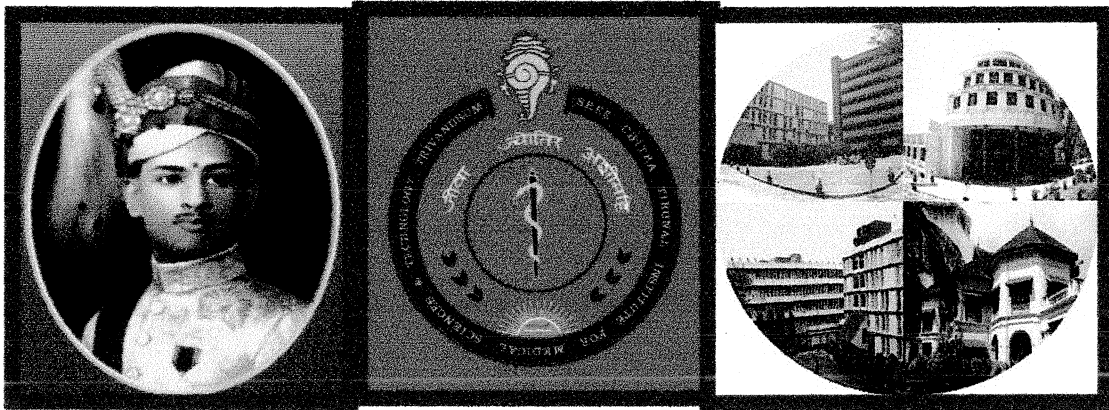


**SREE CHITRA TIRUNAL INSTITUTE OF
MEDICAL SCIENCES & TECHNOLOGY
TRIVANDRUM -695011**



DIPLOMA IN ADVANCED MEDICAL IMAGING TECHNOLOGY

Work book submitted by:

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Place: *Thiruvananthapuram*
Date: *30-11-2006.*

2005-2006

SREE CHITRA TIRUNAL INSTITUTE FOR MEDICAL SCIENCES AND TECHNOLOGY



CERTIFICATE

I Mr. GEO JOSEPH here by declare that I have actually performed all the procedures listed under the course of diploma in advanced imaging technology (DAMIT)

Place : Thiruvananthapuram.

Date : 30-11-2006.

Signature *Geo Joseph*
name GEO JOSEPH.

Geo Joseph
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Signature,

Head of the Dept
Dept .of
Imaging science
&
interventional radiology



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At last, I would like to acknowledge my sincere thanks to PG residents, senior and junior DAMITS for their co-operation at work place and in studies.

PREFACE

This work book, I have done as part of my training in the dept of radiology for diploma in Advanced Medical Imaging Technology (DAMIT) course includes brief details of the equipment used in the Dept, basic physics and working involved with the equipments, the routine protocols and the procedures followed in our different labs, number of cases which I have individually done in X-RAY,CR,CT, MRI,PACS&3D WORKSTATION, and the cases which I have assisted in Neuro and Cardiac Cath Lab, I also have included the seminars and projects I have done .

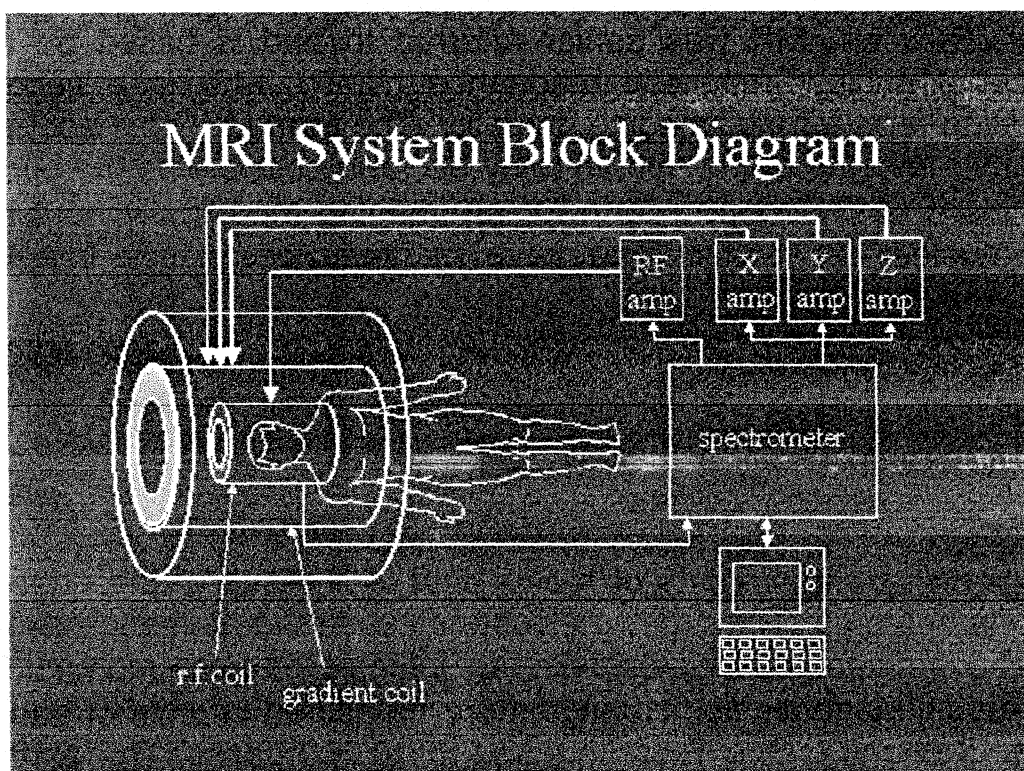
DAMIT is a two years full time residential programme in advanced medical imaging technology for qualified radiographers to excel and learn the newer techniques in medical imaging. Selection is done by a national level entrance examination. At present institute offers 3 seats.

The students are posted in the department of radiology equipped with all modern medical imaging facilities-State of art and top of the line-MRI system, Spiral CT system, DSA suit, Colour Doppler ultra sound scanner and a radiology network with a central workstation with added 3D software and the division of Interventional Radiology make it a distinguished Radiology Dept .The course schedule contains theory classes, practical training, seminar presentations & projects. Diploma is awarded after successful completion of 2 Year term based on a written examination with viva-voce and internal assessment.

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MAGNETISM



History of Magnetic Resonance

In 1946, Felix Bloch and Edward Purcell independently discovered that a magnetically energized substance bombarded with RF emitted a "tone" similar to a tuning fork. They found that the nuclei of different atoms absorbed radio waves at different frequencies. In 1952, Bloch and Purcell received the Nobel Prize for their discovery of what was referred to as Nuclear Magnetic Resonance (NMR), eventually to be known as Magnetic Resonance (MRI). In 1970, the medical imaging world significantly changed with the contributions of Dr. Raymond Damadian. Dr. Damadian discovered that the structure and abundance of water in the human body was the key to MR imaging, and that the water (hydrogen) emitted a signal that was both detectable and recordable. Dr. Damadian and his team spent the next seven years

diligently designing and creating the first MRI scanner for medical imaging of the human body. It was Paul Lauterbur, however, who implemented the concept of tri-plane gradients used for exciting selective areas of the body (G_x , G_y , and G_z). Felix Bloch

Felix Bloch MR was earlier referred to and is based on the physics of nuclear magnetic resonance (NMR). MRI is a second generation term for this imaging modality. The term MRI was adopted soon after imaging of the human body became possible as the public could more easily adopt a term for an imaging modality without the word "nuclear" in it. Thus we have magnetic resonance imaging (MRI).

- Magnetism
- Resonance and RF
- Relaxation

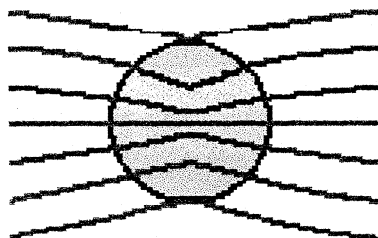
Magnetism

Magnetism is a property of matter that is a result of the orbiting electrons in atoms. The orbiting electrons cause the atoms to have a magnetic moment associated with an intrinsic angular momentum called 'spin'. Magnetic field strengths are measured in units of gauss (G) and Tesla (T). One Tesla is equal to 10,000 gauss. The earth's magnetic field is about 0.5 gauss. The strength of electromagnets used to pick up cars in junk yards is about the field strength of MRI machines (1.5-2.0T). Four terms describing the magnetic properties of materials, used in MRI. These terms are ferromagnetism, paramagnetism, superparamagnetism, and diamagnetism

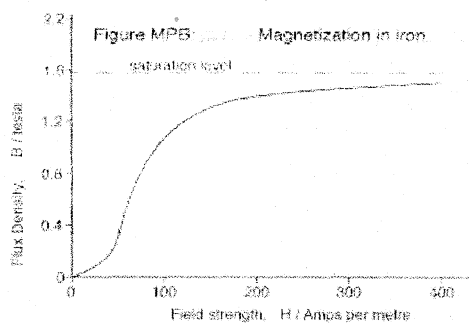
- Ferromagnetism
- Paramagnetism
- Superparamagnetism
- Diamagnetism

Ferromagnetism

Ferromagnetic Material

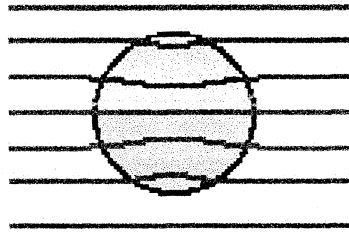


Ferromagnetic materials generally contain iron, nickel, or cobalt. These materials include magnets, and various objects one might find in a patient, such as aneurysm clips, parts of pacemakers, shrapnel, etc. These materials have a large positive magnetic susceptibility, i.e., when placed in a magnet field, the field strength is much stronger inside the material than outside. Ferromagnetic materials are also characterized by being made up of clusters of 10^{17} to 10^{21} atoms called magnetic domains, that all have their magnetic moments pointing in the same direction. The moments of the domains is random in unmagnetized materials, and point in the same direction in magnetized materials. The figure to the above illustrates the effect of a ferromagnetic material (grey circle) on the magnetic field flux lines (blue). The ability to remain magnetized when an external magnetic field is removed is a distinguishing factor compared to paramagnetic, superparamagnetic, and diamagnetic materials. On MR images, these materials cause susceptibility artifacts characterized by loss of signal and spatial distortion. This can occur with even fragments too small to be seen on plain x-ray. This is a common finding in a cervical spine MRI post anterior fusion.



Paramagnetism

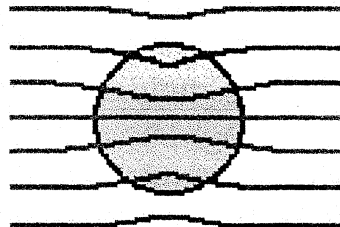
Paramagnetic Material



Paramagnetic materials include oxygen and ions of various metals like Fe, Mg, and Gd. These ions have unpaired electrons, resulting in a positive magnetic susceptibility. The magnitude of this susceptibility is less than one one-thousands of that of ferromagnetic materials. The effect on MRI is increase in the T1 and T2 relaxation rates (decrease in the T1 and T2 times).

Superparamagnetism

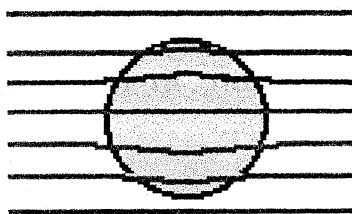
Superparamagnetic Material



Superparamagnetic materials consist of individual domains of elements that have ferromagnetic properties in bulk. Their magnetic susceptibility is between that of ferromagnetic and paramagnetic materials. The figure to the left illustrates the effect of a superparamagnetic material (grey circle) on the magnetic field flux lines (blue). Examples of a superparamagnetic materials include iron containing contrast agents for bowel, liver, and lymph node imaging.

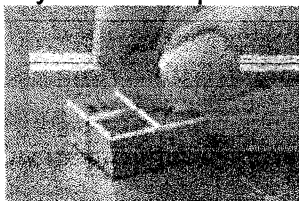
Diamagnetism

Diamagnetic Material



Diamagnetism is a very weak form of magnetism that is only exhibited in the presence of an external magnetic field. It is the result of changes in the orbital motion of electrons due to the external magnetic field. The induced magnetic moment is very small and in a direction opposite to that of the applied field. When placed between the poles of a strong electromagnet, diamagnetic materials are attracted towards regions where the magnetic field is weak. Diamagnetism is found in all materials; however, because it is so weak it can only be observed in materials that do not exhibit other forms of magnetism. Also, diamagnetism is found in elements with paired electrons. Oxygen was once thought to be diamagnetic, but a new revised molecular orbital (MO) model confirmed oxygen's paramagnetic nature.

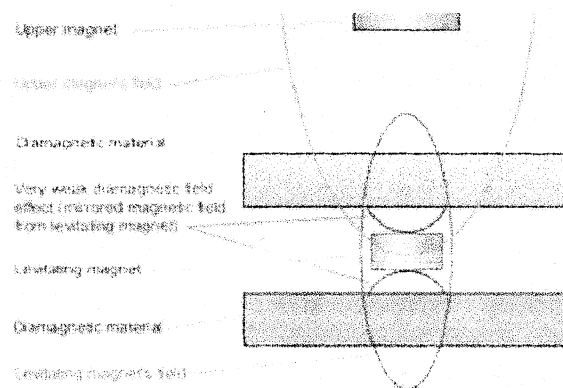
An exception to the "weak" nature of diamagnetism occurs with the rather large number of materials that become superconducting, something that usually happens at lowered temperatures. Superconductors are perfect diamagnets and when placed in an external magnetic field expel the field lines from their interiors (depending on field intensity and temperature).



Superconductors also have zero electrical resistance, a consequence of their diamagnetism. Superconducting structures have been known to tear themselves apart with astonishing force in their attempt to escape an external field. Superconducting magnets are the major

component of most magnetic resonance imaging systems, perhaps the only important application of diamagnetism

Perhaps the substance that displays the strongest diamagnetism is bismuth, used as an alternative to lead shot in shotgun shells. Melting down bismuth and then molding it is a very efficient way of capturing the diamagnetic properties. Pyrolytic Graphite is stronger than Bismuth in only one direction; in the other two directions it is still diamagnetic, but it is weaker

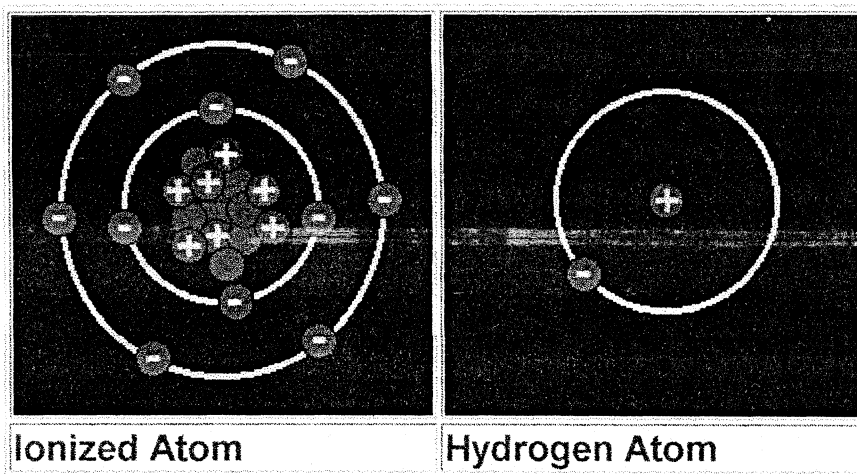


Super Conductors

Superparamagnetic materials consist of individual domains of elements that have ferromagnetic properties in bulk. Their magnetic susceptibility is between that of ferromagnetic and paramagnetic materials. The figure to the left illustrates the effect of a super paramagnetic material (grey circle) on the magnetic field flux lines (blue). Examples of a super paramagnetic materials include iron containing contrast agents for bowel, liver, and lymph node imaging. Diamagnetic materials have no intrinsic atomic magnetic moment, but when placed in a magnetic field weakly repel the field, resulting in a small negative magnetic susceptibility. Materials like water, copper, nitrogen, barium sulfate, and most tissues are diamagnetic. The figure above illustrates the effect of a diamagnetic material (grey circle) on the magnetic field flux lines (blue). The weak negative magnetic susceptibility contributes to the loss of signal seen in bowel on MRI after administration of barium sulfate suspensions.

The Atom

In order to understand MRI, it is necessary to understand the properties of atoms. Atoms consist of a dense nucleus surrounded by orbiting electrons. The nucleus of most atoms is made up of positively charged particles called protons, and neutrally charged particles called neutrons. The nucleus of an atom is always positively charged due to the positive protons. The electrons orbiting around the nucleus are extremely small negatively charged particles which balance the positive charge of the nucleus. Chemical properties of elements vary based on the electrical charge of the atoms. An atom with the same number of protons and electrons is chemically neutral and stable. Often, electrons are added or knocked out of the orbits around nuclei, which alters the charge of atom thus creating a positively or negatively charged particle. An atom which is positively or negatively charged is referred to as ionized



The simple nucleus of the **hydrogen atom** consists of one **proton**, and no **neutrons**. The hydrogen atom has a positive charge and an atomic number of 1 due to the presence of only one proton in its nucleus. For the purposes of MR, the hydrogen atom is referred to as a **proton**. The abundance of the hydrogen atoms in the human body, and the large **magnetic moment** (discussed below) created by the single proton in the nucleus of the atom, make hydrogen atoms extremely sensitive to magnetic resonance. Based on these facts, we will concentrate on the **hydrogen atom** for the duration of this section. Hydrogen has the simplest atomic structure compared to all other elements. There is an abundance of hydrogen in the human

body. Approximately 70% of the body is made up of water which contains two hydrogen atoms and one oxygen atom. It is the hydrogen atoms that are focused on to produce an MR image.

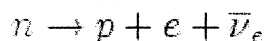
The proton

With spin and parity $1/2^+$, charge +1, and mass of 938 MeV, the proton is the nucleus of a hydrogen atom ($^1\text{H}_1$). It has a magnetic moment of 2.79 nuclear magnetons. The electric dipole moment is consistent with zero; the bound on it is that it is less than 0.54×10^{-23} e-cm.

In some speculative grand unified theories it may decay. The half-life for this decay has been limited to be greater than 2.1×10^{29} years. The charge radius is measured mainly through elastic electron-proton scattering and is 0.870 fm. For specific decay modes, into antilepton or lepton and a meson, the bound is often better than 10^{32} years. The proton is therefore taken to be a stable particle, and baryon number is assumed to be conserved.

The neutron

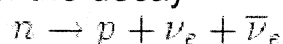
The neutron has no charge, has spin and parity of $1/2^+$, and mass of 940 MeV. The most precise measurements of its decay lifetime are mainly from traps of various kinds and in beams. The lifetime of a neutron is 886 s. It has the weak decay



Its magnetic moment is -1.91 nuclear magnetons. Both time reversal and parity invariance of the strong interactions implies that the neutron's electric dipole moment must be zero; the current observational bound is that it is less than 0.63×10^{-23} e-cm. The mean-square charge radius related to the scattering length measured in low energy electron-neutron scattering for the neutron is -0.116 fm^2 .

Violation of baryon number conservation may give rise to oscillations between the neutron and antineutron, through processes which change **B** by two units. Using free neutrons from nuclear reactors, as well as neutrons bound inside nuclei, the mean time for these transitions is found to be greater than 1.3×10^8 seconds. The much poorer bound, as compared to protons, is related to the difficulty of the observations.

A limit on electric charge non-conservation comes from the observed lack of the decay



The observations which limit the branching fraction of the neutron in this decay channel to less than 8×10^{-27} are all done looking for appropriate decays of nuclei ($A \rightarrow A$ and $Z \rightarrow Z+1$).

Spin

The moving (**spinning**) hydrogen protons create a **magnetic field**, and thus perform as a tiny magnet with a north and a south pole. Since there are two magnetic poles, the protons are referred to **magnet dipoles**. Based on the laws of electromagnetism, any electrically charged particle which moves creates a magnetic field called a **magnetic moment**. This is the property that allows hydrogen protons to behave predictably within an external magnetic field. The motion or the "spin" of the hydrogen atoms can be described as a random combination of the spinning of a top, the spin of a bowling ball and the rotation of the earth around its axis. Understanding the magnetic moment of the hydrogen protons, will lead to an understanding of the **alignment** of the protons within a magnet.

Nuclear spin and magnets

Electrons, neutrons and protons, the three particles which constitute an atom, have an intrinsic property called spin. This spin is defined by the fourth quantum number for any given wave function obtained by solving relativistic form of the Schrödinger equation (SE). It represents a general property of particles which we can describe using the properties of electrons. Electrons flowing around a coil generate a magnetic field in a given direction; this property is what makes electric motors work. In much the same way electrons in atoms circulate around the nucleus, generating a magnetic field. This generated field has an angular momentum associated with it. It so turns out that there is also an angular momentum with the electron particle itself, denoted the spin, and this gives rise to the spin quantum number, m_s .

Spin angular momentum is quantized and can take different integer or half-integer values depending on what system is under study. If we solve the relativistic SE for the electron we get the values $+\frac{1}{2}$ and $-\frac{1}{2}$. Since the Pauli principle states that no two fermions can have the same quantum number, it is why only two electrons, paired

antiparallel (one with positive spin and one negative with negative spin), can appear in a single atomic orbital.

Like the electron, protons and neutrons also have a spin angular momentum which can take values of $+\frac{1}{2}$ and $-\frac{1}{2}$. In the atomic nucleus, protons can pair with other antiparallel protons much in the same way that electrons pair in a chemical bond. Neutrons do the same. Paired particles, with one positive and one negative spin, thus have a net spin of zero "0". We can see that a nucleus with unpaired protons and neutrons will have an overall spin, with the number unpaired contributing $\frac{1}{2}$ to the overall nuclear spin quantum number, I . When this is larger than zero, a nucleus will have a spin angular momentum and an associated magnetic moment, μ , dependent on the direction of the spin. *It is this magnetic moment that we manipulate in modern NMR experiments.*

It is worth noting here that nuclei can have more than one unpaired proton and one unpaired neutron, much in the same way that electronic structures in transition metals can have many unpaired spins. For example ^{27}Al has an overall spin $I=5/2$.

A technique related to nuclear magnetic resonance is electron spin resonance that exploits the spin of electrons instead of nuclei. The principles are otherwise similar.

Values of spin angular momentum

The spin angular momentum of a nucleus can take ranges from $+I$ to $-I$ in integral steps. This value is known as the magnetic quantum number, m . For any given nucleus, there is a total $(2I+1)$ angular momentum states. Spin angular momentum is a vector quantity. The z component of which, denoted I_z , is quantised:

$$I_z = mh/2\pi$$

where h is Planck's constant.

The resultant magnetic moment of this nucleus is intrinsically connected with its spin angular momentum. In the absence of any external effects the magnetic moment of a spin $\frac{1}{2}$ nucleus lies approximately 52.3° from the angular momentum axis or 127.7° for the opposing spin. This magnetic moment is intrinsically related to I with a proportionality constant γ , called the gyromagnetic ratio:

$$\mu = \gamma I$$

Spin behaviour in a magnetic field

Consider the case of nuclei which have a spin of a half, like ^1H , ^{13}C or ^{19}F . The nucleus thus has two possible magnetic moments it could take, often referred to as up or down, $+1/2$ or $-1/2$, which are also called the spin states α and β . The energies of each state are degenerate - that is to say that they are the same. The effect is that the number of atoms, their *population*, in the up or α state is the same as the number of atoms in the β state.

If a nucleus is placed in a magnetic field, the angular momentum axis coincides with the field direction. The resultant magnetic momenta, space quantised from the angular momentum axis, *no longer have the same energy* since one state has a z-component aligned with an external field and are lower in energy (positive l values) and the other opposes the external field and is higher in energy. This causes a population bias toward the lower energy states.

The energy of a magnetic moment μ when in a magnetic field B_0 , the zero subscript is used to distinguish this magnetic field from any other applied field, is the negative scalar product of the vectors:

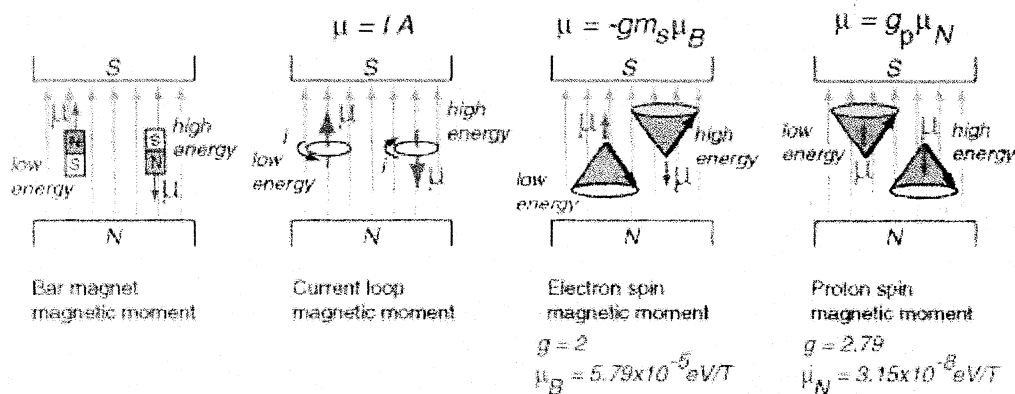
$$E = -\mu_z B_0$$

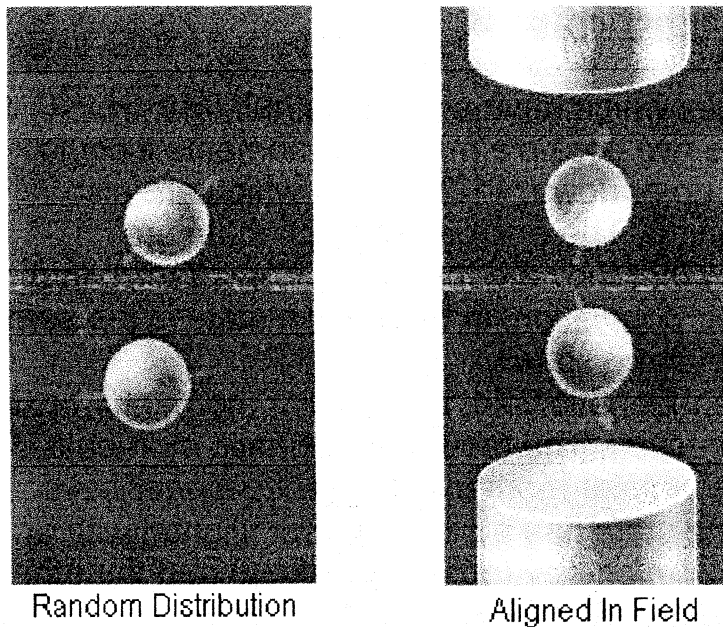
We've already defined $\mu_z = \gamma I_z$. So placing this in the above equation we get:

$$E = -m\hbar\gamma B_0 / 2\pi$$

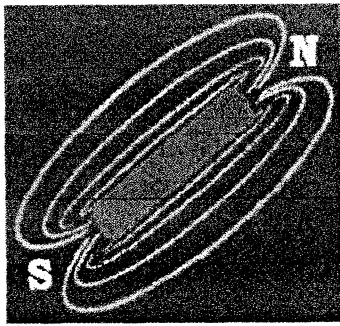
Where γ is the gyromagnetic ratio of the nucleus being scanned

Alignment of protons in magnet

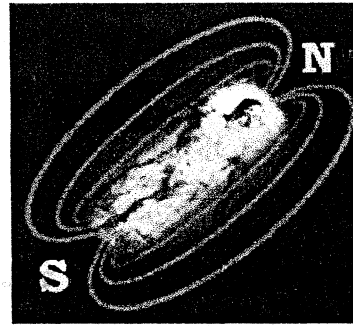




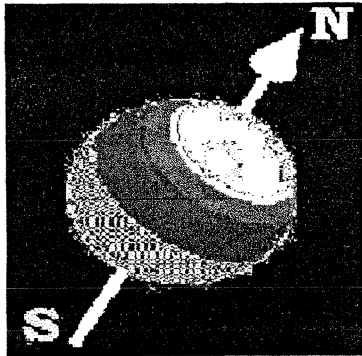
Hydrogen nuclei magnetic moments are randomly oriented in the absence of an external magnetic field and are considered to have a **net magnetization** of zero. Once hydrogen protons are placed in the presence of an external magnetic field, they align themselves in one of two directions, **parallel** or **anti-parallel** to the net magnetic field which is commonly referred to as the vector B_0 . The strength of the external magnetic field and the thermal energy of the atoms are the factors which affect the direction of alignment of the hydrogen protons. The high energy protons are strong enough to align themselves against or anti-parallel to the magnetic field, whereas the lower energy protons will align themselves with or parallel to the magnetic field. As the magnetic field increases, there are fewer protons which are strong enough to align anti-parallel to the magnetic field. There are always a larger number of protons aligned parallel with the magnetic field (B_0), so once the parallel and anti-parallel protons cancel each other out, only the small number of low energy protons left aligned with the magnetic field create the overall net magnetization of the patient's body. Even though there are relatively only a few protons creating the net magnetization of the patient's body, this **difference** is all that counts. The magnetic moments of these protons are added together and are referred to as **net magnetization vector (NMV)** or the symbol '**M**'.



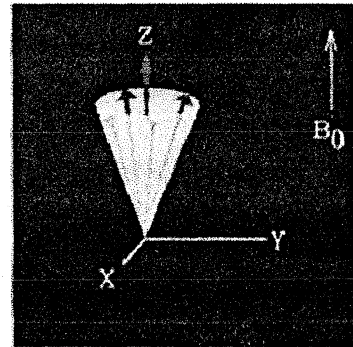
Bar magnet



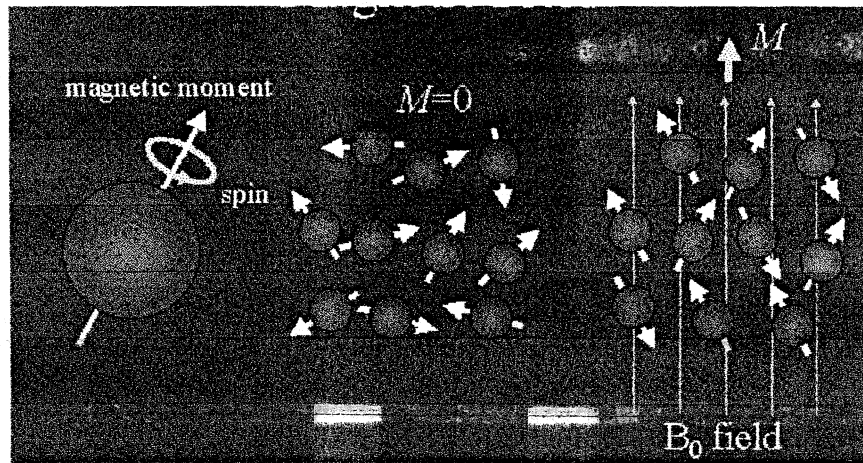
Human magnet



Spinning proton



Precession



Aliengment of spins in a magnetic field

The Larmor Frequency

Tesla (T) or **gauss** is the measure of strength of the magnetic field. One Tesla is equivalent to 10,000 gauss, and is about 20,000 times stronger than the earth's magnetic field. All protons precess at the same frequency within a magnetic field. The **gyro-magnetic ratio** of hydrogen is 42.57 MHz/Tesla. The gyro-magnetic ratio is different for each nucleus of different atoms. The frequency is determined by the gyro-magnetic ratio of atoms and the strength of the magnetic field. The **Larmor** equation is important because it is the frequency at which the nucleus will absorb energy. The absorption of that energy will cause the proton to alter its alignment. In MR imaging, the energy that is transferred is radio frequency waves (RF) and ranges from 1-100 MHz. The Larmor equation governs the value of the precessional frequency, and is follows:

$$\text{Precessional frequency } (\omega_0) = B_0 \times \gamma$$

B_0 = the magnetic field strength of the magnet

γ = the gyro-magnetic ratio.

At 1 T the precessional frequency of hydrogen is 42.57 MHz (1 T x 42.57 MHz)

At 1.5 T the precessional frequency of hydrogen is 63.86 MHz (1.5 T x 42.57 MHz)

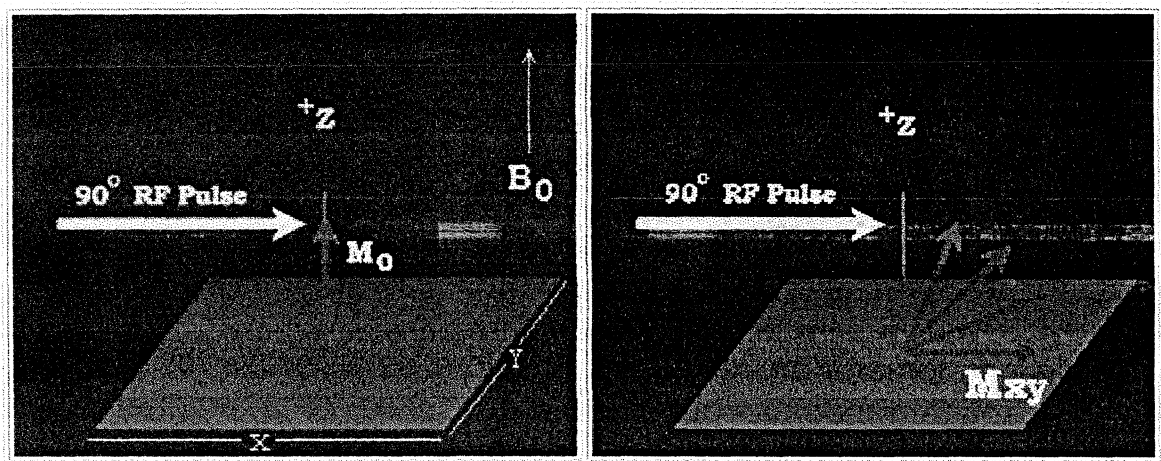
The stronger the magnetic field, the higher the **precessional frequency**. If an RF pulse at the Larmor frequency is applied to the nucleus of an atom, the protons will alter their alignment from the direction of the main magnetic field to the direction opposite the main magnetic field. As the proton tries to realign with the main magnetic field, it will emit energy at the frequency of the Larmor frequency.

Resonance is referred to as the property of an atom to absorb energy only at the Larmor frequency. This is the basis of MR. An atom will only absorb external energy if that energy is delivered at precisely its resonant frequency. The energy must also be delivered at 90° to the net magnetic vector (NMV) and main magnetic field (B_0).

Otherwise, no energy will be absorbed, resonance will not have occurred and an image cannot be created. **Excitation** occurs when the proton absorbs the applied energy or resonates. As resonance occurs and the NMV moves out of alignment with the B_0 to a pre-specified angle. The deflection of the magnetization or total angle created after the end of the RF pulse is referred to as the **flip angle**.

Longitudinal and Transverse Magnetization

The stronger the RF energy applied to the protons, the greater the angle of deflection for the magnetization. The two most common flip angles in MR are 90° and 180° . A 90° pulse will flip the magnetization into the **x-y plane (M_{xy})**. A 180° pulse will flip the magnetization through the x-y plane and into the opposite direction of B_0 .



When a 90° pulse is applied and the protons are given enough energy to be flipped into the x-y plane, the net magnetization vector is now in the transverse plane. B_0 or z-axis is now referred to as the longitudinal plane. The protons are now rotating in the transverse plane at the Larmor frequency. As well as flipping into the transverse plane, the protons also begin rotating in phase with each other. When resonance occurs, all the magnetic moments move into the same path or all flip the same number of degrees, and they all precess in phase with other.

With the net magnetization in the transverse plane (created with a 90° flip angle), and a receiver coil or antenna in the transverse plane, a voltage is induced within the receiver coil. This oscillating signal voltage over time is the MR signal. The magnitude of the signal is dependent on the magnetization present in the transverse plane. At the termination of the RF, the freely precessing protons in the transverse plane (**M_{xy}**) give up energy (RF) in order to try to realign with B_0 . As the transverse magnetization starts to **decay** due to the loss of phase coherence, the protons eventually realign with B_0 . This **signal** produced by the decay of transverse magnetization is called **free induction decay (FID)**. The amplitude of the FID signal becomes smaller over time as net magnetization returns to equilibrium. Simultaneously, the longitudinal magnetization begins to recover and return to B_0 to a state of equilibrium just as if nothing had occurred.

After the external RF signal is turned off, two phenomenons simultaneously occur.

- Longitudinal magnetization gradually increases and is called **T1 recovery**
- Transverse magnetization gradually decreases and is called **T2 decay**

These phenomena are discussed in more detail below. In summary, atoms rotate randomly outside the presence of a magnetic field. When in the presence of an external magnetic field, the atoms align either with or opposed to the main magnetic field. The parallel and anti-parallel protons cancel each other out, leaving a relatively small number of protons aligned with the main magnetic field. As an RF signal is applied at the Larmor frequency, the individual protons resonate, or absorb the applied energy, and precess in phase. Depending on the strength of the applied energy, the protons will flip

into the x-y plane (transverse magnetization), or exactly the opposite direction of the main magnetic field. The transverse magnetization induces a voltage in an antenna or receiver coil which will be eventually become the MR signal. As the RF is turned off, the protons dephase and lose their coherence as they try to realign with B_0 . Two phenomenons occur simultaneously. Transverse magnetization

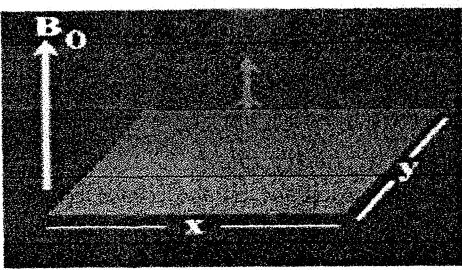
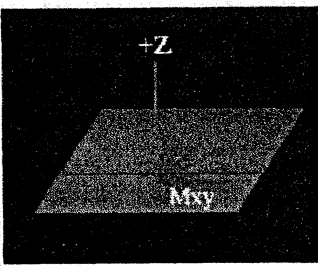
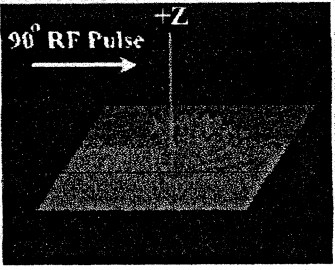
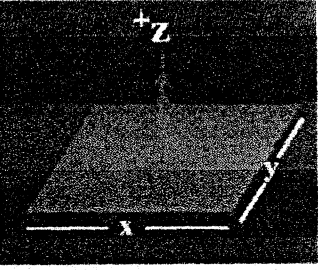
resonate, or absorb the applied energy, and precess in phase. Depending on the strength of the applied energy, the protons will flip into the x-y plane (transverse magnetization), or exactly the opposite direction of the main magnetic field. The transverse magnetization induces a voltage in an antenna or receiver coil which will be eventually become the MR signal. As the RF is turned off, the protons dephase and lose their coherence as they try to realign with B_0 . Two phenomenons occur simultaneously. Transverse magnetization decreases (T2 decay), while longitudinal magnetization increases (T1 recovery).

Longitudinal and Transverse Relaxation

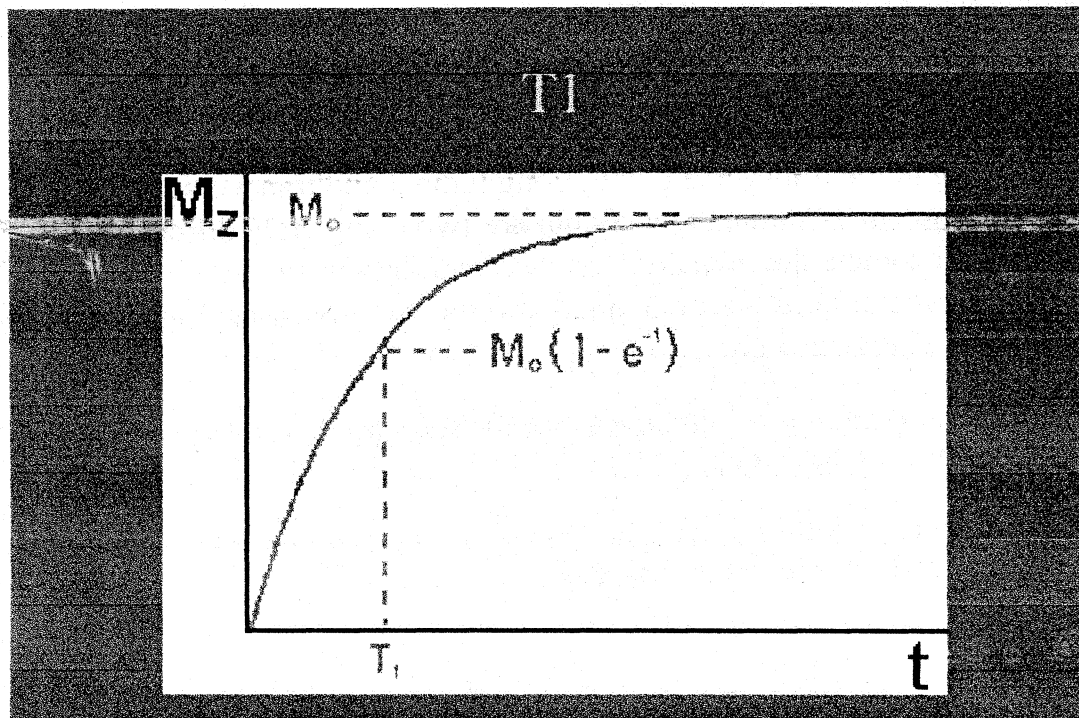
Relaxation is a very important process in MR imaging as it determines the type of signal obtained greatly impacting the type of image generated. When the RF pulse is terminated, the net magnetic vector (NMV) is once again influenced by the main magnetic field (B_0) and tries to re-align with it along the longitudinal axis. Protons attempt to return from a state of non-equilibrium to a state of equilibrium. As the NMV gives up its absorbed RF energy while trying to return to B_0 , the process of relaxation occurs. As relaxation is occurring, magnetization is recovering in the longitudinal plane while decaying in the transverse plane. Longitudinal and transverse magnetization occur at simultaneously but are two completely different processes. To examine the relaxation phenomena further, the net magnetization vector needs to be divided into its two processes; longitudinal and transverse magnetization.

Longitudinal relaxation is the return of longitudinal magnetization to equilibrium (B_0) and is termed **T1 recovery**.

- **Transverse relaxation** is the return of transverse magnetization to equilibrium and is termed **T2 decay**.

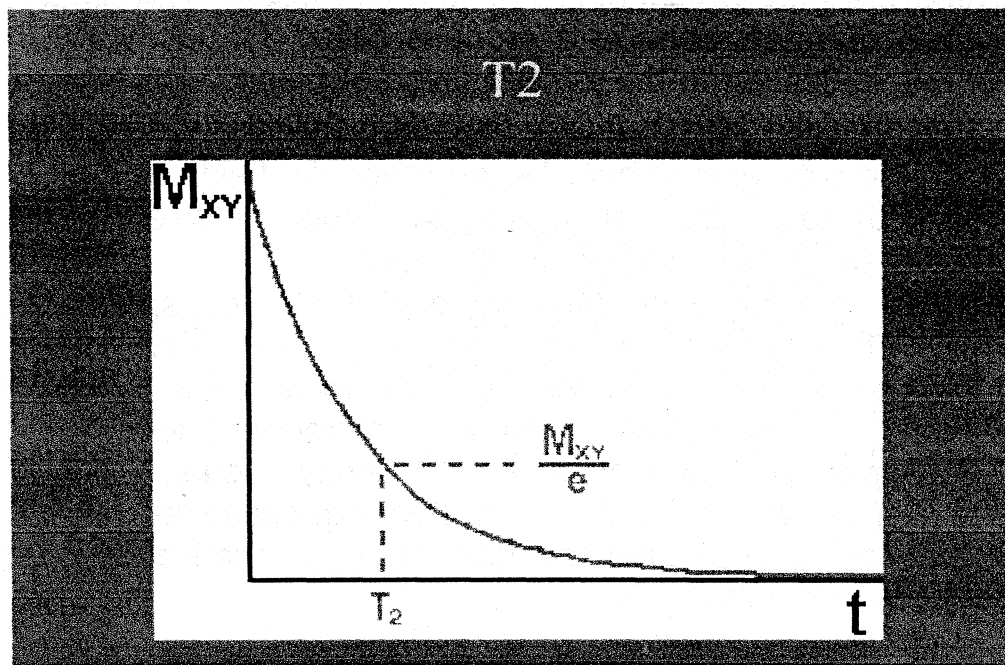
			
Longitudinal Magnetization	90 Degree RF Pulse	T2 Dephasing	Longitudinal Relaxation - T1

Longitudinal Relaxation



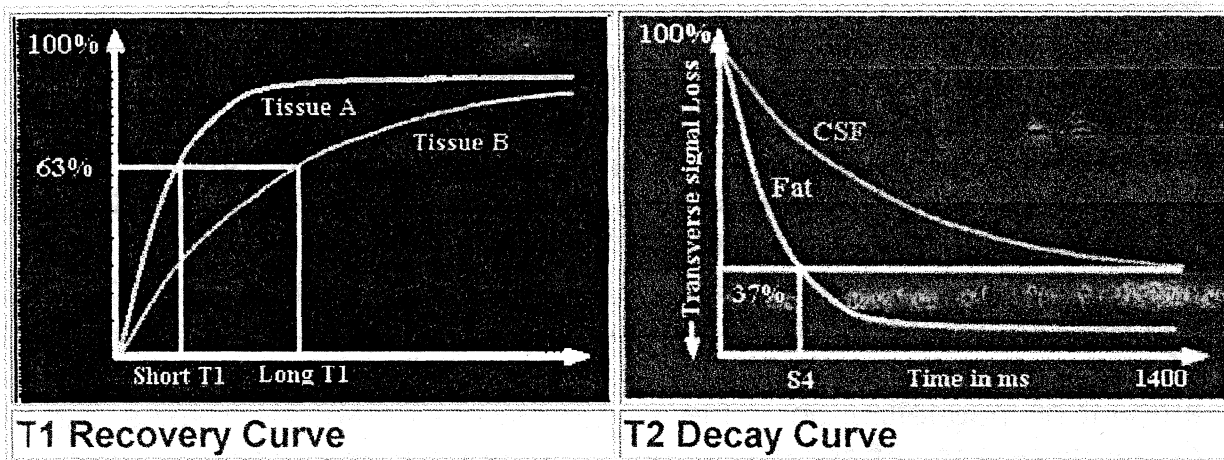
Longitudinal relaxation is the return of longitudinal magnetization to equilibrium (B_0) and is termed T1 recovery. As the hydrogen atoms release their energy they previously absorbed to the surrounding tissue (lattice), in their attempt to realign with the main magnetic field, T1 relaxation or T1 recovery occurs. This phenomenon is sometimes

Transverse Relaxation



As longitudinal magnetization is occurring, it is accompanied by transverse relaxation. **Transverse relaxation** is the return of transverse magnetization to equilibrium. Although these two processes occur at the same time, they are quite different and thus contribute significantly different information to the resulting MR image. Unlike longitudinal relaxation, transverse relaxation is not a process of dissipation or absorption of energy into tissue. The **decay of transverse magnetization** is a process called T2 decay. As the amount of magnetization in the x-y (transverse) plane decreases, T2 decay increases. After an RF pulse, hydrogen nuclei are spinning in unison or in-phase with each other. As the magnetic fields of all the nuclei interact with each other, energy is exchanged between those nuclei. The nuclei which began spinning "in-phase", lose their phase coherence or dephase over time and spin in a random fashion. This process results in an exponential decrease or decay in transverse magnetization. Because T2 decay is the result of the exchange of energy between spinning hydrogen nuclei, it is referred to as "**spin-spin**" relaxation. As T2 decay occurs, the MR signal dies out. The rate of T2 decay is also expressed as a time constant. T2 decay occurs when the transverse magnetization has decreased to 37% of its initial value

image. Unlike longitudinal relaxation, transverse relaxation is not a process of dissipation or absorption of energy into tissue. The **decay of transverse magnetization** is a process called T2 decay. As the amount of magnetization in the x-y (transverse) plane decreases, T2 decay increases. After an RF pulse, hydrogen nuclei are spinning in unison or in-phase with each other. As the magnetic fields of all the nuclei interact with each other, energy is exchanged between those nuclei. The nuclei which began spinning "in-phase", lose their phase coherence or dephase over time and spin in a random fashion. This process results in an exponential decrease or decay in transverse magnetization. Because T2 decay is the result of the exchange of energy between spinning hydrogen nuclei, it is referred to as "**spin-spin**" relaxation. As T2 decay occurs, the MR signal dies out. The rate of T2 decay is also expressed as a time constant. T2 decay occurs when the transverse magnetization has decreased to **37%** of its initial value



Longitudinal relaxation is a regrowth or an increase in value, whereas transverse relaxation is a decrease or decay. Although these two processes occur together, T2 decay almost always occurs more rapidly than the regrowth of longitudinal magnetization.

Tissue Contrast

Due to the T1 and T2 relaxation properties, we can differentiate between various tissues in the body. As well as T1 and T2 contrast in tissues, proton density can also be measured. **Proton density** is measured by the **number of protons per unit of tissue**. Various

tissues have different T1 and T2 values. These T1 and T2 values significantly influence the type of signal generated during MRI and thus contribute greatly to the MR image. **Tissue contrast** is affected by not only the T1 and T2 values of specific tissues, but the differences in the magnetic field strength, temperature changes and many other factors. It is not imperative that one memorize the absolute T1 and T2 values in different tissues, but being aware of the values may make a difference when the technologist is programming values for pulse sequences.

Fat Versus Water

Due to the slow molecular motion of fat nuclei, longitudinal relaxation occurs rather rapidly and longitudinal magnetization is regained quickly. The net magnetic vector realigns with B_0 leading to a short T1 time for fat. Water is not as efficient as fat in T1 recovery due to the high mobility of the water molecules. Water nuclei do not give up their energy to the lattice (surrounding tissue) as quickly as fat, and therefore take longer to regain longitudinal magnetization resulting in a long T1 time.

As we know, T2 decay is dependent on the interaction of nuclei and the exchanging of energy with near by nuclei. Fat has a very efficient energy exchange and therefore has a relatively short T2. Water is less efficient than fat in the exchange of energy, and therefore has a long T2.

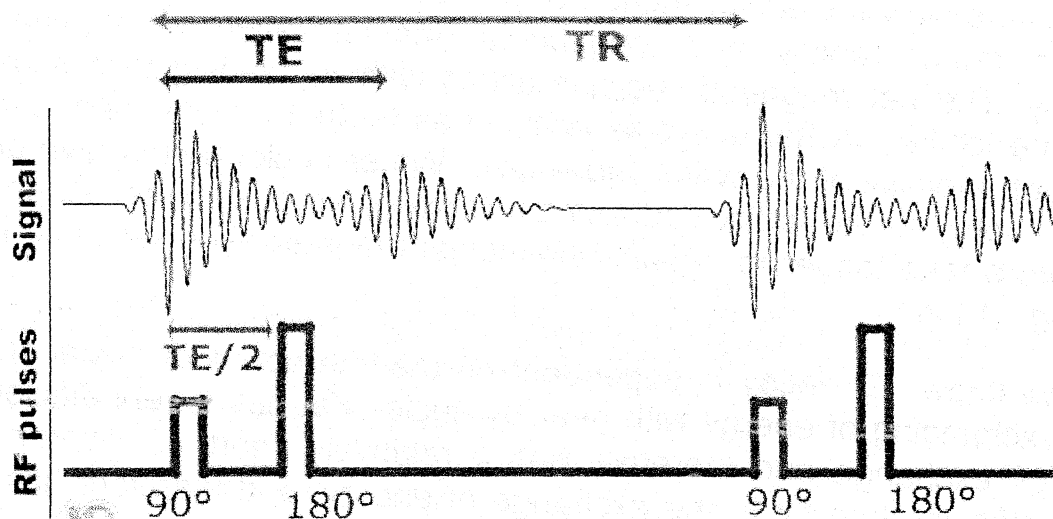
T1 and T2 CONSTANTS

	<i>T1 Constants at 1.5 T Controlled by TR</i>	<i>T2 Constants at 1.5 T Controlled by TE</i>
Fat		85
Muscle	860	45
White matter	780	90

Gray matter	920	100
CSF	3000	1400

TR and TE are parameters controlled by the operator and are usually measured in milliseconds.

- TR stands for repetition time, or the elapsed time between successive RF excitation pulses.
- TE stands for echo delay time, or the time interval between the RF pulse and the measurement of the first echo.



The T1 constants above will indicate how quickly the spinning nuclei will emit their absorbed RF into the surrounding tissue. The T2 constants above will indicate how quickly the spinning nuclei will decay to 37% of the initial transverse magnetization.

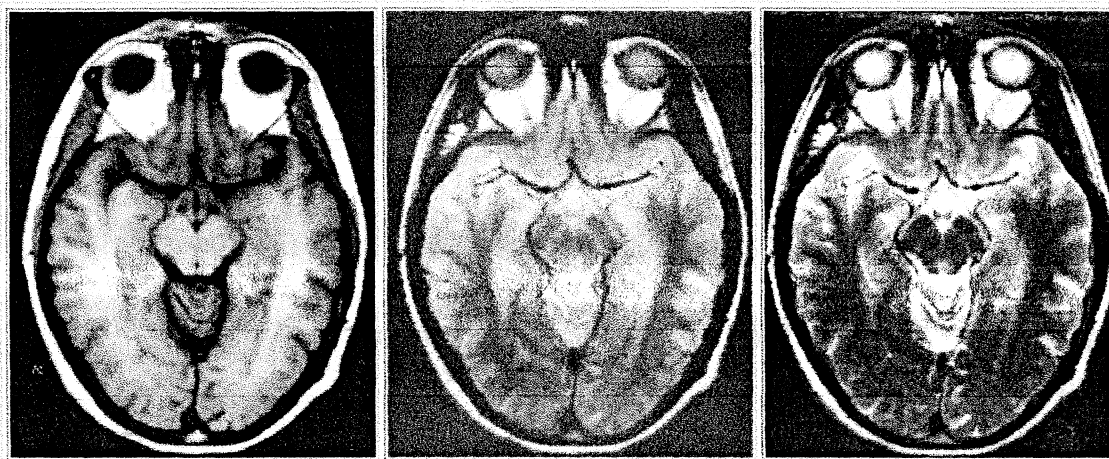
T1, T2 and Proton Density Contrast

Fat has a shorter T1 time than water, therefore the fat vector will realign more quickly with the main magnetic field. It is obvious then that fat has a larger longitudinal component than water. After a 90° pulse, the longitudinal magnetization of both fat and water are flipped into the transverse plane. As previously mentioned, fat has a larger longitudinal component prior to an RF pulse, and it has a larger

transverse component after an RF pulse. Due to the larger longitudinal and transverse magnetization, fat has a higher signal and will appear bright on a T1 contrast MR image. Conversely, water has less longitudinal magnetization prior to an RF pulse, therefore less transverse magnetization after an RF pulse yielding low signal appearing dark on a T1 contrast image. Images created with TR's and TE's to enhance T1 contrast are referred to as T1-weighted images.

The previously learned concepts of transverse magnetization apply for T2 contrast. Fat has a shorter T2 time than water and relaxes or decays more readily than water. Since the amount of transverse magnetization in fat is small, fat generates very little signal on a T2 contrast image and appears dark. Water has a very high T2 constant, therefore has very high T2 signal and thus appears bright on a T2 contrast image. Images created with TR's and TE's to enhance T2 contrast are referred to as T2-weighted images.

Proton density contrast is a quantitative summary of the number of protons per unit tissue. The higher the number of protons in a given unit of tissue, the greater the transverse component of magnetization, and the brighter the signal on the proton density contrast image. Conversely the lower the number of protons in a given unit of tissue, the less the transverse magnetization and the darker the signal on the proton density image.



T1 Weighting	Proton Weighting	Density	T2 Weighting
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T1 AND T2 WEIGHTING (Proton weighting imaging)

Most all MR imaging will entail T1 and T2-weighted images among many other types of imaging. T1 and T2 images are the most common contrasts obtained in MRI. T1-weighted images are obtained to compare the T1 differences in tissues or to compare the relaxation rates of the tissue being examined. T2-weighted images are obtained to compare the T2 contrast in tissues and compare the transverse relaxation rates. Parameters are manipulated by the user to obtain the type of image contrast desired

<i>RF SIGNAL INTENSITIES IN TISSUE</i>	
High-Intensity Signal	Low-Intensity Signal
Short T1	Long T1
Long T2	Short T2

As previously stated, TR will control the T1-weighting of an MR image short TR maximizing T1-weighting and a long TR maximizing proton weighting

TE controls the T2-weighting of an MR image with a short TE minimizing T2-weighting and a long TE maximizing T2-weighting.

<i>TYPICAL TE and TR Values</i>		<i>IMAGE WEIGHTING PARAMETERS</i>	
Short TE	10-25 ms	TR	TE

Long TE	60-100+ ms	T1 Weighting	Short	Short
Short TR	100-900 ms	Proton Density Weighting	Long	Short

T1-Weighted Imaging:

Emphasis is made on the differences on longitudinal relaxation rates by utilizing a short TR and short TE. The short TR allows full recovery of tissues with a short T1 (fat) to recover quickly while allowing only partial recovery of tissues with a long TR (CSF). The short TE used to obtain a T1-weighted image will allow minimal loss of transverse magnetization due to T2 relaxation. Basically, short TR increases T1 effects with the short TE minimizes T2 effects of tissue.

T2-Weighted Imaging:

Emphasis is made on the differences on transverse relaxation rates by utilizing a long TR and long TE. The long TR allows tissues to reach complete longitudinal magnetization which will reduce T1 effects. The long TE will allow for the loss of transverse signal enhancing T2 effects. Fluid has a very long T2 and is frequently associated with pathology, so it is important to take advantage of the imaging parameters to accentuate signal differences.

There are many parameters which significantly contribute to the contrast of an image. Most of these parameters are controlled or at least influenced by the technologist creating the images. Care must be taken to create the best possible MR images which will ensure a more accurate diagnosis by the radiologist. Parameters contributing to an MR image should be manipulated only by an experienced MR technologist who is aware of all the factors which are changing during the manipulation

On a macroscopic level, exposure of an object or person to RF radiation at the Larmor frequency, causes the net magnetization to spiral away from the B_0 field. In the rotating frame of reference, the

net magnetization vector rotate from a longitudinal position a distance proportional to the time length of the RF pulse. After a certain length of time, the net magnetization vector rotates 90 degrees and lies in the transverse or x-y plane. It is in this position that the net magnetization can be detected on MRI. The angle that the net magnetization vector rotates is commonly called the 'flip' or 'tip' angle. At angles greater than or less than 90 degrees there will still be a small component of the magnetization that will be in the x-y plane, and therefore be detected.

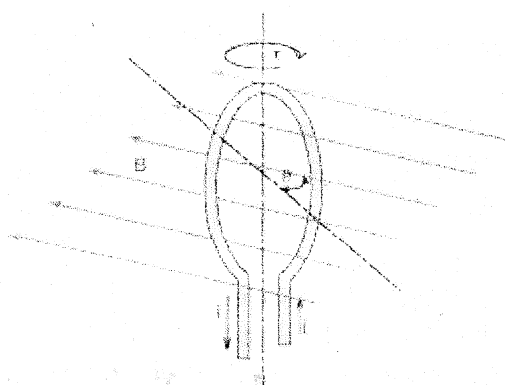
INSTRUMENTATION

This section will cover the following topics:

- Magnets
- RF and Gradient Coils
- Electronics and Data Processing

MAGNET

Fig. MPM The concept of magnetic moment.



Magnetic resonance imaging (MRI) is based on the magnetic resonance phenomenon, and is used for medical diagnostic imaging since ca. 1977 (see also MRI History). The first developed MRI devices were constructed as long narrow

tunnels. In the meantime the magnets became shorter and wider. In addition to this short bore magnet design, open MRI machines were created. MRI machines with open design have commonly either horizontal or vertical opposite installed magnets and obtain more space and air around the patient during the MRI test. The basic hardware components of all MRI systems are the magnet, producing a stable and very intense magnetic field, the gradient coils, creating a variable field and radio frequency (RF) coils which are used to transmit energy and to encode spatial positioning. A computer controls the MRI scanning operation and processes the information.

The range of used field strengths for medical imaging is from 0.15 to 3 T. The open MRI magnets have usually field strength in the range 0.2 Tesla to 0.35 Tesla. The higher field MRI devices are commonly solenoid with short bore superconducting magnets, which provide homogeneous fields of high stability.

There are THREE different types of magnets:

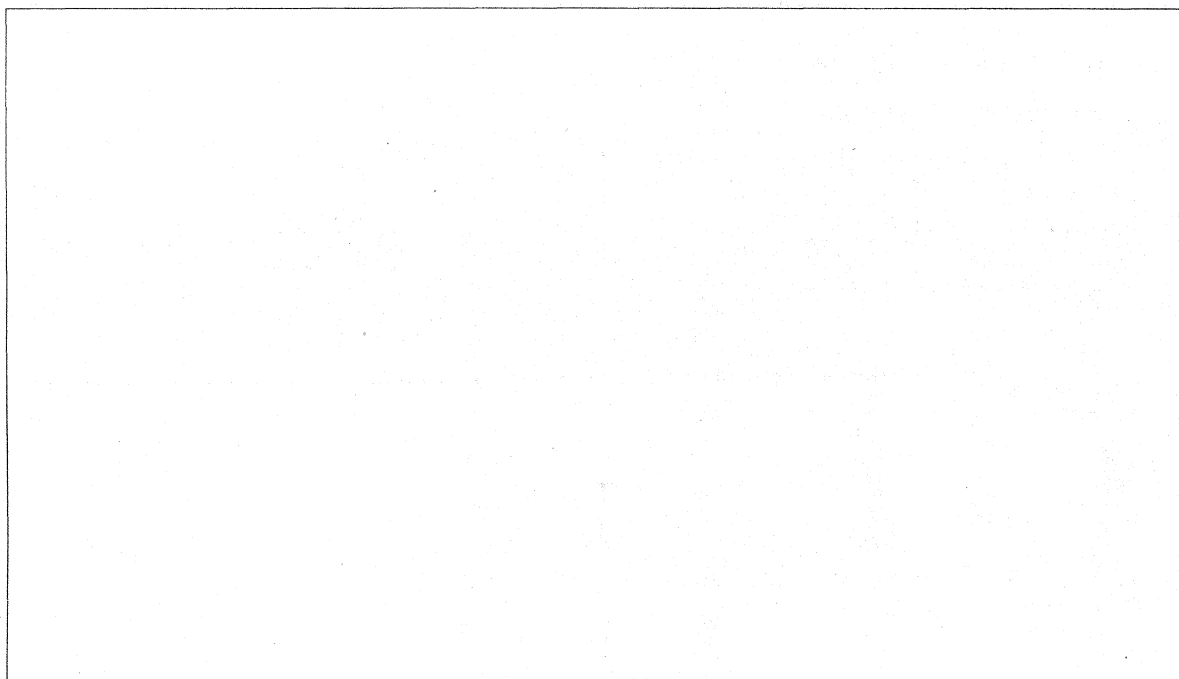
- Resistive Magnet
- Permanent Magnet
- Super conducting magnet



Resistive Magnet

A type of magnet that utilizes the principles of electromagnetism to generate the magnetic field. Typically large current values and significant cooling of the magnet coils is required. The **resistive magnet** does not require cryogen's, but needs a constant power supply to maintain a homogenous magnetic field, and can be quite expensive to maintain.

Resistive magnets fall into two general categories - iron-core and air-core. Iron-core electromagnets provide the advantages of a vertically oriented magnetic field, and a limited fringe field with little, if any, missile effects due to the closed iron-flux return path. Air-core electromagnets exhibit horizontally oriented fields, which have large fringe fields (unless magnetically shielded) and are prone to missile effects. Resistive magnets are typically limited to maximum field strengths of approximately 0.6T.

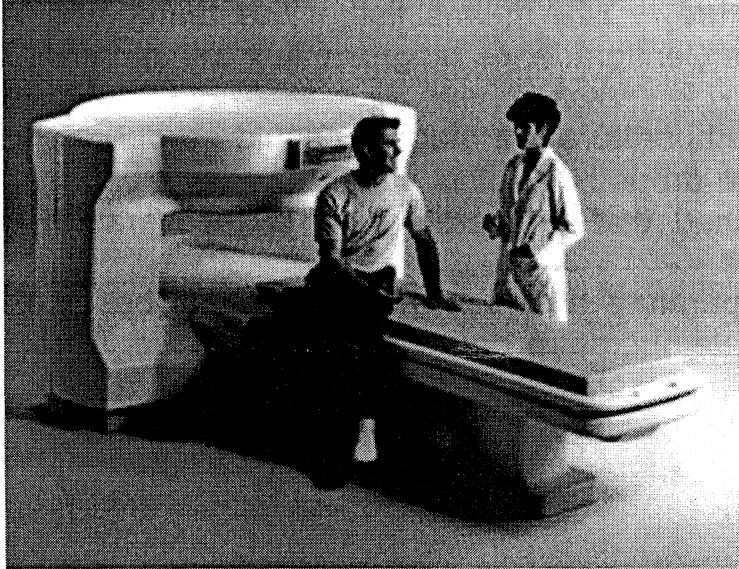


The main advantages of these types of magnets are:

- 1) No liquid cryogen,
- 2) The ability to "turn off" the magnetic field,
- 3) Relatively small fringe field.

Permanent Magnet

field and no missile effect. Due to weight considerations, **permanent magnets** are usually limited to maximum field strengths of 0.4 T. The main disadvantages of a **permanent magnet** are the cost of the magnet itself and supporting structures and the varying changes in the magnetic field. Field homogeneity can be an on-going problem in **permanent magnets**.



The patient lies on a scanning table between these two plates. Advantages of these systems are: 1) Relatively low cost, 2) No electricity or cryogenic liquids are needed to maintain the magnetic field, 3) Their more open design may help alleviate some patient anxiety, 4) Nearly nonexistent fringe field. It should be noted that not all vertical field magnets are permanent magnets.

Materials used for permanent magnets

- There are four classes of permanent magnets:
 - 1, Neodymium Iron Boron (NdFeB or NIB)
 - 2, Samarium Cobalt (SmCo)
 - 3, Alnico
 - 4, Ceramic or Ferrite

Br is the measure of its residual magnetic flux density in Gauss, which is the maximum flux the magnet is able to produce. (1 Gauss is like 6.45 lines/sq in)

Hc is the measure of the coercive magnetic field strength in Oersted,

or the point at which the magnet becomes demagnetized by an external field. (1Oersted is like 2.02 ampere-turns/inch)
BHmax is a term of overall energy density. The higher the number, the more powerful the magnet.

Tcoef of Br is the temperature coefficient of Br in terms of % per degree Centigrade. This tells you how the magnetic flux changes with respect to temperature. -0.20 means that if the temperature increases by 100 degrees Centigrade, its magnetic flux will decrease by 20%!

Tmax is the maximum temperature the magnet should be operated at. After the temperature drops below this value, it will still behave as it did before it reached that temperature (it is recoverable). (Degrees Centigrade)

Tcurie is the Curie temperature at which the magnet will become demagnetized. After the temperature drops below this value, it will not behave as it did before it reached that temperature. If the magnet is heated between Tmax and Tcurie, it will recover somewhat, but not fully (it is not recoverable). (degrees Centigrade)

Superconducting Magnet

A type of magnet whose field is generated by current in wires made of a superconducting material such as niobium-titanium, that has no resistance when operated at temperatures near absolute zero (-273,15°C, -459°F, 0K).

The coil of wire is kept at a temperature of 4.2K by immersing it in liquid helium. The coil and liquid helium is kept in a large dewar. In early magnet designs, this dewar was typically surrounded by a liquid nitrogen (77.4K) dewar, which acts as a thermal buffer between the room temperature (293K) and the liquid helium. In later magnet designs, the liquid nitrogen region was replaced by a dewar cooled by a refrigerator. This design eliminates the need to add liquid nitrogen to the magnet.

Superconducting magnets typically exhibit field strengths of greater than 0.5 T, operate clinically up to 3 T and have a horizontal field orientation, which makes them prone to missile effects without significant magnetic shielding.

Boil off Rate

Rate of cryogen (liquid helium) evaporation in **superconducting magnets**, usually measured in liters of liquid per hour. It increases during ramping of the magnet and with eddy currents induced in the cryoshields by pulsed field gradients. In calculating cryogen consumption additional transfer and filling losses have to be considered

Cryogen

A cooling agent, typically liquid helium or liquid nitrogen used to reduce the temperature of the magnet windings in a **superconducting magnet**. All cryogenic liquids are gases at normal temperatures and pressures. Different cryogens become liquids under different conditions of temperature and pressure, but all have two properties in common: they are extremely cold, and small amounts of liquid can expand into very large volumes of gas. The boiling points of cryogens are commonly below -150°C (-238°).

Ramping Up.

Ramping up of oxford magnet.

The supercon contains minimum 80% He and 20% Nitrogen. The normal ramp up procedure consumes 7 – 10 % of He. Normal ramp down consume 5% of He. Normal ramp up takes 5 hrs. and normal ramp down takes 4 hrs. to complete. Before start to ramp up ensure that all magnet power supplies are okay and install the magneto meter at the centre probe. Do not start ramp up procedure with below 80% He. For proceeding the ramp up first give a warm up heating to the heater (resistance) for a period of 3 – 5 minutes and set voltage as 2V (for high current). After the warm-up procedure set the current to 250A. The current will first jump to 4 A and comes down to 1A per minute. Synchronize the magnetometer until the field is approximately 7000 gauss. Once the magneto meter shows the reading of 14750 gauss, set the current level which is less than 250A and allow the field to stabilize for 45 minutes. The magnetometer reading is stable at the stabilization time. By making the fine control we can get a magnetic field strength of 15008 +/- 0.2 gauss. When

the field reaches 15008 turns, off the heater switch and allow 5 minutes to cool. After this turn the current and voltage limit to zero. When reaches zero wait 5 minutes for stabilization. Selective manual ramp down can also do with proper current and voltage to the heater to overcoming this problem is to use a secondary compensating magnetic field generated by a **set** of so-called shim coils to bring the overall field to the level of desired homogeneity.

Shimming

Correction of inhomogeneity of the magnetic field produced by the main magnet of a MRI system due to imperfections in the magnet or to the presence of external ferromagnetic objects. May involve changing the configuration of the magnet or the using of shim coils (active shimming) or adding or removing steel from the magnets poles (passive shimming) to fine-tune the magnetic field.

COILS USED IN MRI SYSTEM

RF COILS And Gradient coils

RF coils are the "antenna" of the MRI system that broadcasts the RF signal to the patient and/or receives the return signal. RF coils can be receive-only, in which case the body coil is used as a transmitter; or transmit and receive (transceiver).

A coil consist of one or more loops of a conductor used to create a magnetic field or to detect a changing magnetic field by voltage induced in the wire. A coil is usually a physically small antenna. The perfect coil produces a uniform magnetic field without significant radiation.

Radio Frequency Coil

RF coils create the B_1 field which rotates the net magnetization in a pulse sequence. They also detect the transverse magnetization as it precesses in the XY plane. RF coils can be divided into three general categories; RF coils can be divided into three general categories

- transmit and receive coils
- receive-only coils
- transmit-only coils

Transmit Receive Coil

(T/R) Also called transceiver coil. An RF coil that acts as a transmitter (T) producing the B1 excitation field and as a receiver (R) of the MRI signal. Such a coil requires a T/R switching circuit to switch between the two modes. A body coil is typically a T/R coil, but smaller volume T/R coils (head/extremities) are often used at high field as a possibility of reducing RF power absorption (SAR).

Transmitter Coil

The coil of the RF transmitter, inside the MR imager is used in excitation of the spins. Also called transmit-only coil it is used to create the B1 field. As a radio frequency generator send this coil bursts of RF pulses. These pulses serve to disturb the spins in the patient.

Receiver Coil

A coil, or antenna, positioned within the imaging volume and connected to the receiver circuitry that is used to detect or receive the MR signal from the patient as the disturbed spins relax back into their equilibrium distribution. Also called receive-only coil. Special-purpose coils are designed to optimize the SNR from a given region of the body. State-of-the-art coil systems include the use of four or more coils with four separate receivers. This method is often referred to as a phased array system. Receiver coil types include also solenoid, planar, volume and quadrature coils. The quality of the MR images depends on the SNR of the acquired signal from the patient. SNR is of the utmost importance in obtaining clear images of the interior of the human body

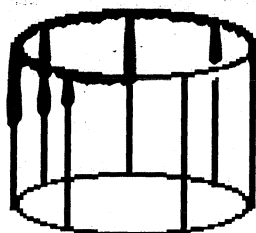
Multiply Tuned Coil

RF coil designed to operate at more than one resonance frequency, so that NMR of more than one kind of nucleus can be observed with the same coil.

VOLUME COIL

Radio frequency coil that surrounds a portion of the body. Volume coils have a better RF homogeneity than surface coils

Bird Cage Coil



Bird Cage
Coil

A transmit and receive RF imaging coil, which looks like a bird cage. The bird cage coil works on a different principle to conventionally tuned local and surround coils in that it behaves like a tuned transmission line with one complete cycle of standing wave around the circumference. The bird cage coil provides the best RF homogeneity of all the RF coils. One advantage is that it is simple to produce an exceedingly uniform B1 radio frequency field over most of the coil's volume, with the result of images with a high degree of uniformity. A second advantage is that nodes with zero voltage occur 90° away from the driven part of the coil, thus facilitating the introduction of a second signal in quadrature, which produces a circularly polarized radio frequency field. The bird cage coil is the coil of choice for imaging the head and brain. This type of coil is also used occasionally for imaging of the extremities, such as the knees.

Circularly Polarized Coil

A coil designed to excite or detect spins using two orthogonal transmit and/or receive channels. As a transmitter coil, there is a factor of 2 reductions in power required. The receiver coil has a better SNR than a linearly polarized coil.

Crossed Coil

RF coil pair arranged with their magnetic fields at right angles to each other in such a way as to minimize their mutual magnetic interaction.

Helmholtz Pair Coil

The Helmholtz pair coil consists of two circular coils parallel to each other. They are used as z gradient coils in MRI scanners. They are also used occasionally as RF coils for pelvis imaging and cervical spine imaging

Paired Saddle Coil

Paired saddle coils are commonly used for imaging of the knee. These coils provide better homogeneity of the RF in the area of interest and are used as volume coils, unlike surface coils. Paired saddle coils are also used for the x and y gradient coils. By running current in opposite directions in the two halves of the gradient coil, the magnetic field is made stronger near one and weaker near the other

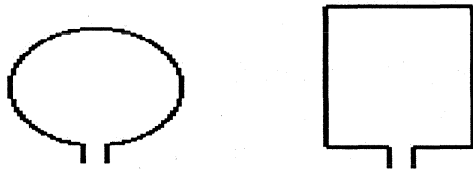
Quadrature Coil

A coil that produces an RF field with circular polarization. The RF power received from the RF power amplifier comes in two signals (quadrature detection), which have a phase difference of 90° . The RF transmit coil converts the power into a circularly polarised RF magnetic field. Quadrature coils can be used as both, transmit and/or receive coil. When used as a transmitter coil a factor of two power reduction over a linear coil results; as a receiver an increase in SNR of up to a factor of 1.41 can be achieved.

Single Turn Solenoid

A transmit and receive RF imaging coil that, in general, has a cylindrical shape. This small volume coil differs somewhat from a flat surface coil. The single turn consists of a significantly wider copper tape whose width is wider than the diameter of the coil. The single turn solenoid is useful for imaging extremities, such as wrist or knee.

SURFACE COIL



Surface Coils

A type of receiver coil, which is placed directly on or over the region of interest for increased magnetic sensitivity. Surface coils have a good SNR for tissues adjacent to the coil and the signal decrease with the distance, an eligibility homogeneity correction will balance this over the FOV. As only the region close to the surface coil will contribute to the noise, there is an improvement in the SNR for regions close to the coil, compared to the use of receiver coils that surround the corresponding part of the body. These coils are specifically designed for localized body regions, and provide improved signal to noise ratios by limiting the spatial extent of the excitation or reception. When a surface coil is used, a larger coil on the imager is used as the transmitter of RF energy to producing the 90° and 180° pulses.

Array Coil

An array coil combines the advantages of smaller coils (high SNR) with those of larger coils (large measurement field). This type of RF coil is composed of separate multiple smaller coils, which can be used individually (switchable coil) or combined. When used simultaneously, the elements can either be: coupled array coils - electrically coupled to each other through common transmission lines or mutual inductance

isolated array coils - electrically isolated from each other with separate transmission lines and receivers and minimum effective mutual inductance, and with the signals from each transmission line processed independently or at different frequencies

phased array coils - multiple small coils arranged to efficiently cover a specific anatomic region and obtain high-resolution, high-SNR images of a larger volume. The data from the individual coils is integrated by special software to produce the high-resolution images.

Body Wrap Around Coil(BWA)

A flexible surface coil for body imaging.

Linearly Polarized Coil(LP Coil)

A coil designed to excite or detect spins using one RF transmit and/or receive channel. The magnetic field has predominantly a single direction

Phased Array Coil

The phased array coils are typically operated as a receive only coil. A larger coil on the imager is used as the transmitter of RF energy to producing the 90° and 180° pulses. State-of-the-art coil systems include the use of four or more coils with four separate receivers. This method is often referred to as a phased array system, although the signals are not added such that the signal phase information is included. The use of phased array coils allows the number of signal averages to be decreased with their superior SNR and resolution, thereby decreasing scan time. Fast parallel imaging techniques using surface phased array multichannel coils, such as sensitivity encoding (SENSE) or simultaneous acquisition of spatial harmonics (SMASH) to further improve spatial and temporal resolution. High-sensitivity RF coils and digital processing algorithms have been developed that speed image acquisition and reconstruction during MRI. Also called linear array coil or synergy coil.

Saddle Coil

A coil geometry, which has two loops of a conductor wrapped around opposite sides of a cylinder. RF coil configuration design commonly used when the static magnetic field is coaxial with the axis of the coil along the long axis of the body (e.g., superconducting magnets and most resistive magnets) as opposed to solenoid or surface coil.

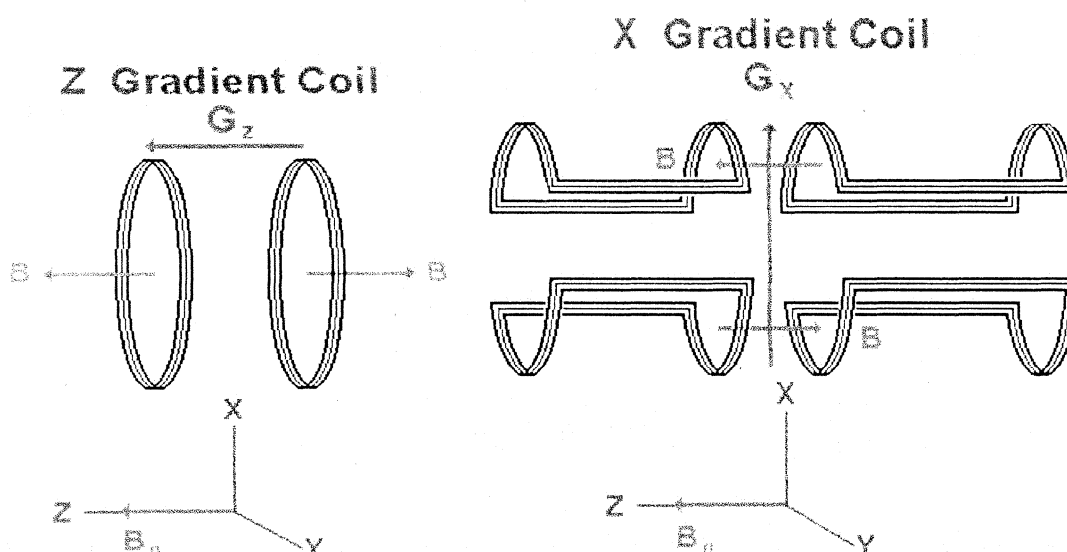
Surface Coil NMR

A surface coil placed over a region of interest will have an effective selectivity for a volume approximately subtended by the coil circumference and one radius deep from the coil center. Such a coil can be used for simple localization of sites for measurement of chemical shift spectra, especially of phosphorus, and blood flow studies. Some additional spatial selectivity can be achieved with magnetic field gradients.

GRADIENT COIL

Gradient coils are used to produce controlled variations in the main magnetic field (B_0) to provide spatial localization of the signals and to apply reversal pulses in some imaging techniques. The standard magnetic resonance coordinate system, a gradient in B_0 in the Z direction is achieved with an antihelmholtz type of coil. Current in the two coils flow in opposite directions creating a magnetic field gradient between the two coils. The B field at one coil adds to the B_0 field while the B field at the center of the other coil subtracts from the B_0 field

The X and Y gradients in the B_0 field are created by a pair coils. The X axis coils create a gradient in B_0 in the X direction due to the direction of the current through the coils. The Y axis coils provides a similar gradient in B_0 along the Y axis.



Current carrying coils designed to produce a desired magnetic field gradient (so that the magnetic field will be stronger in some locations than others). Proper design of the size and configuration of the coils is necessary to produce a controlled and uniform gradient. Three paired orthogonal current-carrying coils located within the magnet that are designed to produce desired gradient magnetic fields, which collectively and sequentially are superimposed on the main magnetic field (B_0) so that selective spatial excitation of the imaging volume can occur. Gradients are also used to apply reversal pulses in some fast imaging techniques. Gradient coils are used to produce deliberate variations in the main magnetic field (B_0). The set of gradient coils for the z-axis are, e.g. Helmholtz pairs, and for the x- and y-axis, paired saddle coils.

Figure 8 Coil

A magnetic field gradient coil shaped like the number eight.

Golay Coil

Term commonly used for a particular kind of gradient coil, commonly used to create magnetic field gradients perpendicular to the main magnetic field

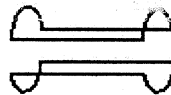
Helmholtz Coil

**Helmholtz
Pair Coil**

Pair of current carrying coils used to create uniform magnetic field in the center of the space between them. For circular coils, their separation equals their radius

Maxwell Coil

A particular kind of gradient coil, commonly used to create magnetic field gradients along the direction of the main magnetic field

Paired Saddle Coil

**Paired Saddle
Coil**

Paired saddle coils are commonly used for imaging of the knee. These coils provide better homogeneity of the RF in the area of interest and are used as volume coils, unlike surface coils. Paired saddle coils are also used for the x and y gradient coils. By running current in opposite directions in the two halves of the gradient coil, the magnetic field is made stronger near one and weaker near the other.

Shielded Gradient Coils

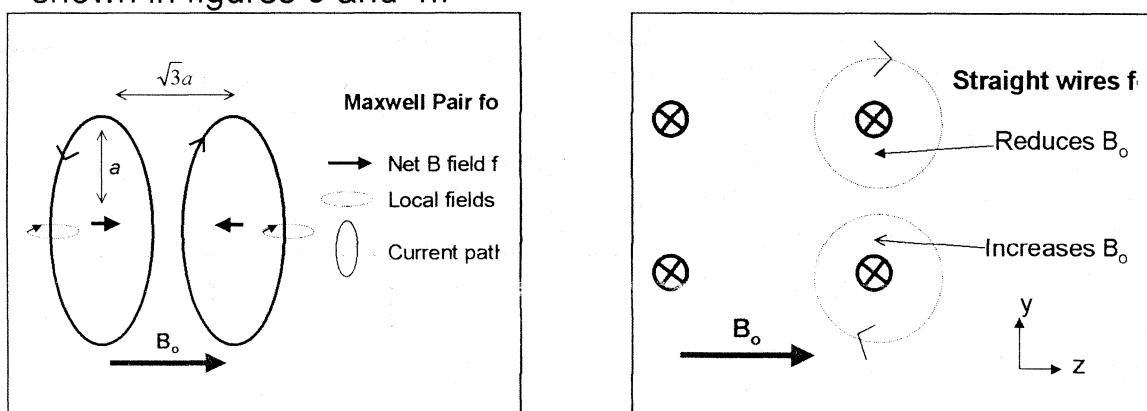
Current-carrying gradient coils with reduced gradient fringe field inside of the magnet cryostat structures like cryoshields and He-vessel. The shielding can be accomplished by secondary actively driven coils or by passive screens, which are inductively coupled to the gradient coils. In both cases eddy currents outside of the gradient system will be reduced.

Gradient coil designs

Gradient coils are carefully designed to produce linearly varying magnetic fields, as well as to have other features. The two simplest coil designs for use with a cylindrical whole body imaging system are shown in figures 3 and 4.:

Gradient coil designs

Gradient coils are carefully designed to produce linearly varying magnetic fields, as well as to have other features. The two simplest coil designs for use with a cylindrical whole body imaging system are shown in figures 3 and 4.:



The movement of the gradient coil wires in the magnetic field as electric currents are pulsed through them is the reason that MRI is so acoustically noisy.

Slice selection

To select a slice, a magnetic field gradient is applied in a certain direction, let us say z . This means that the Larmor frequency now varies along z (eqn.2, fig.5).

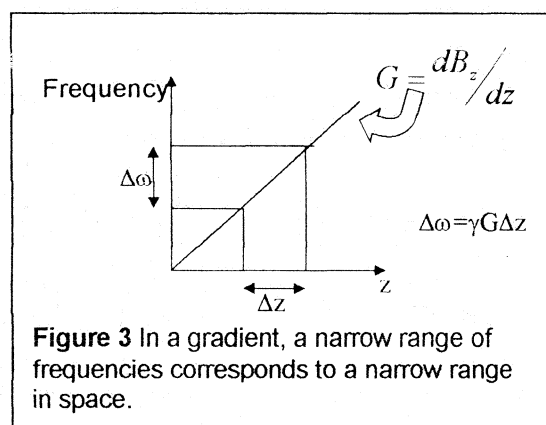
In order to select a slice it is necessary to apply an RF radiation over a narrow bandwidth of frequency, so that only spins in that range of frequencies are excited. In this way it is possible to select a slice perpendicular to the z -axis.

But how can a narrow range of frequencies be excited?

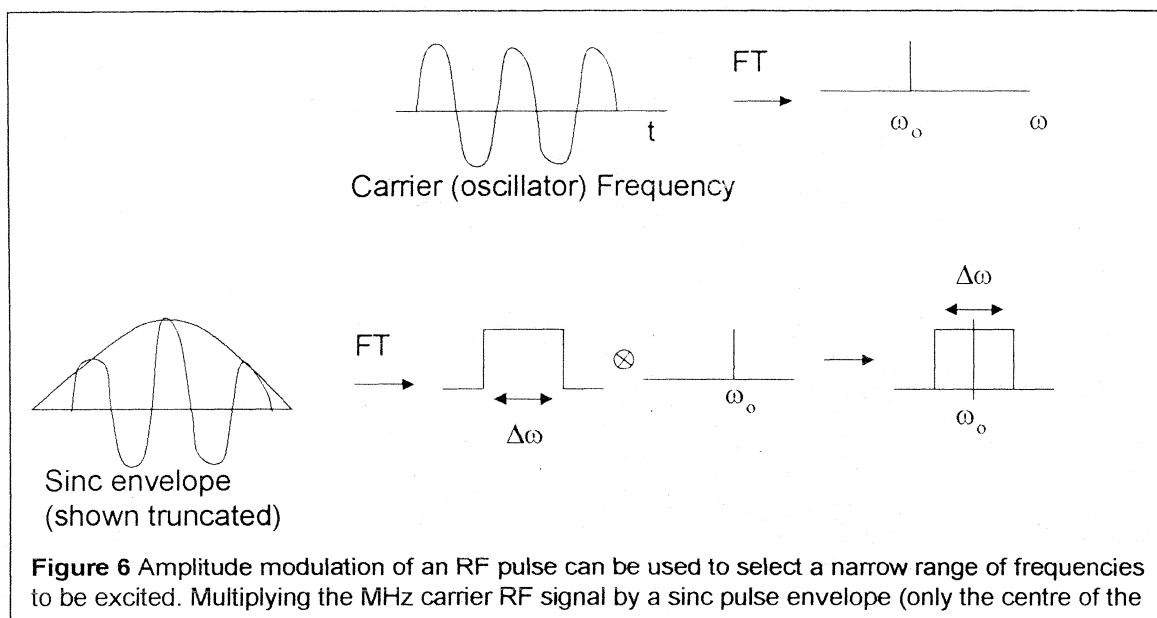
RF radiation oscillates at a single carrier frequency (corresponding to a single spike in the Fourier transform of the radiation time course). RF pulses are created by modulating the carrier frequency at audio frequencies (ie carrier signal is multiplied by an audio frequency window. The audio frequency window defines the pulse shape).

Remember that if you multiply two functions in time, it is equivalent to convolving them in frequency space. Therefore if the carrier signal is multiplied by a shaped envelope (eg a sinc function), this corresponds to convolving the spike with the Fourier transform of the shape of the envelope (for a sinc function in time, this is a rectangle in frequency).

Hence an RF pulse can be designed to excite a range of frequencies about ω_0 . By varying the shape of the pulse you can control thickness (and in fact the position) of the imaging slice.



You may be interested to know that Sir Peter Mansfield (from Nottingham) invented the slice selection technique. The patent income from this discovery paid for a large part of the building of the SPMMRC Centre.



K-SPACE

The inverse relationship between time and frequency: if the period of a function is T s, then its frequency is 1/T Hz..

Spatial encoding using magnetic field gradients

Virtually all modern MRI scanners use spatially variations in the applied magnetic field (B_0), to make the Larmor frequency spatially dependant. Magnetic field gradients can be formed in any direction

$$(G_x = \frac{dB_z}{dx}, G_y = \frac{dB_z}{dy}, G_z = \frac{dB_z}{dz}).$$

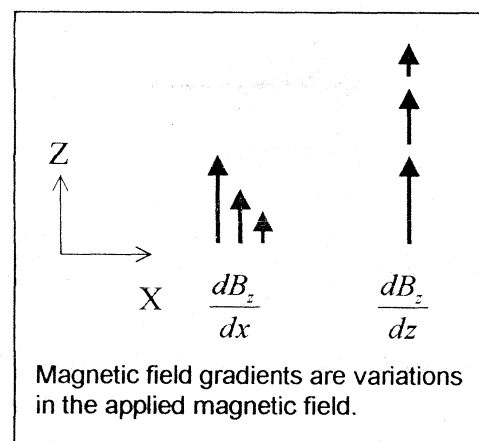
Therefore in the presence of a magnetic field gradient, the local applied magnetic field is given by:

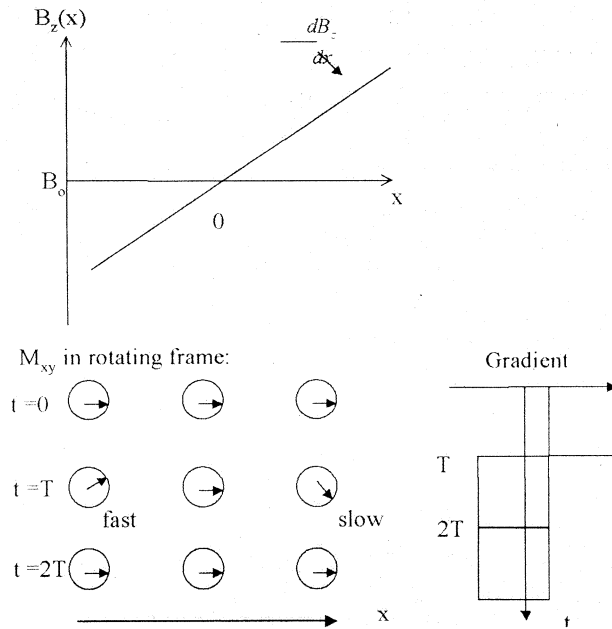
$$B(r) = B_0 + G_r r$$

where

B_0 = applied field from magnet (T)

G_r = applied gradient along r (mT/m) from the gradient coils





Therefore the Larmor frequency now depends on the position of the spins within the sample:

$$\omega = \gamma(B_0 + G_x r)$$

i.e. different parts of sample will produce transverse magnetisation precessing at different frequencies.

FOR Imaging sequences

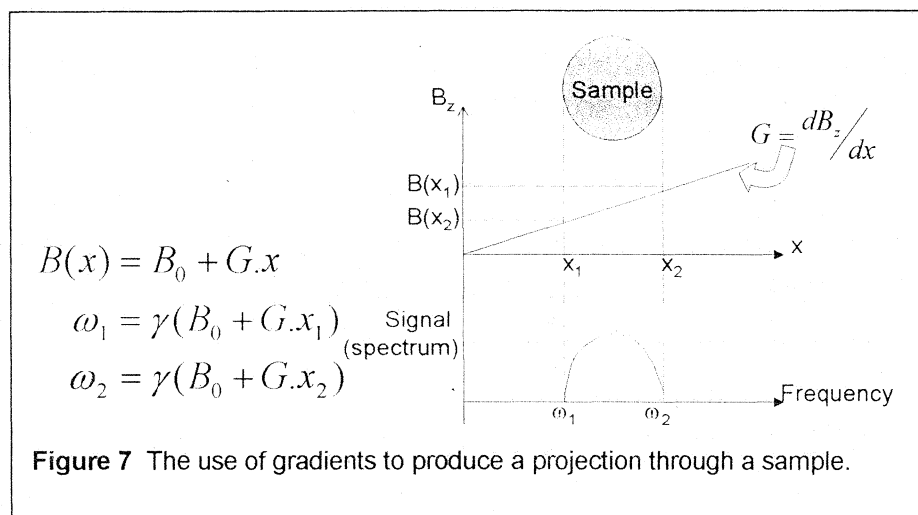
Once a slice has been selected we want to be able to create a picture the distribution of the spins within it. There are many ways of producing an image in MRI, and we are going to consider just a few of them.

Projection reconstruction

This is closely related to the method of CT imaging. After a slice has been excited using a gradient along z, another gradient is applied along x. Now the precessional frequency of the spins that have been excited depends on their position in x. Therefore

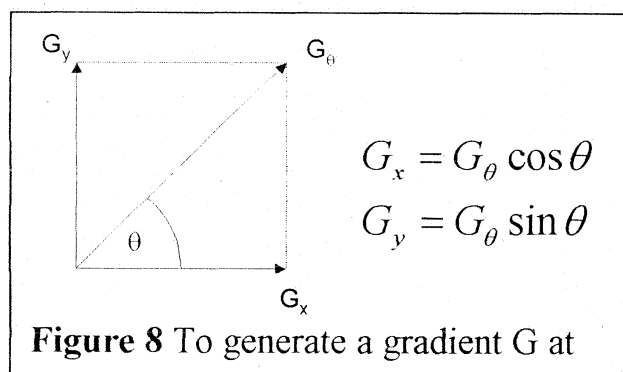
the time varying signal will be the sum of signals at a range of frequencies.

If the signal is Fourier transformed, so that the frequency spectrum is obtained, then the spectrum is the same as the projection through the object along the gradient.



If this process is repeated for different directions of the gradient (ie at various angles between x and y), then projections through the object can be acquired at a variety of angles. The image can then be reprocessed back projections

Gradients can be produced at various angles between x and y, by using combinations of the x and y gradients. The relative amplitudes of the currents through the Gx and Gy coils, determines the angle of the gradient with respect to the x axis. (figure 8).



Look at this analytically. The signal from a sample at position x is given by:

$$S_x(t) = \rho(x)e^{i\omega t}$$

where

$\omega = \gamma(G_x x + B_0)$ i.e. the frequency of rotation in the x y plane

$\rho(x)$ = total spin density at position x (ie projection along x)

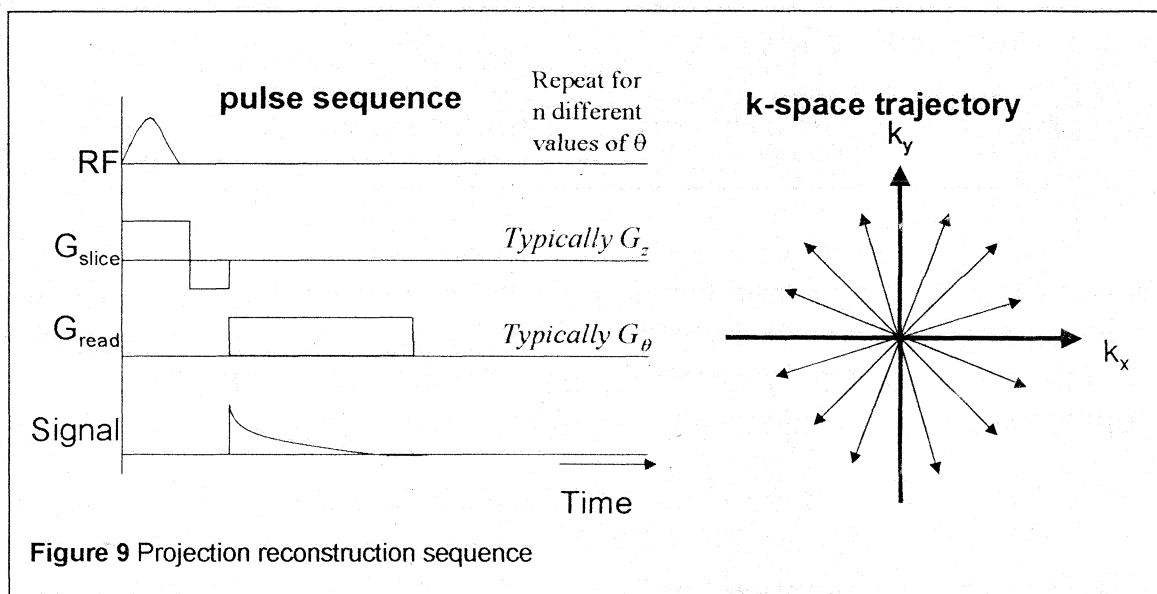
However the signal we measure is summed over the whole sample, ie over all values of x:

$$S_{total}(t) = \int \rho(x) e^{i\gamma(B_0 + G_x x)t} dx$$

As we know, phase sensitive detection mixes out the central Larmor frequency, ie the B_0 term, so that the total signal after detection is given by

$$S(t) = \int \rho(x) e^{i\gamma G_x x t} dx$$

ie the projection through the object along the direction of the gradient can be obtained by taking the Fourier transform of the signal (NB this



analysis neglects $T2^*$ decay). (If filtered backprojection reconstruction is to be used, then the signal can be multiplied by the appropriate Fourier domain filter before Fourier transforming).

The projection reconstruction imaging 'pulse sequence' is shown in figure 9. This shows the time course of the gradient and RF waveforms. The first line shows the slice selective RF pulse, and the second line shows the slice selection gradient. The third line shows the image encoding (frequency encoding/ readout) gradient. The fourth line shows the magnitude of the signal that will be acquired (ie the magnitude of the transverse magnetization, so net frequency offsets cannot be seen).

Applications of back projection reconstruction:

because imaging can start sampling very quickly after excitation, projection reconstruction is useful for imaging things with a short T_2^*
eg

Lungs (which have a very short T_2^* because of susceptibility differences between lung tissue and lung gas)

Sodium imaging (which has a very short T_2)

Microscopy (where susceptibility artefacts can also be a problem)

The images can be badly affected by inhomogeneities in the main magnetic field (B_0)

The total scan time is long.

k-space imaging

MR pulse sequences are best compared in k-space (reciprocal space), (which contains the spectrum of spatial frequencies contained in the image. If we define

$$k = \gamma \int G(t) dt$$

(for the general case where G is varying) so that the Fourier transform in eqn. 5 can be rewritten

$$S(k) = \int \rho(x) e^{-ik \cdot x} dx$$

i.e. signal obtained can be directly mapped into 'reciprocal' space (spatial frequency space). To sample higher spatial Frequencies (responsible for sharp edges and high resolution) you need higher (or longer) gradients.

Conventional Fourier Imaging

Fourier imaging resolves spatial information by three distinct procedures, called **selective excitation**, **phase encoding**, and **frequency encoding**). The slice-selection gradient spreads out the Larmor frequency over a broad range so that the frequencies contained in the RF pulse affect only a slice. The phase-encoding gradient is pulsed briefly after the RF excitation pulse has been

turned off, making the Larmor frequency depend on spatial position along the phase-encoding direction. Afterwards, magnetization components everywhere in the slice regain the same Larmor frequency, but the phase depends on their position along the phase-encoding direction. Spatial information in the phase-encoding direction can be resolved if many separate MR signals are collected. The amplitude of the phase-encoding gradient for each signal is decreased systematically. Each of these signals is measured as an echo while the frequency-encoding gradient is active, which creates a distribution of Larmor frequencies along the frequency-encoding direction. The first pulse of the frequency-encoding gradient is necessary for an echo to form during the middle of the second pulse.

Spin-warp imaging

if you remember, the problems of crude back projection reconstruction stem from the fact that k-space is not sampled uniformly. In spin warp imaging k space is sampled on a regular grid.

Remember that moving across k space you are increasing the number of radians per meter acquired by the spins. Therefore in order to move across the k space in the k_y direction, you must apply a y direction gradient (so that the spins gains relative phase shifts along y).

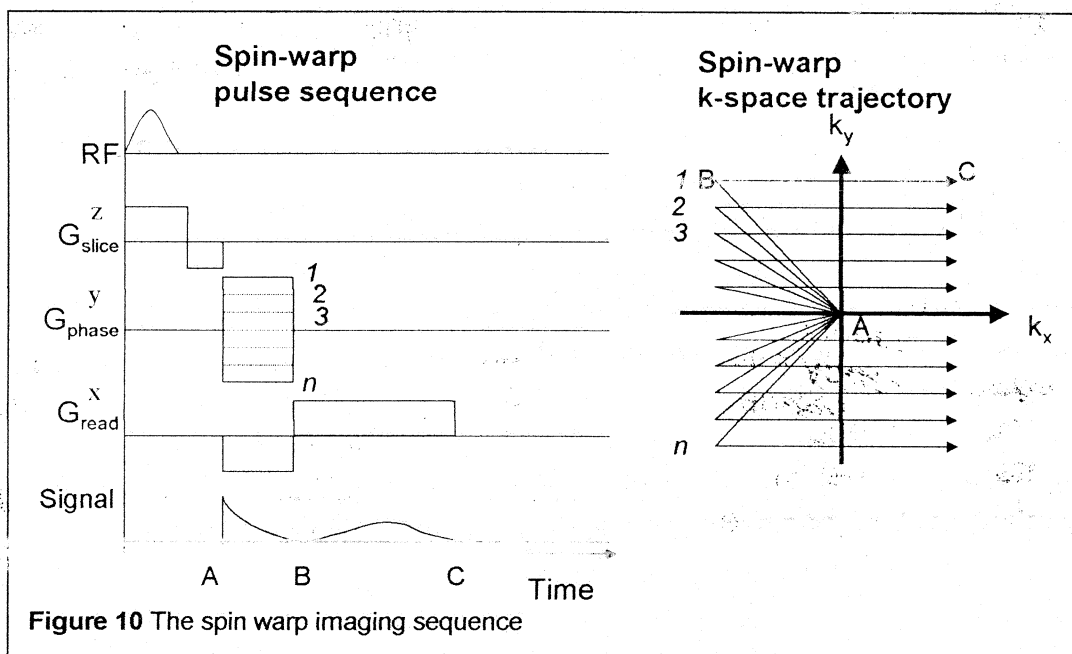


Figure 10 matches up the k space trajectory and the imaging sequence. Frequency encoding: The line BC in k space is acquired between the points B and C in the pulse sequence diagram, whilst the spins evolve under the read gradient. This part of the sequence looks very much like a standard projection reconstruction sequence: whilst the gradient is applied (and the signal is readout), we move across k space in the reciprocal direction of the gradient. You may have noticed that a negative read gradient is actually applied during the period AB, so that a gradient echo is acquired in the period BC. This corresponds to the fact that the line AB in the k-space trajectory is moving in a negative k_x direction. Phase encoding During the period AB an additional phase encoding gradient is applied. This makes the initial phase of the signal depend on both G_y and the position of sample in y (Morris 1985) and moves us in the y direction in k space.

Advantages:

Most widely used imaging sequence Field errors cause distortion, not blurring

Disadvantages

Long imaging times... motion causes ghost artifacts

Cine imaging.

Uses a single thick slice (10 – 40 mm) is acquired in the plane that best include the vessel. Acquisition is carried indexed to the patient's heart rate. The heart beat interval is divided in to several segments called cardiac phases. This is achieved using a gradient echo with small flip angle leaving enough magnetization on the longitudinal axis of the magnet for a multiple data collection from same slice throughout the cardiac cycle, as heart rate varies the time for T1 relaxation recovery also vary from one encoding cycle to the next. This motion can be corrected by giving ECG triggering and respiratory compensation. When displayed in a movie mode this images give excellent functional aspect of flow during the cardiac cycle through the vessel.

PROPELLER IMAGING

Periodically rotated overlapping parallel lines with enhanced reconstruction.

Multiple sets of parallel lines in k-space are filled

An angle is then shifted, and another sets of parallel lines are filled

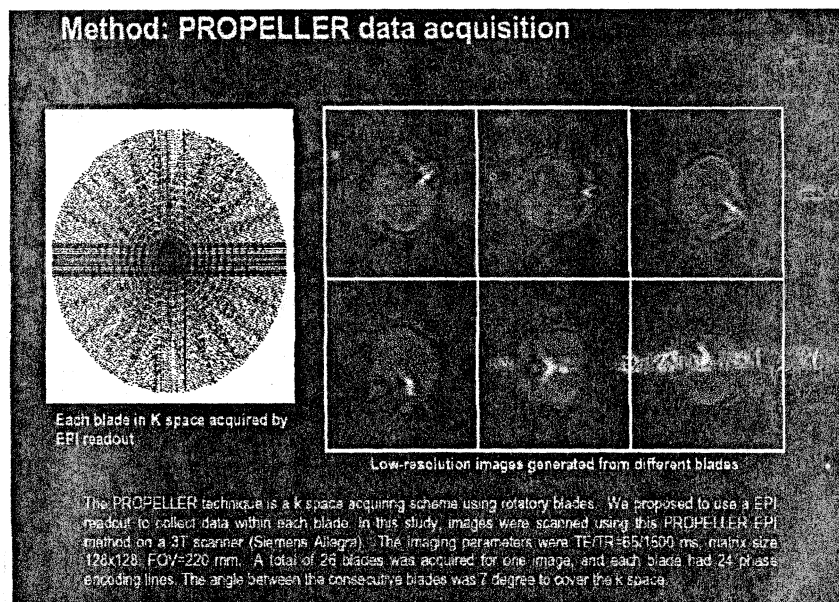
It overlapping first sets of parallel lines in the center It is continued until full range of k-space is filled

Central region of the k-space is sampled repeatedly

It suppress motion artifact.

FFT is required to collect the row data

Chemical shift artifact are more



VIEW SHARING

The acquired views to reconstruct a given image slice or volume, the data can be shared between one or more images. This will decrease the amount of data that must be newly sampled per images. It increase the temporal resolution of fast mri techniques Approach is similar to the radiographic fluoroscopy

Real time images are generated by continuously replacing the alternate lines in the CRT .It effectively doubling the apparent temporal resolution .

MR Fluoroscopy –intermittently refreshing some of the image data .

An image with 128 lines in the Kx, replacing 8 lines we can generate 16 images .

view sharing-increase the efficiency of dual TE FSE

Center of the k-space obtain separately with in each of the echo train portion Cardiac imaging,

Dynamic contrast –enhanced MR images

Dynamic events during fMRI ,

View sharing –impressive reduction imaging time

KEYHOLE IMAGING

A variant of the shared echo techniques K-space is covered completely on the first image only the central part (20%),which provide the most of the contrast,is updating on subsequent image It speed up the imaging by a factor of 5 Is very useful in fast recitative imaging of the same slice Its disadvantages is high frequency outer 80% of k space is shared information

Two major limitation of keyhole imaging techniques

If any motion between the baseline higher resolution images& dynamic contrast enhanced images

It will produce misregistration between the contrast &fine detail of the image

It produce some distortion in the contrast enhanced small vessel or finely heterogeneous pattern of tissue

Keyhole hybrid technique

It intermittently updating the peripheral views
Periphery of the k-space is filled in an interleaved manner. This has
been applied to 3D MRA
Contrast enhanced application, 3D-TRICKS .
Hybrid keyhole-spin echo, fse, gre

ZIP IMAGING

Zero fill interpolation processing.
Zip are used with any MR acquisition

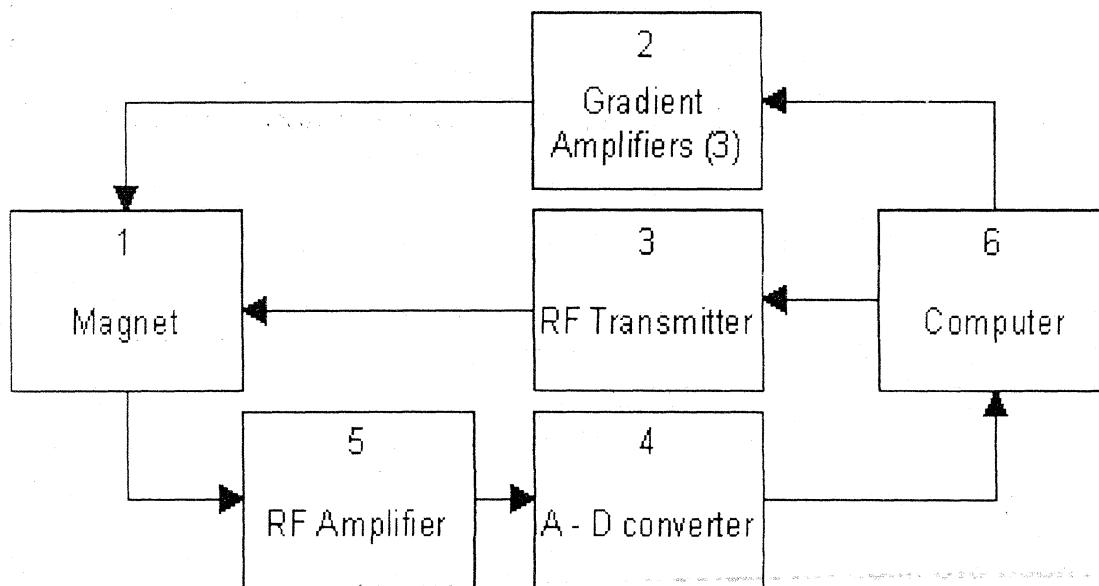
The portion of the K-space are not directly filled during the read out

These are filled with zeros ,FFT can be performed This techniques
allows for interpolation of large matrix
Zip create 512X512 matrix image from 256X256 image

Using this we can increase the apparent resolution
It reducing the scanning time with out much loss in resolution & SNR
It increase the truncation artifact

ELECTRONICS AND DATA PROCESSING

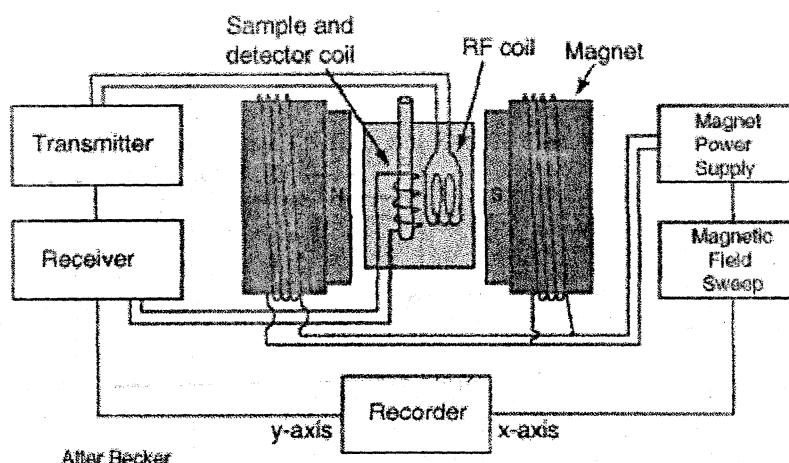
This will be an abbreviated description of the MRI scanner. To
get a general idea of the major components of a MRI scanner I have
included the following schematic diagram:



Starting from the right hand side, we have the computer that directs all of the action in the MRI acquisition and acquires and processes the data. The computer tells the gradient amplifiers and RF transmitter when to turn on and off to obtain the proper pulse sequence. The RF receiver amplifier is also controlled by the computer and relays the signal received by the RF coil from the patient to the A-D converter that digitizes the signal, and from there to the computer to be reconstructed into

Equipment: The MR Console

The MR Console is the "brains" of the system. It consists of one or more computers, complete with software, and the control electronics. It controls all the hardware, collects the data and presents it to the user.



MRI PULSE SEQUENCE

A preselected set of defined RF and gradient pulses, usually repeated many times during a scan, wherein the time interval between pulses and the amplitude and shape of the gradient waveforms will control NMR signal reception and affect the characteristics of the MR images. Pulse sequences are computer programs that control all hardware aspects of the measurement process.

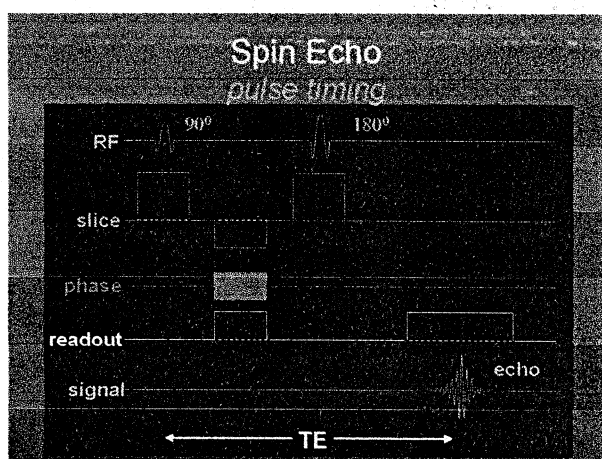
A recommended shorthand designation of interpulse times and excitation pulse used to generate a particular image is to list the repetition time (TR), the echo time (TE) and, if using inversion recovery, the inversion time (TI) with all times given in milliseconds and if using a gradient echo sequence the flip angle. For example, 3000/30/1000 would indicate an inversion recovery pulse sequence with TR of 3000 msec, TE of 30 msec, and TI of 1000 msec. Specific pulse sequence weightings are dependent on the field strength, the manufacturer and the pathology. See also Interpulse Times.

SPIN ECHO SEQUENCE

The most common pulse sequence used in MR imaging is based on the detection of a spin or Hahn echo. It uses 90° radio frequency

pulses to excite the magnetization and one or more 180° pulses to refocus the spins to generate signal echoes. In the pulse sequence timing diagram, the simplest form of a spin echo sequence is illustrated. The 90° pulse is first applied to the spin system and rotates the magnetization down into the $X'Y'$ plane. The transverse magnetization begins to dephase. At some point in time after the 90° pulse, a 180° pulse is applied. This pulse rotates the magnetization by 180° about the X' axis. The train of refocusing 180° pulses serves to generate repetitive signal echoes (hence the name spin echo). The 180° pulses occur at time, $\frac{1}{2}TE + iTE$ $i = 0, 1, \dots, n$, and the signal echoes at iTE . The echoes serve to rephase all the coherence in the x y -magnetization which are lost during the time between 90° and 180° pulse due to dephasing.

The recovery of the z -magnetization occurs with the T_1 relaxation time and typically at a much slower rate than the T_2 -decay, because in general T_1 is greater than T_2 for living tissues and is in the range of 100–2000 ms. The SE pulse sequence was devised in the early days of NMR days by Carr and Purcell and exists now in many forms: the multi echo pulse sequence using single or multislice acquisition, the fast spin echo (FSE/TSE) pulse sequence, echo planar imaging (EPI) pulse sequences and the GRASE pulse sequence are all basically spin echo (SE) sequences. In the simplest form of SE imaging, the pulse sequence has to be repeated as many times as the image has lines.



A slice selective 90° RF pulse is applied in conjunction with a slice selection gradient. A period of time equal to $TE/2$ elapses and a 180° slice selective 180° pulse is applied in conjunction with the slice selection gradient. A phase encoding gradient is applied between the 90° and 180° pulses. The phase encoding gradient could be applied after the 180° pulse, however if we want to minimize the TE period the pulse is applied between the 90° and 180° RF pulses. The frequency encoding gradient is applied after the 180° pulse during the time that echo is collected. The recorded signal is the echo. The FID, which is found after every 90° pulse, is not used. One additional gradient is applied between the 90° and 180° pulses. This gradient is along the same direction as the frequency encoding gradient. It dephases the spins so that they will rephase by the center of the echo. This gradient in effect prepares the signal to be at the edge of k-space by the start of the acquisition of the echo. The entire sequence is repeated every TR seconds until all the phase encoding steps have been recorded.

PDWeighted:ShortTE(20Ms)AndLongTR.

T1Weighted:ShortTE(10-20Ms)AndShortTR(300-600Ms)

T2Weighted:LongTE(GreaterThan60Ms)AndLongTR(GreaterThan1600Ms)

With Spin Echo Imaging No $T2^*$ Occurs.

Dual Echo Sequen (DE - dual / double echo)

Dual echo sequences include images with different weightings and or echo times and are used to obtain both, proton density and T2 weighted images or in phase and out of phase gradient echo images, simultaneously without increasing the measurement time.

Modified Spin Echo(MSE)

A spin echo technique with a flip angle over 90° .

Multi Echo Multiplanar (MEMP)

Sequence with a multislice and multi echo acquisition in one TR.

Partial Saturation Spin Echo (PSSE)

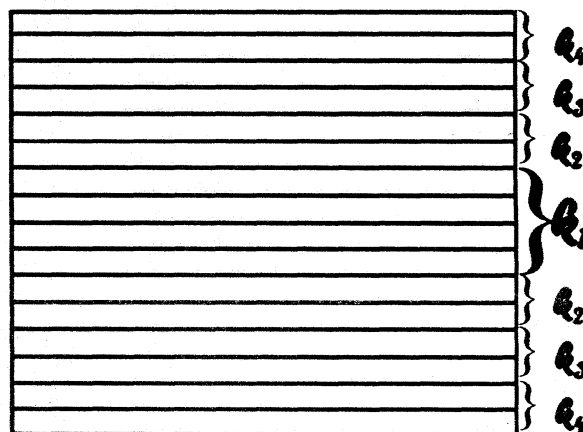
Partial saturation sequence in which the signal is detected as a spin echo. Even though a spin echo is used, there will not necessarily be a significant contribution of the T2 relaxation time to image contrast, unless the echo time, TE, is on the order of or longer than T2.

Variable Echo Multiplanar (VEMP)

MR imaging spin echo pulse sequence in which signals for multiple variable echoes are collected

FAST SPIN ECHO (FSE)

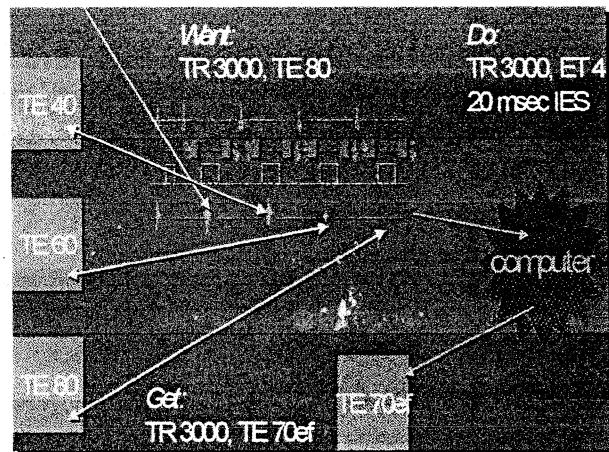
In conventional spin echo, depending up up on the phase encoding steps we repeat the TR and fill the k-space for each echo (e.g. for an 8 echo train we will get a eight different images), but in FSE we will get only one K-space and also we get 8 lines at a time depending up on the ETL. In this one TR we will get 8 lines next TR we will get 8 more lines. If your phase encoding steps is 256 & ETL-8 ,we only have to repeat the k-space to 32 times



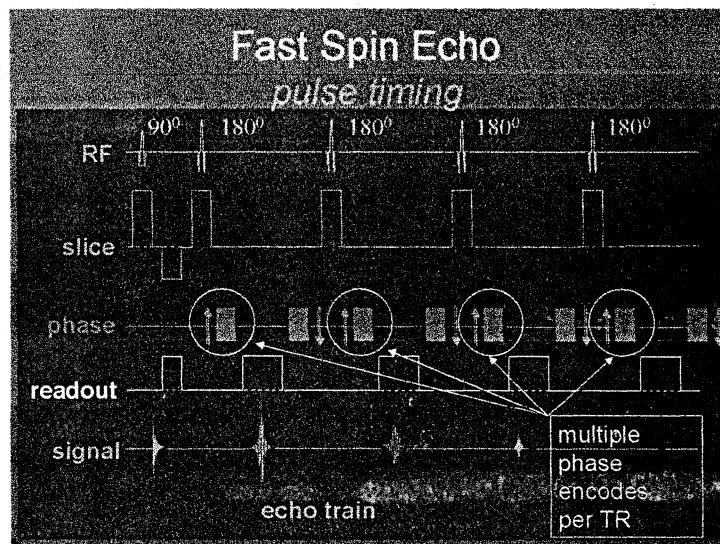
ETL

It refers to the number of echoes used in-FSE and it ranges typically from 3-32. The time interval between successive echoes is called the echo spacing (ESP), the typical ESP is in the order of 16-20ms at a

typical high field band width of 32 khz .in FSE the only TEs we can choose are integral multiples of the echo spacing this is called effective TE_{eff}



Center of k-space has maximum signal .Edges of k-space ,we get less signal .we use phase encoding with lowest strength for effective TE. Each subsequent phase encoding steps will have a gradient with more and more amplitude



In case of the acquisition of 2 echoes this type of a sequence is named double fast spin echo / dual echo sequence, the first echo is usually density and the second echo is T2 weighted. Other terms for this technique are

with 90° phase shift in the rotating frame of reference between the 90° pulse and the subsequent 180° pulses in order to reduce accumulating effects of imperfections in the 180° pulses. Suppression of effects of pulse error accumulation can alternatively be achieved by switching phases of the 180° pulses by 180°

Double Fast Spin Echo(DFSE)

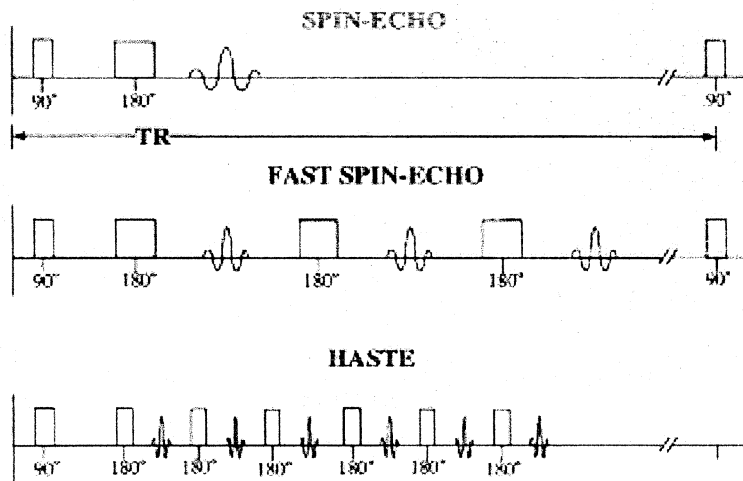
Simultaneously acquired T2 and proton density weighted TE in FSE echo images

Double Turbo Spin Echo(DTSE / DE TSE)

Simultaneously acquired T2 and proton density weighted echoes in a TSE sequence.

Half Fourier Acquisition Single Shot Turbo Spin Echo(HASTE)

A pulse sequence with data acquisition after an initial preparation pulse for contrast enhancement with the use of a very long echo train (Single shot TSE), whereat each echo is individually phase encoded. This technique is a heavily T2 weighted, high speed sequence with partial Fourier technique, a great sensitivity for fluid detection and a fast acquisition time of about 1 sec per slice. This advantage makes it possible for using breath-hold with excellent motionless MRI, e.g. used for liver and lung imaging.



Rapid Acquisition with Refocused Echoes(RARE) (GE)

If the image lines from multiple echoes are used for the same image, this results in the RARE pulse sequence. The sequence is similar to fast spin echo.

Turbo Spin Echo (TSE)

A pulse sequence characterized by a series of rapidly applied 180° rephasing pulses and multiple echoes.(SIEMENS)

Ultrashort Turbo Spin Echo

A turbo spin echo (TSE / FSE) technique with extremely short echo spacing, resulting in shorter scan times. This is an advantage in areas where motion is a problem, for example dynamic or abdominal imaging. The shorter scan time and echo spacing are achieved by using a higher TSE factor and an increased data sampling. Disadvantages are the decrease in SNR (caused through the increase of the bandwidth) and artifacts if minimum echo spacing is used (incomplete dephasing of the 180° pulse FID).

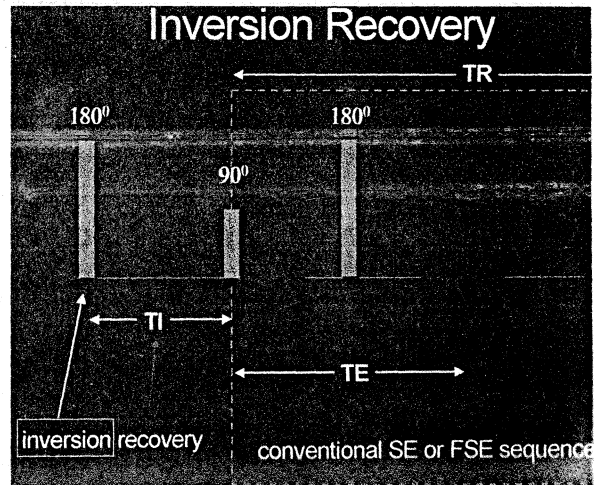
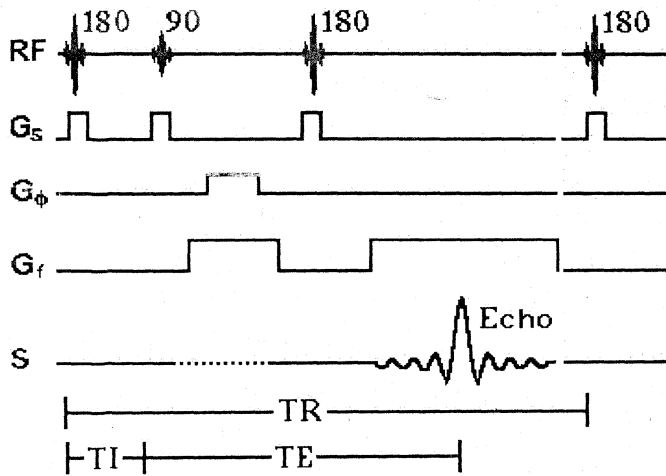
INVERSION RECOVERY(IR)

Inversion recovery is a MR technique, which can be incorporated into MR imaging, wherein the nuclear magnetization is inverted at a time on the order of T_1 before the regular imaging pulse-gradient sequences. The resulting partial relaxation of the spins in the different structures being imaged can be used to produce an image that depends strongly on T_1 . This may bring out differences in the appearance of structures with different T_1 relaxation times. Note that this does not directly produce an image of T_1 . T_1 in a given region can be calculated from the change in the MR signal from the region due to the inversion pulse compared to the signal with no inversion pulse or an inversion pulse with a different inversion time. This sequence involves successive 180° and 90° pulses. The **inversion recovery** sequence is specified in terms of three parameters, inversion time (TI), repetition time (TR) and echo time (TE).

Inversion Recovery Sequence

This pulse sequence produces signals, which represent the longitudinal magnetization existent after the application of a 180° inversion RF pulse and rotates the net magnetization down to the negative z-axis. The magnetization undergoes spin Lattice relaxation and returns toward its equilibrium position along the positive z-axis. Before it reaches equilibrium, a 90° pulse is applied that rotates the longitudinal magnetization into the xy plane. Once magnetization is present in the xy plane it rotates about the z-axis and dephases giving a FID.

In the pulse sequence timing diagram, the basic **inversion recovery** sequence is illustrated. The 180° inversion pulse is attached prior to the 90° excitation pulse of a spin echo acquisition. The **inversion recovery** sequence has the advantage, that it can provide very strong contrast between tissues having different T_1 relaxation times or to suppress tissues like fluid or fat. But the disadvantage is, that the additional inversion radio frequency RF pulse makes this sequence less time efficient than the other pulse sequences.



Inversion Recovery Spin Echo (IRSE)

Form of **inversion recovery** imaging in which the signal is detected as a spin echo. For TE short compared to the T2 relaxation time, there will be only a small effect of T2 differences on image intensities; for longer TE's, the effect of T2 may be significant

Short T1 Inversion Recovery (STIR)

Also called Short Tau (TI-inversion time) **Inversion Recovery**. It is an **inversion recovery** sequence where the TI time is set to $T1 \ln 2$, where T1 is the spin Lattice relaxation time of the component that should be suppressed. For fat suppression that component is fat, for water suppression it is water. This technique only works when the T1 values for the two components are different. Inversion recovery doubles the distance spins will recover, allowing more time for T1 differences. A 180° preparation pulse inverts the net magnetization to the negative longitudinal magnetization prior to the 90-degree excitation pulse. This specialized application of the **Inversion Recovery** pulse sequence set the inversion time (TI) of the sequence at 0.69 times

the T1 of fat. The T1 of fat at 1.5T is approximately 250 with a null point of 170 ms while at 0.5T its 215 with a 148 ms null point. At the moment of excitation, about 120 to 170 ms after the 180° inversion pulse (depending of the magnetic field) the magnetization of the fat signal has just risen to zero from its original, negative, value and no fat signal is available to be flipped into the transverse plane. When deciding on the optimal T1 time, factors to be considered include not only the main field strength, but also the tissue to be suppressed and the anatomy. In comparison to a conventional spin echo where tissues with a short T1 are bright due to faster recovery, fat signal is reversed or darkened. Because body fluids have both a long T1 and a long T2, it is evident that STIR offers the possibility of extremely sensitive detection of body fluid. This is of course, only true for stationary fluid such as edema, as the MR signal of flowing fluids is governed by other factors.

Flow Sensitive Alternating Inversion Recovery (FAIR)

In this sequence 2 inversion recovery images are acquired, one with a nonselective and the other with a slice selective inversion pulse. The z-magnetization in the first sequence is independent of flow. Inflowing spins give z-magnetization from second pulse. A major signal loss in FAIR is the T1 relaxation of tagged blood in transit to the imaging slice. Sharper edges of the inversion pulse give narrow spacing between the inversion edge and the 1st slice because reduced transit time gives lower T1 relaxation induced signal loss. The difference of the images in a consequence contains information proportional to flow (blood partition coefficient). Standard adiabatic inversion RF pulse does not have good slice-profile, because of power/SAR limitation. A c-shaped frequency offset corrected inversion (FOCI) RF pulse can help to increase the signal. Perfusion imaging, e.g. myocardial, using tissue water as endogenous contrast is suggested

Fluid Attenuation Inversion Recovery (FLAIR)

A special inversion recovery sequence with long TI to remove the effects of fluid from the resulting images. The TI time is set to the relaxation time of the component that should be suppressed. For fluid suppression the inversion time (long TI) is set to the zero crossing point of fluid, resulting in the signal being 'erased'.

Lesions that are normally covered by bright fluid signals using conventional T2 contrast are made visible by the dark fluid technique. Therefore it is an important technique for the differentiation of brain and spine lesions. This sequence is used in the brain to suppress CSF to bring out the periventricular hyperintense lesion, such as multiple sclerosis plaques.

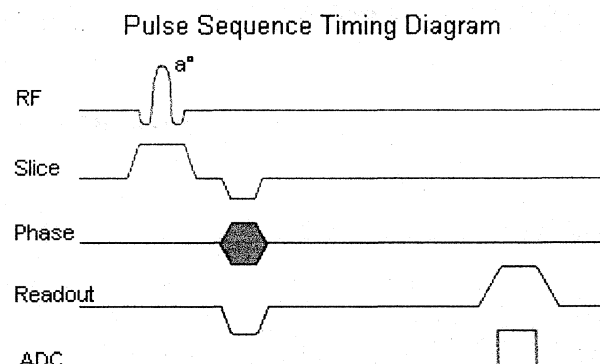
Turbo Inversion Recovery (TIR / TIRM / IR-TSE)

Turbo Spin Echo / FIR - **Fast Inversion Recovery**
A turbo / fast **spin echo sequence** with long TI for fluid suppression (FLAIR) or with short TI for fat suppression (STIR). This sequence allows for a true **inversion recovery** display that shows the arithmetic sign of the signal.

TIRM means a turboIR with a **magnitude** display.

GRADIENT ECHO SEQUENCE (GRE - sequence)

A gradient echo is generated by using a pair of bipolar gradient pulses. In the **pulse sequence timing diagram**, the basic gradient echo sequence is illustrated. There is no refocusing 180° pulse and the data are sampled during a gradient echo, which is achieved by **dephasing** the spins with a negatively pulsed gradient before they are rephased by an opposite gradient with opposite polarity to generate the echo.



There you will find a description of the components. The excitation pulse is termed the alpha pulse. It tilts the magnetization by a flip angle α , which is typically between 0° and 90° . With a small flip angle there is a reduction in the value of transverse magnetization that will affect subsequent RF pulses. The flip angle can also be slowly increased during data acquisition (variable flip angle: tilt optimized non-saturation excitation). The data are not acquired in a steady state, where z-magnetization recovery and destruction by ad-pulses are balanced. However, the z-magnetization is used up by tilting a little more of the remaining z-magnetization into the xy-plane for each acquired imaging line. Gradient echo imaging is typically accomplished by examining the FID, whereas the read gradient is turned on for localization of the signal in the readout direction. T_2^* is the characteristic decay time constant associated with the FID. The contrast and signal generated by a gradient echo depend on the size of the longitudinal magnetization and the flip angle. When flip angle is 90° the sequence is identical to the so-called partial saturation or saturation recovery pulse sequence. In standard GRE imaging, this basic pulse sequence is repeated as many times as image lines have to be acquired. Additional gradients or radio frequency pulses are introduced with the aim to spoil to refocus the xy-magnetization at the moment when the spin system is subject to the next pulse.

As a result of the short repetition time, the z-magnetization cannot fully recover and after a few initial α pulses there is an equilibrium established between z-magnetization recovery and z-magnetization reduction due to the α pulses.

Gradient echoes have a lower SAR, are more sensitive to field inhomogeneities and have a reduced crosstalk, so that a small or no slice gap can be used. In or out of phase imaging depending on the selected TE (and field strength of the magnet) is possible. As the flip angle is decreased, T1 weighting can be maintained by reducing the TR. T_2^* weighting can be minimized by keeping the TE as short as

possible, but pure T2 weighting is not possible. By using a reduced flip angle, some of the magnetization value remains longitudinal (less time needed to achieve full recovery) and for a certain T1 and TR, there exist one flip angle that will give the most signal, known as the "Ernst angle"

Contrast values:

PD weighted: Small flip angle (no T1), long TR (no T1) and short TE (no T2*)

T1 weighted: Large flip angle (70°), short TR (less than 50ms) and short TE

T2* weighted: Small flip angle, some longer TR (100 ms) and long TE (20 ms)

Classification of GRE sequences

1. T1 weighted or incoherent/(RF or gradient) spoiled GRE sequences
2. T1/T2* weighted or coherent/refocused GRE sequences.
3. T2 weighted contrast enhanced GRE sequencesultrafast GRE sequences.

GRADIENT RECALLED ECHO SEQUENCE

An MRI sequence producing signals called gradient echoes as a result of the application of a refocusing echo. In the gradient recalled echo sequence, a slice selective RF pulse is applied to the imaged object. This RF pulse typically produces a rotation angle of between 10° and 90°.

A slice selection gradient is applied with the RF pulse. A phase encoding gradient is applied and a dephasing frequency encoding gradient is applied at the same time so as to cause the spins to be in phase at the center of the acquisition period. This gradient is negative in sign from that of the frequency-encoding gradient turned on during the acquisition of the signal. An echo is produced when the frequency encoding gradient is turned on because this gradient refocuses the dephasing, which occurred from the gradients.

GRADIENT ECHO FAST SCANNING TECHNIQUES

DRIVEN EQUILIBRIUM FAST GRADIENT RECALLED ACQUISITION IN THE STEADY STATE DEPHASING GRADIENT(DE FGR)

A gradient echo sequence using a pulse, which sensitizes the sequence to variations in T2, rather than waiting for T1 relaxation

Fast Gradient Recalled Echo(FGRE)

In the gradient recalled echo sequence, a slice selective RF pulse is applied to the imaged object.

SPGR(Spoiled Gradient)

With SPGR T2 dependence is almost completely eliminated. If we use this sequence with short TE and moderate flip angle, one could achieve T1 weighted gradient echo images. This pulse sequences uses very short TR values neither a transverse or longitudinal state has enough time to become fully established during the course of imaging. Because the image acquisition has not taken place under steady state condition, non uniform weighting of the data along phase encoded axis will occur and can cause loss of image resolution. Because the TR value are so short, small flip angle may be used to minimize saturation to preserve the SNR. In SPGR uses a simple preparative non-selective 180° pulse along the phase encoding axis(Spoiling gradient) which invert the tissue magnetization across the sample and produces T1 weighted images. The SPGR spoiling is achieved by adding phase shift to each successive RF excitation pulse. This result in tilting of residual magnetization in out of phase with each other to prevent the build up of steady state component and thus eliminate T2 contribution. This approaches only works if the transmitter and receiver keep a constant phase relationship for each transmit and receive function.

There are various way to get SPGR images

- 1,By applying RF spoiling
- 2,BY applying variable gradient spoilers
- 3,BY lengthening TR

RF spoiling

In this scheme, a phase offset is added to each successive RF pulse. This causes a corresponding phase shift in successive Mss vectors. By maintaining a constant phase relationship between the transmitter and the receiver

Variable Gradient Spoilers

Spoiling can also be achieved by using gradient spoilers. This is accomplished by introducing an additional gradient with variable strength from cycle to cycle

Lengthening TR

When TR is sufficiently large, there is enough time to allow complete dephasing of the spin in the transverse plane

CE-FFE-T1: Contrast Enhanced Fast Field Echo with T1 Weighting,
 GFE: Gradient Field Echo,
 FLASH: Fast Low Angle Shot,
 PS: Partial Saturation,
 RF spoiled FAST: RF Spoiled Fourier Acquired Steady State Technique,
 RSSARGE: Radio Frequency Spoiled Steady State Acquisition Rewound Gradient Echo
 S-GRE: Spoiled Gradient Echo,
 SHORT: Short Repetition Techniques,
 SPGR: Spoiled Gradient Recalled (spoiled GRASS),
 STAGE: T1W T1 weighted Small Tip Angle Gradient Echo,
 T1-FAST: T1 weighted Fourier Acquired Steady State Technique,
 T1-FFE :T1 weighted Fast Field Echo.

Abdominal Imaging

General MRI of the abdomen can consist of T1 or T2 weighted spin echo, fast spin echo (FSE, TSE) or gradient echo sequences with fat suppression and contrast enhanced MRI techniques. The examined organs include liver, pancreas, spleen, kidneys, adrenals as well as parts of the stomach and intestine. Respiratory compensation and breath hold imaging is mandatory for good image quality.

T1 weighted sequences are more sensitive for lesion detection than T2 weighted sequences at 0.5 T, while higher field strengths (greater than 1.0 T), T2 weighted and spoiled gradient echo sequences are used for focal lesion detection. Gradient echo in phase T1 breath hold can be performed as a dynamic series with the ability to visualize the blood distribution. Phases of contrast enhancement include the capillary or arterial dominant phase for demonstrating hypervascular lesions, in liver imaging the portal venous phase demonstrates the maximum difference between the liver and hypovascular lesions, while the equilibrium phase demonstrates interstitial disbursement for edematous and malignant tissues.

Out of phase gradient echo imaging for the abdomen is a lipid-type tissue sensitive sequence and is useful for the visualization of focal hepatic lesions, fatty liver, haemochromatosis, adrenal lesions and renal masses. The standards for abdominal MRI vary according to clinical sites based on sequence availability and MRI equipment. Specific abdominal imaging coils and liver-specific contrast agents targeted to the reticuloendothelial system (RES) of the liver and spleen, achieve the goals of improved detection and localization in the liver.

Contrast Enhanced Fast Field Echo with T1 Weighting(CE-FFE-T1)

A T1 weighted gradient echo sequence

Fast Spoiled Gradient Echo(FSPGR)

A sequence similar to Turbo FLASH or Turbo Field Echo

Refocused Gradient Echo Sequence

Refocused GRE sequences use a refocusing gradient in the phase encoding direction during the end module to maximize (refocus) remaining xy- (transverse) magnetization at the time when the next excitation is due, while the other two gradients are, in any case, balanced.

When the next excitation pulse is sent into the system with an opposed phase, it tilts the magnetization in the \square direction. As a result the z-magnetization is again partly tilted into the xy-plane, while the remaining xy-magnetization is tilted partly into the z-

direction.

Companies use different acronyms to describe certain techniques.

Different terms for these gradient echo pulse sequences:

R-GRE: Refocused Gradient Echo,

FAST: Fourier Acquired Steady State,

FFE: Fast Field echo,

FISP: Fast Imaging with Steady State Precession,

F-SHORT SHORT Repetition Technique Based on Free Induction Decay,

GFEC: Gradient Field Echo with Contrast,

GRASS: Gradient Recalled Acquisition in Steady State,

ROAST: Resonant Offset Averaging in the Steady State,

SSFP: Steady State Free Precession.

STERF: Steady State Technique with Refocused FID

Fast Field Echo(FFE)

An echo signal generated from a FID by means of a bipolar switched magnetic gradient. The preparation module of the pulse sequence consists of an excitation pulse. The magnetization tilts by a flip angle between 0° and 90° .

Fast Gradient Recalled Echo(FGRE)

In the gradient recalled echo sequence, a slice selective RF pulse is applied to the imaged object.

Fast Low Angle Recalled Echoes (FLARE)

A sequence producing signals called gradient echoes as a result of the application of a refocusing echo with low flip angles.

SSFP (Steady State Free Precession) / PSIF

Steady state gradient sequence is designed to operate either single or multi-section mode by using extremely short TR values ($TR < 50\text{mSec.}$) and to produce a coherent steady state free precession. The steady state can be controlled in two ways. Through the use of resonant offset averaging on the read and slice select gradients. Through the application of rewinder gradients on the phase encoded axis. Steady state is the sum of two components-

FID occurring early in the cycle (just after each RF pulse) and A stimulated echo (St.E) occurring late in the cycle (just before next RF pulse.) It is possible to recover both signals simultaneously and coherently, provided gradient profiles are perfectly balanced.

If the sequence were run for only one cycle no echo is formed. An echo is formed in this sequence by magnetization brought in to transverse plane by RF pulse in the preceding cycle. The effective echo time is $TR + TE$ [SSFP => $TE = 2(TR - 9)$

Relatively long evaluating period before echo collection allows for natural transverse decay of magnetization to occur. So the images from SSFP sequence is highly T2 weighted. In this each \square pulse contains some 90° and 180° pulses embedded in it

ADVANTAGES

- 1, Decreased dephasing due to inhomogeneities in B_0 compared with GRASS and SPGR
- 2, Decreased magnetic susceptibility artifacts compared with GRASS and SPGR
- 3, Decreased chemical shift artifacts

DIS ADVANTAGES

- 1, Decreased SNR secondary to the use of longer TE
- 2, Increased sensitivity to non stationary tissue

ULTRAFAST GRADIENT ECHO SEQUENCE

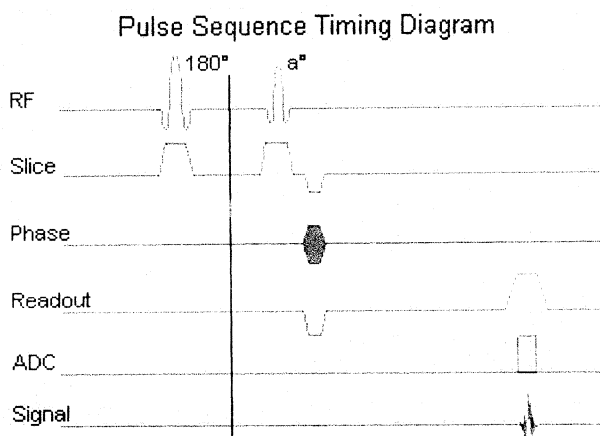
In simple ultrafast GRE imaging, TR and TE are so short, that tissues have a poor imaging signal and — more importantly — poor contrast except when contrast media enhanced (contrast enhanced angiography). Therefore, the magnetization is 'prepared' during the preparation module, most frequently by an initial 180° inversion pulse.

In the pulse sequence timing diagram, the basic ultrafast gradient echo sequence is illustrated. The 180° inversion pulse is executed one time (to the left of the vertical line), the right side represents the data collection period and is often repeated depending on the acquisition parameters.

Ultrafast GRE sequences have a short TR, TE, a low flip angle and TR is so short that image acquisition lasts less than 1 second and typically less than 500 ms. Common TR: 3-5 msec, TE: 2 msec, and the flip angle is about 5° . Such sequences are often labeled with the

prefix "Turbo" like TurboFLASH, TurboFFE and TurboGRASS. This allows one to center the subsequent ultrafast GRE data acquisition around the inversion time TI, where one of the tissues of interest has very little signal as its z-magnetization is passing through zero.

Unlike a standard IR sequence, all lines or a substantial segment of k-space image lines are acquired after a single inversion pulse, which can then together be considered as readout module. The readout module may use a variable flip angle approach, or the data acquisition may be divided into multiple segments ("shots"). The latter is useful particularly in cardiac imaging where acquiring all lines in a single segment may take too long relative to the cardiac cycle to provide adequate temporal resolution. If multiple lines are acquired after a single pulse, the pulse sequence is a type of gradient echo echo planar imaging (EPI) pulse sequence



Balanced Sequence

This family of sequences uses a balanced gradient waveform. This waveform will act on any stationary spin on resonance between 2 consecutive RF pulses and return it to the same phase it had before the gradients were applied. A balanced sequence starts out with a RF pulse of 90° or less and the spins in the steady state. Before the next TR in the slice, phase and frequency encoding direction, gradients are balanced so their net value is zero. Now the spins are prepared to accept the next RF pulse, and their corresponding signal can become part of the new transverse magnetization. If the balanced gradients maintain the longitudinal and transverse magnetization, the result is that both T1 and T2 contrast are represented in the image.

This pulse sequence produces images with increased signal from fluid (like T2 weighted sequences), along with retaining T1 weighted tissue contrast. Because this form of sequence is extremely dependent on field homogeneity, it is essential to run a shimming prior the acquisition.

This sequences include e.g. Balanced Fast Field Echo (bFFE), Balanced Turbo Field Echo (bTFE), Fast Imaging with Steady Precession (TrueFISP), Completely Balanced Steady State (CBASS) and Balanced SARGE (BASG).

Balanced Fast Field Echo(bFFE)

A FFE sequence using a balanced gradient waveform. A balanced sequence starts out with a RF pulse of 90° or less and the spins in the steady state. Before the next TR in the slice phase and frequency encoding, gradients are balanced so their net value is zero. Now the spins are prepared to accept the next RF pulse, and their corresponding signal can become part of the new transverse magnetization. Since the balanced gradients maintain the transverse and longitudinal magnetization, the result is, that both T1 and T2 contrast are represented in the image. This pulse sequence produces images with increased signal from fluid, along with retaining T1 weighted tissue contrast. Because this form of sequence is extremely dependent on field homogeneity, it is essential to run a shimming prior the acquisition. A fully balanced (refocused) sequence would yield higher signal, especially for tissues with long T2 relaxation times.

Balanced SARGE(BASG)

The spoiled steady state acquisition rewinded gradient echo sequence with balanced waveform.

Balanced Turbo Field Echo(BTFE)

A gradient echo pulse sequence with a balanced gradient waveform and data acquisition after an initial preparation pulse for contrast enhancement.

Fast Imaging with Steady Precession(TrueFISP)

True fast imaging with steady state precession is a coherent technique that uses a fully balanced gradient waveform. The image contrast with TrueFISP is determined by $T2^*/T1$ properties and mostly depending on TR. The speed and relative motion insensitivity of acquisition help to make the technique reliable, even in patients who have difficulty with holding their breath. Recent advances in gradient hardware have led to a decreased minimum TR. This combined with improved field shimming capabilities and signal to noise ratio, has allowed TrueFISP imaging to become practical for whole-body applications. There's mostly $T2^*$ weighting. With the used ultra short TR-times $T1$ weighting is almost impossible. One such application is cardiac cine MR with high myocardium-blood contrast. Spatial and temporal resolution can be substantially improved with this technique, but contrast on the basis of the ratio of $T2^*$ to $T1$ is not sufficiently high in soft tissues. By providing $T1$ contrast, TrueFISP could then document the enhancement effects of $T1$ shortening contrast agents. These properties are useful for the anatomical delineation of brain tumors and normal structures. With an increase in SNR ratio with minimum TR, TrueFISP could also depict the enhancement effect in myoma uteri. True FSIP is a technique that is well suited for cardiac MR imaging. The imaging time is shorter and the contrast between the blood and myocardium is higher than that of FLASH.

Reduced k-Space Sampling

. Reduced k-space sampling methods include MR Fluoroscopy, Fourier-encoded keyhole imaging, random k-space sampling methods. Spiral scan imaging and other fast or ultra-fast techniques can be included among these methods if acquisition yields incomplete k-space filling. Also, the Half Excitation method] uses half k-space filling exploiting k-space symmetry. Images acquired using reduced k-space sampling may suffer from reduced spatial resolution, artifacts, low tissue contrast, or be restricted to a small FOV. Reduced k-space sampling methods exploit different imaging algorithms - Fourier spatial encoding ordering strategies - for

temporal efficiency, and, in theory, can utilize a number of fundamental pulse sequence approaches. Also, some of these methods use the strategy of updating, rather than waiting to refill, the k-space image reconstruction buffer so that fast computing hardware can reconstruct and display images (from the mixture of image data in the buffer) in real-time without the delay of the standard acquire-delay-reconstruct-display paradigm.

MR FLUOROSCOPY

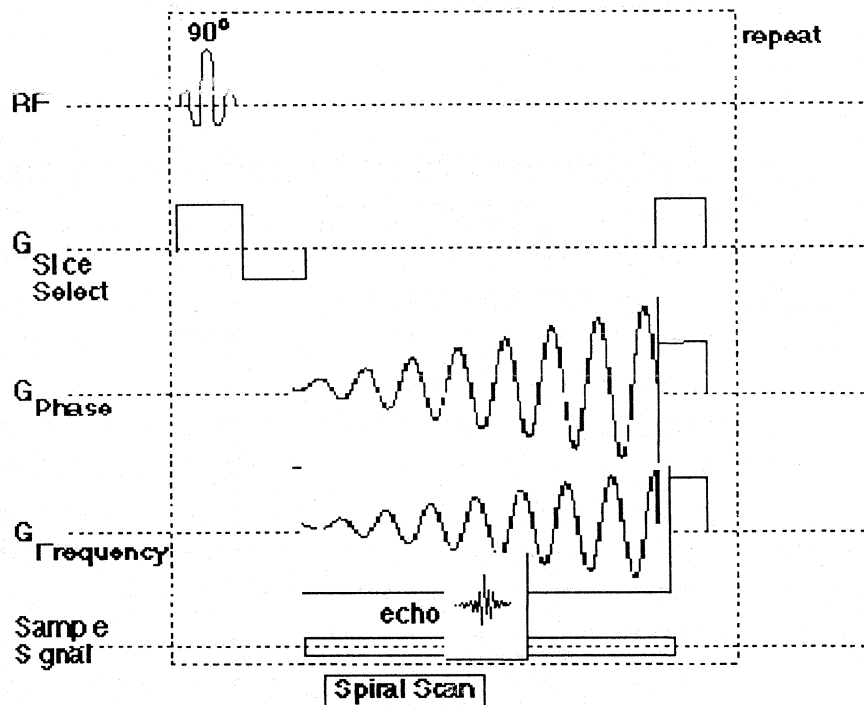
MR fluoroscopy techniques have been available for the last decade but predominantly based on gradient echo or EPI acquisitions providing T1 or proton weighted imaging. They have primarily been applied to cardiac and angiographic imaging. As a result of their limited contrast and spatial resolution, combined with their sensitivity to off resonance artefacts,

MR Fluoroscopy combines a fast MR pulse sequence (GRASS) and an efficient imaging algorithm (partial zero-filling of k-space) with specialized hardware for image reconstruction, and was developed for real-time visualization of dynamic processes. The approach of the MR Fluoroscopy method is to acquire less data for image reconstruction in order to improve temporal resolution in dynamic studies and to create the appearance of real-time image display. Improved temporal resolution is obtained, however, at the expense of reduced spatial resolution since the acquisition of higher spatial frequency information is bypassed in order to more rapidly update low spatial frequency image data

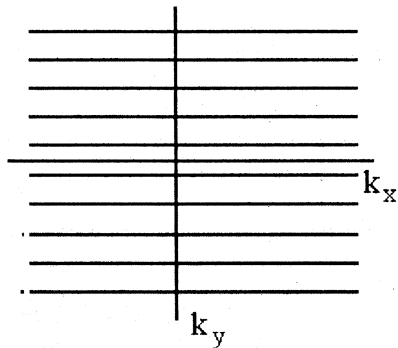
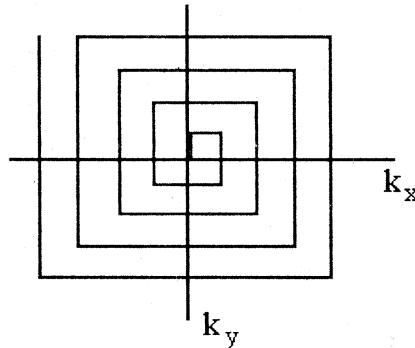
SPIRAL SCANNING,

A way of sampling image data for MR imaging in k space. As opposed to a standard procedure, where k-space is examined by sampling data points linearly, in spiral scanning k-space is scanned in one or several intertwined. Spiral scanning is more demanding on image reconstruction than standard scanning. When moving through k-space in a spiral fashion, the gradient fields are continuously changing in a sinus-like form with decreasing frequency. Data acquisition requires nonlinear sampling. Potential advantages of

spiral scanning are: k-space can be sampled more efficiently, omitting data points at its edge where image information is less important; and image data in the centre of k-space are acquired first, where the gradient strength needed is small and thus flow effects are smaller, i. e. images acquired using spiral scanning are less prone to flow artefacts.



Spiral scanning is an echo planar imaging EPI sequence, as after a single excitation pulse the entire k-space is sampled, entailing the problem that the signal amplitude decays with T_2^* during the measurement. Hence, the signal from image data measured further out on the spiral is underestimated to an unknown and variable degree. This problem can be remedied by using an interleaving or "multishot" data acquisition technique. Where multiple intertwined spirals originating at the centre of k-space are traced consecutively, but with fewer turns per radial distance increment

Conventional k-space**Spiral scan k-space**

ECHO PLANAR IMAGING

This was also invented by Professor Sir Peter Mansfield..

In 2003, he got Nobel prize in medicine

1981-82-the first snap shot EPI image

1983-First EPI images of the brain

EPI is a very fast MR imaging technique. It acquires the entire MR image in a fraction of seconds. In this multiple lines of imaging data are acquired after a single RF pulse. Its advantages over conventional

MR imaging is reduced imaging time, reduced motion artifact, possible to imaging the rapid physiological process of human body

In this technique the whole of k space is sampled in a single FID, so that the total scan time is typically less than 100 ms. This is a major advantage of EPI is, there are no motion artefacts, dynamic processes (such as brain activation) can be scanned, quantitative imaging (measurement of T1 and T2 and other parameters) can be performed in a reasonable time. However the images have a low intrinsic signal to noise ratio (although the signal to noise per unit time is very high, and this is what really matters), The magnetic field gradients have to be switched very rapidly. This means that low inductance gradient coils are required, (and the sequence can be very loud at high field) The images are very sensitive to distortions and signal loss arising from gross variations in the magnetic field across the sample (T_2^* effects). High performance of gradient are needed to allow rapid on and off switching of the gradient

TYPES OF EPI

Two main types exist

1, single-shot EPI

2, multi-shot EPI

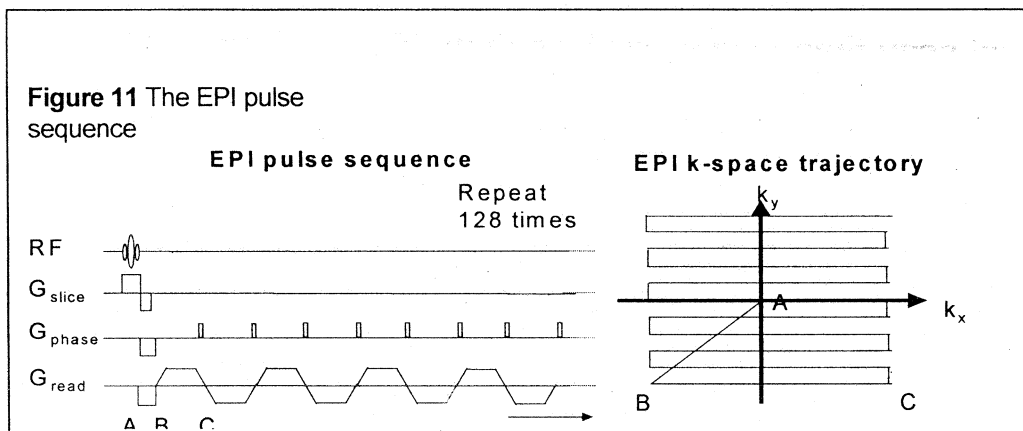
Newer techniques use as blipped phase-encoded gradient

blipped-EPI

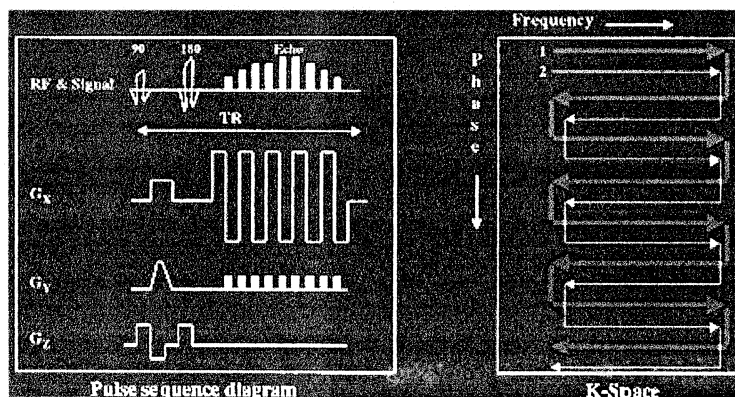
Single-shot EPI

In single-shot EPI all the lines in K-space are filled by multiple gradient reversal, producing multiple echoes in a single RF-pulse. To achieve this, read out gradient must be reversed rapidly from maximum positive to maximum negative value. Each lobe of the read out gradient above or below the base line correspond to a separate K_y line in the k-space. so the number of phase encoding steps N_y is equal to the +ve & -ve lobes of read out gradient. In earlier EPI methods, the phase-encoding gradient was kept on continuously during the acquisition resulting in the zigzag coverage of k-space. This lead to some artifacts during FT compared with conventional k-space trajectories. To rectify this problem, the phase encoding gradient was subsequently applied briefly during the time when the

read out gradient was zero. This method was referred to as blipped phase encoding and this k-space trajectory is much easier on FT.



The susceptibility artifact is more in single shot EPI, and the phase errors tends to propagates along the phase-encode axis,



Multishot- EPI

In multi-shot EPI, the readout is divided into multi-shot EPI or segments (N_s), so that

$$N_y = N_s \times \text{ETL}$$

ETL is echo train length, is the number of lines in each segment. This is also called **segmental EPI**

Advantages of multi-shot EPI (compared with single-shot EPI)

1, this technique places less stress on the gradients compared with single-shot EPI

2, phase errors susceptibility artifact are less

Disadvantages of multi-shot EPI (compared with single-shot EPI)

1, multi-shot EPI takes longer to perform than does single-shot EPI

2, multi-shot EPI is more susceptible to motion artifact

CONTRAST IN EPI

Contrast in EPI depends on the root pulsing sequence. For SE-EPI We are using 90° - 180° -SE root sequences is applied before the EPI module, like that we can produce the gradient echo, IR-sequences, diffusion etc

1, Spin Echo EPI

2, Gradient EPI

Spin Echo EPI

RF refocusing pulses can be applied after the initial excitation pulse. The application of the refocusing pulse helps to clean up some of the artifacts by magnetic field inhomogeneities and chemical shift.

Gradient Echo EPI

This EPI is acquired with a RF pulse followed by a number of gradient blips creating a train of gradient echoes. In this sequence images are acquired in one TR pass in milliseconds.

Gradient EPI is less time than SP EPI

Special Hardware Requirements

gradient strength, rise time, and duty cycle are markedly increased
 gradient coils capable of a maximum amplitude of 20 mT/m

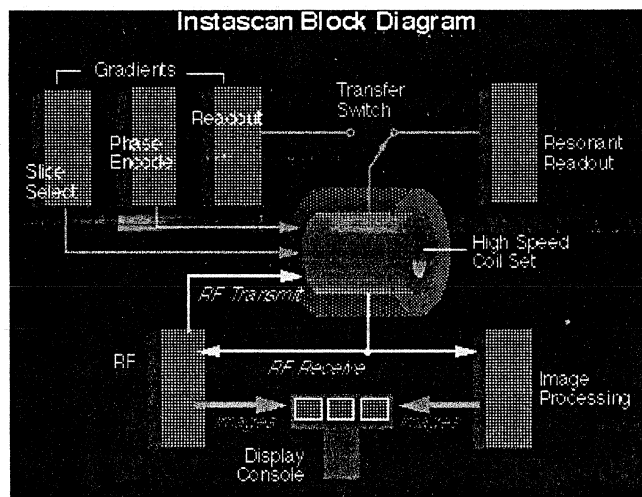
minimum rise time of 0.1 msec

slew rate of 200 T/m per second

duty cycle of 50%–60%

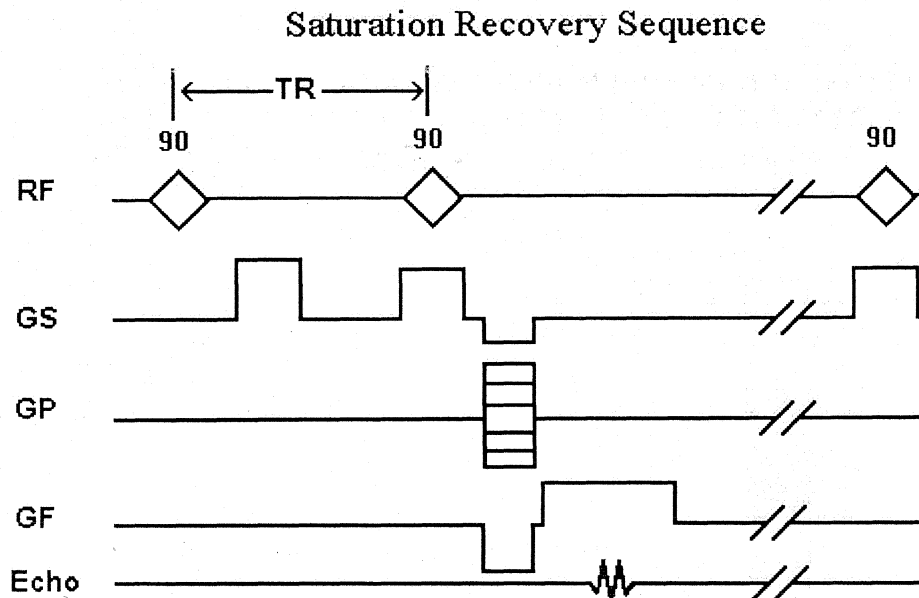
First generation Epi hardware-resonant systems

Second generation Epi hardware-nonresonant systems



Saturation Recovery Sequence

The saturation recovery (SR) sequences are rarely used for imaging now. Their primary use at this time is as a technique to measure T1 times more quickly than an inversion recovery pulse sequence. Saturation recovery sequences consist of multiple 90 degree RF pulses at relatively short repetition times (TR). An example of a SR sequence is shown below. Residual longitudinal magnetization after the first 90-degree RF pulse is dephased by a spoiling gradient (in this case with the slice select gradient). Longitudinal magnetization that develops during the TR period after the dephasing gradient is rotated into the transverse plane by another 90 degree pulse. A gradient echo is acquired immediately after this. The signal will reflect T1 differences in tissues because of different amounts of longitudinal recovery during the TR period.



SMASH (Simultaneous Acquisition Of Special Harmony)

SMASH is such a technique. It uses combinations of component coil signals in a radio-frequency (RF) coil array to substitute for omitted gradient steps, reducing the burden on the gradients and allowing multiple components of the spatial encoding required to generate an MR image to be performed in parallel.¹² This technique has been shown to achieve a four- to eightfold reduction in image acquisition time, with no compromise in spatial resolution.

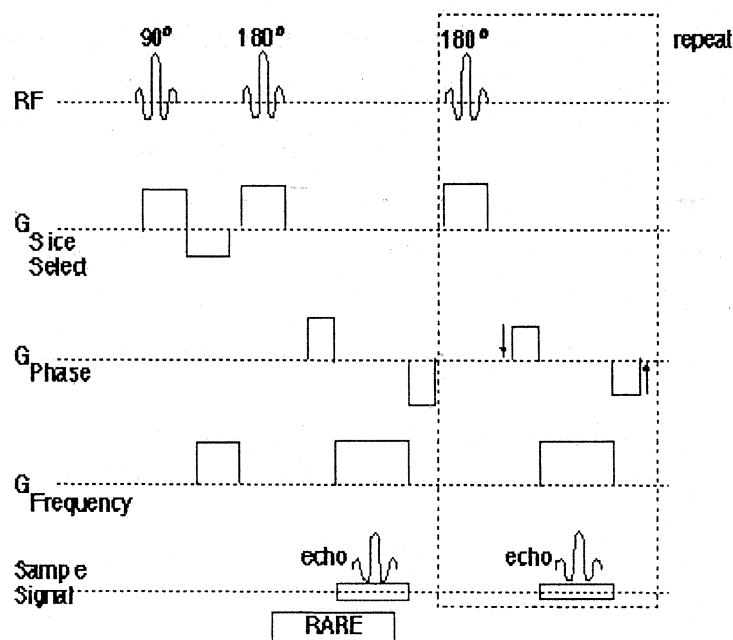
TRICKS

This technique uses an increased sampling rate for lower frequencies, temporal interpolation of k-space views, and zero filling in the slice-encoding direction. When appropriately combined, these elements permit reconstruction of a series of 3-D image sets, having an effective temporal frame rate of one volume every two to six seconds, with no serious compromise in the spatial resolution.

RARE. (Rapid acquisition with relaxation enhancement)

The RARE is a fast spin echo sequence using a series of RF pulses, after an initial P/2 RF excitation pulse, to evoke a train of

multiple individually phase encoded echoes. The RARE method is initiated by a slice-selective P/2 RF excitation pulse applied simultaneously with a gradient in the slice-selective direction, which is inverted after the RF pulse ends for spin rephasing. A short dephasing gradient pulse is then applied in the frequency encoded direction. Next, a P RF pulse is transmitted accompanied by a gradient in the slice-selective direction. A phase encoding gradient pulse follows, after which is applied the readout frequency encoding gradient and the sampling of an echo occurs. Echo acquisition is immediately followed by a rephasing gradient in the phase encoded direction (ie. phase 'unwound'). A subset of sequence events is repeated 4-16 times, including the P RF pulse, slice-selective gradient pulse, phase encoding gradient, readout gradient and rephasing gradient in the phase encoding direction. Optional 'crusher' gradients can be used before and after the slice-selective gradients (simultaneous with the P RF pulse) to improve slice definition. FAISE (Fast acquisition interleaved spin echo) is a hybrid RARE technique with an special phase encode gradient order (ie. special imaging algorithm) that traverses k-space in an interleaved manner giving straightforward T2 contrast manipulation.

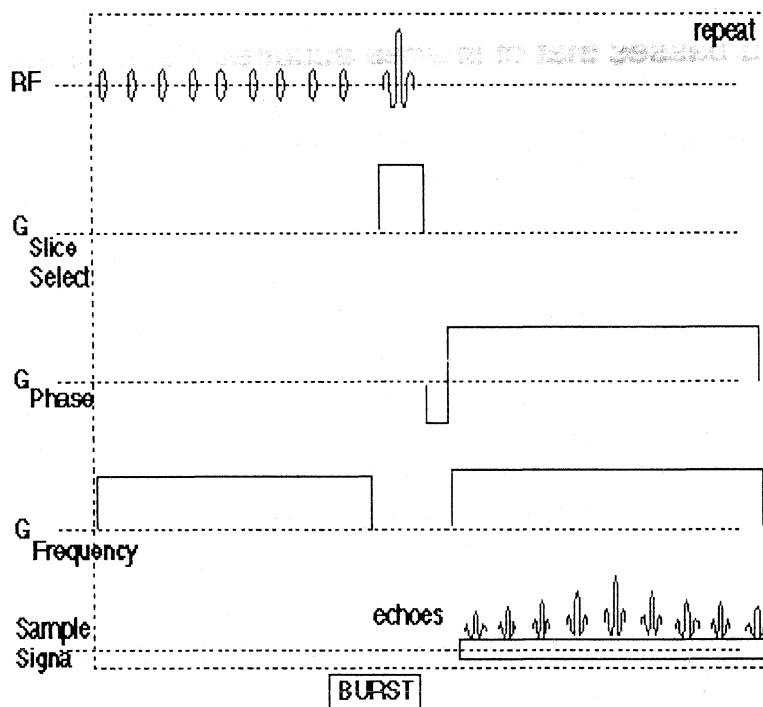


The advantage of fast spin echo methods is that a number of echoes can be acquired in a single TR period yielding a highly efficient method for obtaining proton density-weighted or strongly T2-weighted

images, 3D images [], or for dynamic imaging. The SNR of fast spin echo methods is near that of standard spin echo imaging, and generally surpasses that of images acquired using gradient-recalled echo methods]. Several 256x256 images can be obtained in a multi-slice acquisition in 30-60 sec. Disadvantages of RARE images are the presence of T2-dependent effects along the image phase encoding axis, insensitivity to susceptibility effects and increased RF power deposition (due to the multiple P RF pulses) which can be overcome by using an a degree RF excitation pulse ($\alpha < P$) in place of the repeated P RF pulse

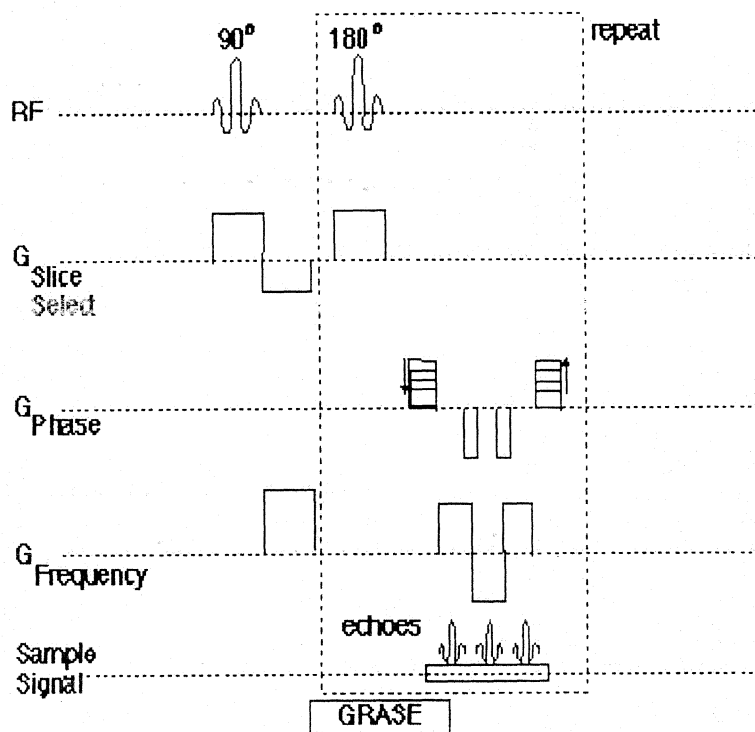
BURST.

The BURST is a ultra-fast imaging method utilizes a rapid sequence of short very low flip angle RF excitations of varying amplitudes to generate multiple echoes accompanied by either constant is an example of a BURST sequence) or switched gradients to improve SNR. BURST sequences can be designed in several ways with either phase and frequency encoding gradients occurring during the BURST RF pulses and/or after the pulse during RF sampling. An image is formed from a small number of repeats of this multi-echo sequence segment with even sub-100 msec acquisition times possible. Related ultra-fast methods referred to as DUFIS (or OUFIS) imaging] have also been described that can similarly acquire images in about 10 msec - similar to BURST in that relatively infrequent gradient switching is required. BURST-type methods have shown promise for rapid 3D imaging. A current disadvantage of BURST-type images is the low SNR.



GRASE.

The gradient- and spin-echo (GRASE) method combines magnetization refocusing using \square RF pulses interleaved with refocusing gradient pulses to form a train of echoes. Using the CPMG method, \square RF refocusing pulses are used to create an echo train, similar to fast spin echo techniques. Three or more gradient-recalled echoes are also then created during the period between each RF echo using a combination of stepped phase encoding gradients and switched frequency encoding gradients. An image is formed from a small number of repeats of this multi-echo sequence segment. (Multi-shot spin echo planar imaging is equivalent to GRASE.) Used in multi-slice 3D imaging GRASE imaging can yield more than 20 T2-weighted 256x256 images in about 30 sec. A disadvantage of GRASE, shared by fast spin echo sequences, is that magnetization remains saturated after a single shot, requiring a delay before the next acquisition. GRASE images are less affected by chemical shift and field inhomogeneities, advantageously more sensitive to magnetic susceptibility differences, and are acquired with reduced RF deposition compared to fast spin echo techniques and is more rapid.



PARALLEL Imaging

Introduction:

Since the development of the NMR phased array in the late 1980s, multicoil arrays have been designed to image almost every part of the human anatomy. These multicoil arrays are primarily used for their increased signal-to-noise ratio (SNR) compared to volume coils or large surface coils. Sensitivity encoding for MRI (SENSE), simultaneous acquisition of spatial harmonics (SMASH), and sensitivity profiles from an array of coils for encoding and reconstruction in parallel (SPACE-RIP) are MRI techniques designed to reduce the scan time. The reduction is achieved by under sampling k-space and recording images simultaneously from multiple imaging coils. Under sampling reduces the acquisition time and the use of multiple RF coils eliminates the wraparound caused by the under sampling.

In this technique the component coils of an array to partially replace spatial encoding which would normally be performed using gradients, thereby reducing imaging time.

In general, higher acceleration factors can be achieved, approaching the number of coil elements.

- Huchinson & Raff (1988) and Kwiat & Einav (1991) suggested using arrays of coils in place of all phase encoding steps.
- Requires one coil for each line of k-space.
- Signals seen by adjacent coils must be sufficiently distinct.
- Kelton et al: "An algorithm for rapid image acquisition using multiple receiver coils," 8th SMRM proc., 1172 (1989).
- Ra & Rim: "Fast imaging using subencoding data sets from multiple detectors," 10th SMRM proc., 1240 (1991); Mag. Res. Med. 30, 143 (1993).
- A set of aliased images are produced by an array of receiver coils using subencoded data.
- These images are resolved to an aliasing-free image by using the distance-dependant sensitivity information for each coil.

Phased Array

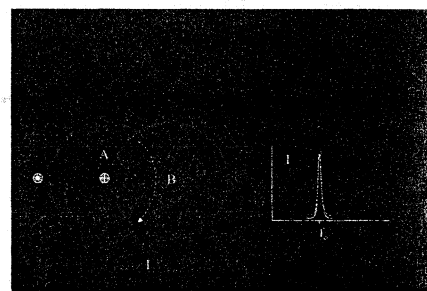
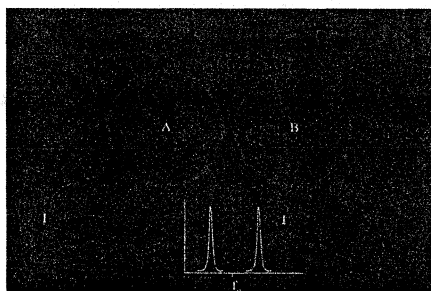
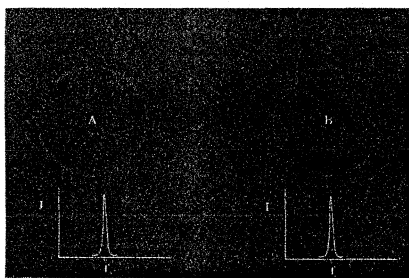
- Phased array is the use of several surface coils.

–Advantages:

- Good SNR
- Large FOV

–Disadvantages:

- Coil coupling - "magic separation" geometry
- Multiple receiver channels (\$)

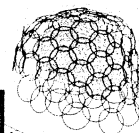


**Extending the phased array to more channels:
23 channel "Bucky" array for 1.5T**



Why stop at 32?
96 Channels on its way...

Receiverarray:
Andreas Potthast



Parallel imaging key points

Aliasing occurs as a result of sub sampling, Parallel imaging algorithms aim at removing aliasing or regenerating the missed k -space lines using the coil sensitivities. Subsampling factor is limited theoretically by the number of coils

Parallel reconstruction techniques

- SENSE (SENSitivity Encoding)
- PILS (Partially Parallel Imaging with Localized Sensitivites)
- SMASH (SiMultaneous Acquisition of Spatial Harmonics)
- GRAPPA (GeneRalized Auto-calibrating Partially Parallel Acquisitions)

- SPACE RIP (Sensitivity Profiles from an Array of Coils for Encoding and Reconstruction In Parallel)

SENSE (SENSitivity Encoding)

Sensitivity encoding (SENSE) is based on the fact that receiver sensitivity generally has an encoding effect complementary to Fourier preparation by linear field gradients. Thus, by using multiple receiver coils in parallel scan time in Fourier imaging can be considerably reduced. The problem of image reconstruction from sensitivity encoded data is formulated in a general fashion and solved for arbitrary coil configurations and k -space sampling patterns. Special attention is given to the currently most practical case, namely, sampling a common Cartesian grid with reduced density. For this case the feasibility of the proposed methods was verified both in vitro and in vivo. Scan time was reduced to one-half using a two-coil array in brain imaging. With an array of five coils double-oblique heart images were obtained in one-third of conventional scan time.

THEORY AND METHODS

In this section SENSE theory is presented and methods for image reconstruction from sensitivity encoded data are derived. The theory addresses the most general case of combining gradient and sensitivity encoding. That is, no restrictions are made as to the coil configuration and the sampling pattern in k -space. Two reconstruction strategies are discussed. The first approach strictly aims at optimal voxel shape and is called *strong* reconstruction for convenience. In *weak* reconstruction, the voxel shape criterion is weaker in favor of the SNR. With both strategies the reconstruction algorithm is numerically demanding in the general case. This is mainly because with hybrid encoding the bulk of the work of reconstruction can usually not be done by fast Fourier transform (FFT). However, it is shown that in weak reconstruction FFT can still be applied if k -space is sampled in a regular Cartesian fashion. For this reason sensitivity encoding with Cartesian sampling is particularly feasible.

Sensitivity Encoding With Cartesian Sampling of k -Space

In two-dimensional (2D) Fourier imaging with common Cartesian sampling of k -space, sensitivity encoding by means of a receiver array permits reduction of the number of Fourier encoding steps. This is achieved by increasing the distance of sampling positions in k -space while maintaining the maximum k -values. Thus scan time is reduced at preserved spatial resolution. The factor by which the number of k -space samples is reduced is referred to as the *reduction factor* R . In standard Fourier imaging, reducing the sampling density results in the reduction of the FOV, causing aliasing. In fact, SENSE reconstruction in the Cartesian case is efficiently performed by first creating one such aliased image for each array element using discrete Fourier transform (DFT). The second step then is to create a full-FOV image from the set of intermediate images. To achieve this one must undo the signal superposition underlying the fold-over effect. That is, for each pixel in the reduced FOV the signal contributions from a number of positions in the full FOV need to be separated. As depicted in Fig. 1, these positions form a Cartesian grid corresponding to the size of key to signal separation lies in the fact that in each single-coil image signal superposition occurs with different weights according to local coil sensitivities. Consider one pixel in the reduced FOV and the

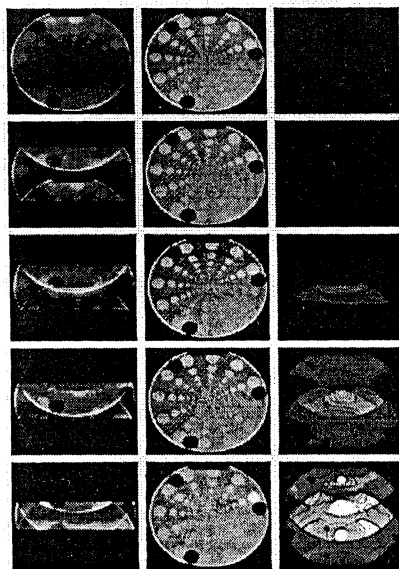
corresponding set of pixels in the full FOV (Fig. 1). Let nP denote the number of pixels superimposed and nC the number of coils used. Assemble in the vector \mathbf{a} the complex image values the chosen pixel has in the intermediate images. The complex coil sensitivities at the nP superimposed positions form an $nC \times nP$ sensitivity matrix S :

$$\mathbf{v} = U\mathbf{a},$$

where the resulting vector \mathbf{v} has length nP and lists separated pixel values for the originally superimposed positions. By repeating this procedure for each pixel in the reduced FOV a non-aliased full-FOV image is obtained.

$$SNR_{\rho}^{red} = \frac{SNR_{\rho}^{full}}{g_{\rho} \sqrt{R}}.$$

This relation confirms an upper bound for SNR characterized by the square root of the number of samples acquired. The geometry factor describes the ability with the used coil configuration to separate pixels superimposed by aliasing. In practice it allows a priori SNR estimates and provides an important criterion for the design of dedicated coil arrays



1 2 3 4 5 6 7 8 9 10 >10

SENSE imaging of a quality phantom with increasing reduction factor R indicated on the left. Phase encoding in vertical direction. Left: conventional sum-of-squares images. Middle: SENSE reconstruction from the same data. Right: maps of the relative noise level as predicted by SENSE theory, colored according to the gray-scale on the far right (arbitrary units).

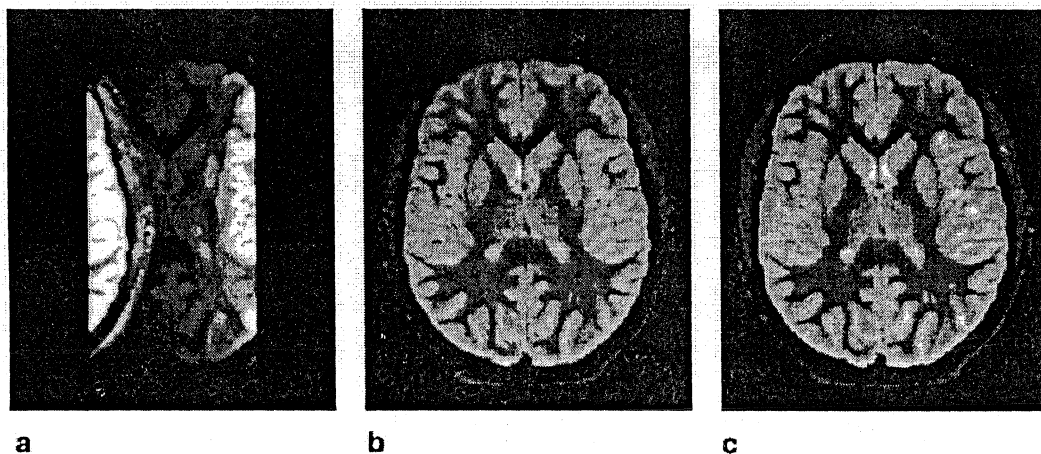


FIG. 7. Transverse brain images obtained with TSE using two coils. a, b: Reduction factor $R=2.0$, conventional single-coil image and SENSE reconstruction. c: SENSE image from fully Fourier encoded data.

SMASH (SiMultaneous Acquisition of Spatial Harmonics)

1997 SMASH is introduced. The SMASH technique exploits sensitivity variations in a surface coil array to substitute for spatial modulations normally produced by phase-encoding gradients.¹¹ This use of coil encoding in place of gradient encoding allows the whole of kspace to be traversed using a reduced number of phase encoding gradient steps, thereby reducing image acquisition times. The function of phase-encoding gradients is to impose sinusoidal modulations of magnetization across the image plane. The MR signal integrated against these sinusoids then corresponds to spatial Fourier components of the image, or the familiar k-space lines. In the SMASH technique, some of these sinusoidal modulations, or 'spatial harmonics', are generated by manipulations of component coil sensitivities, rather than by gradient-induced modulations of magnetization.

An array of rf coils contains spatial information in the form of its component coil sensitivities (Figure 2). In a linear surface coil array with adjacent components, each coil j has a distinct but overlapping sensitivity $C_j(x, y)$. By forming appropriate linear combinations of component coil signals (Figure 3), we may generate composite sensitivity profiles C^{tot} which oscillate in

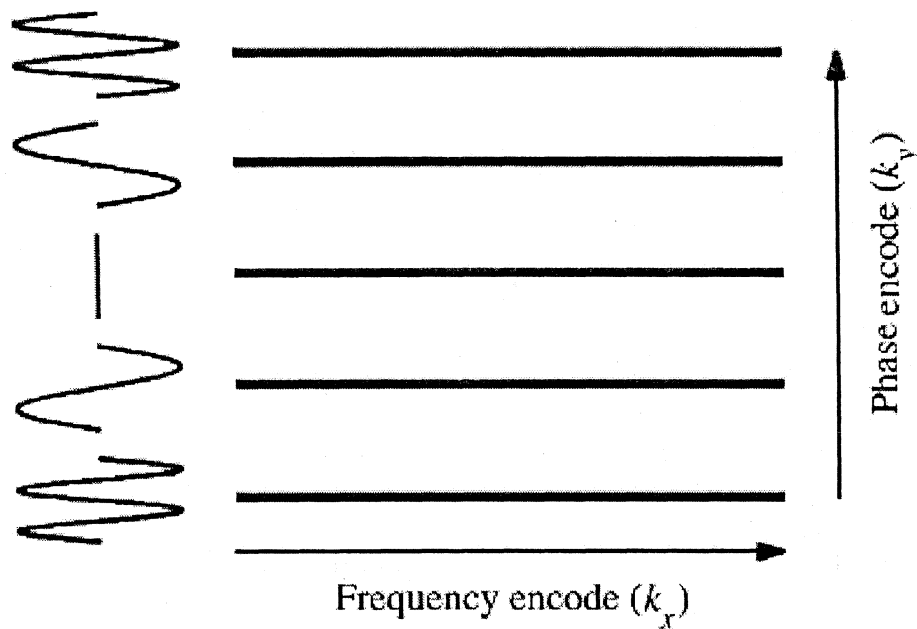


Figure 1 A k -space schematic, indicating the spatial modulations resulting from phase-encoding gradients. Gradient steps on either side of the central $k=0$ line correspond to various harmonics of spin modulation across the image plane

SPATIAL ENCODING USING MULTIPLE RF COILS: SMASH IMAGING AND PARALLEL MRI 3

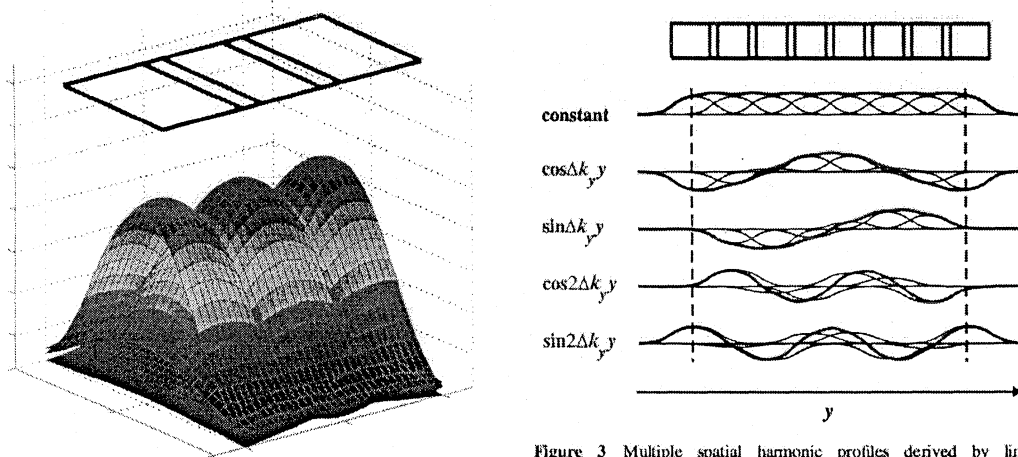


Figure 3 The sensitivity profiles of an eight-element coil array. Each of the three

Figure 3 Multiple spatial harmonic profiles derived by linear combination of component coil sensitivities in an eight-element array

Figure 2 The rf sensitivities of a linear coil array. Each of the three overlapping sensitivity distributions corresponds to one of the three component coils in the array pictured at the top. (The conductor paths of adjacent component coils are also typically overlapped, as shown in this example, to minimize inductive coupling between coils.) Only sensitivity magnitudes are shown here. In practice, coil sensitivities may also have spatially varying phase distributions

Implementation

An important first step in a practical SMASH implementation is to measure the rf sensitivities of the various array elements. These sensitivities may be extracted in a straightforward manner from images of homogeneous phantoms, since any intensity variations in such images may be traced to variations in coil sensitivity. Typically, intensity profiles across an image plane of interest are extracted from a stored phantom data set and are used as sensitivity references.

4 SPATIAL ENCODING USING MULTIPLE RF COILS: SMASH IMAGING AND PARALLEL MRI

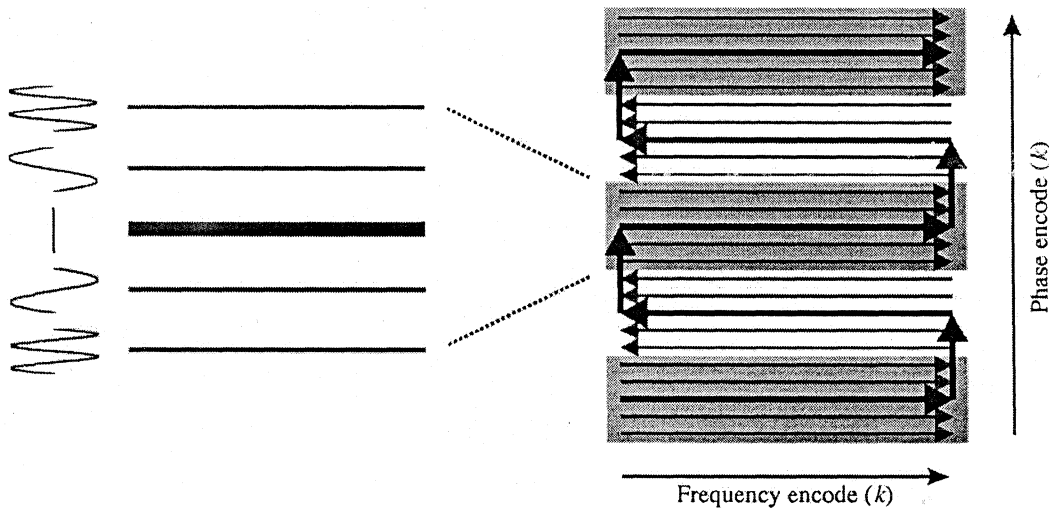


Figure 4. Schematic k -space trajectory for a partially parallel acquisition using first spatial harmonics. Thick lines represent k -space lines corresponding to applied phase-encoding gradient steps, while thin lines represent additional k -space lines reconstructed using the linear combinations shown in Figure 3. The additional reconstructed lines substitute for omitted phase-encoding gradient steps.

AUTO-SMASH,

a small number of reference k -space lines are added to the acquisition, and the relation between these reference lines and the usual MR signal data lines are used to 'train' SMASH reconstructions

In AUTO-SMASH, a small number of reference k-space lines are added to the acquisition, and the relation between these reference lines and the usual MR signal data lines are used to 'train' SMASH reconstructions directly in k-space.

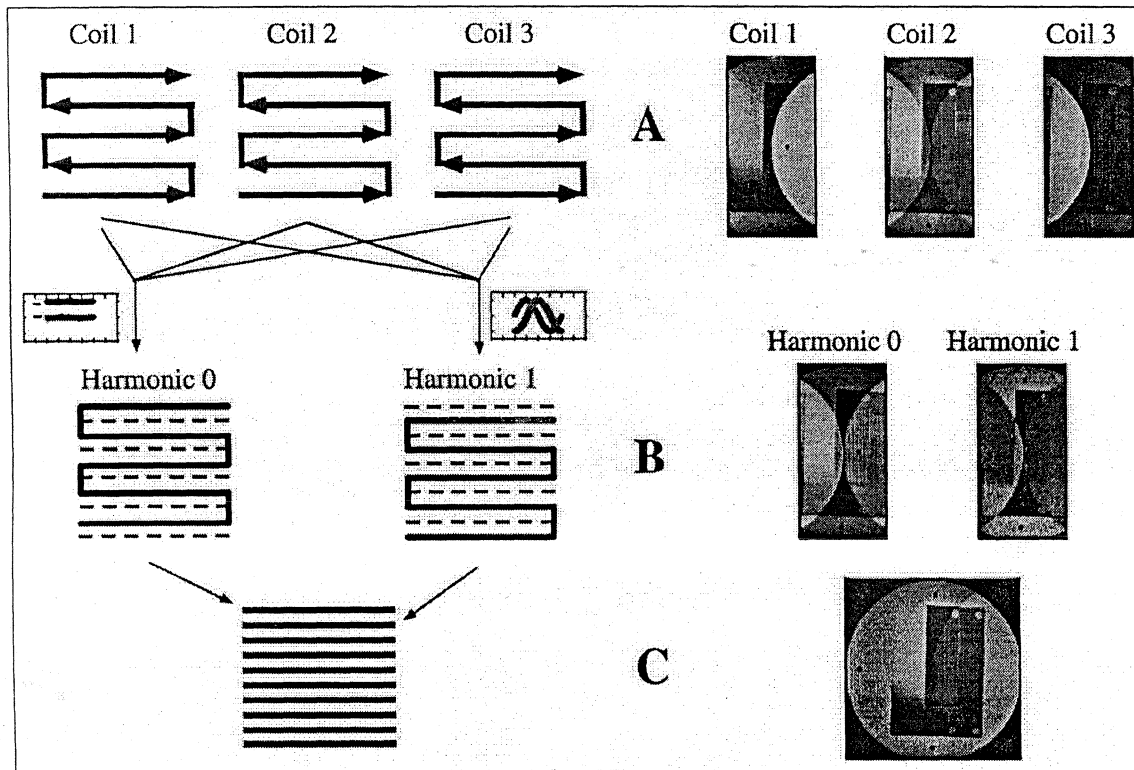
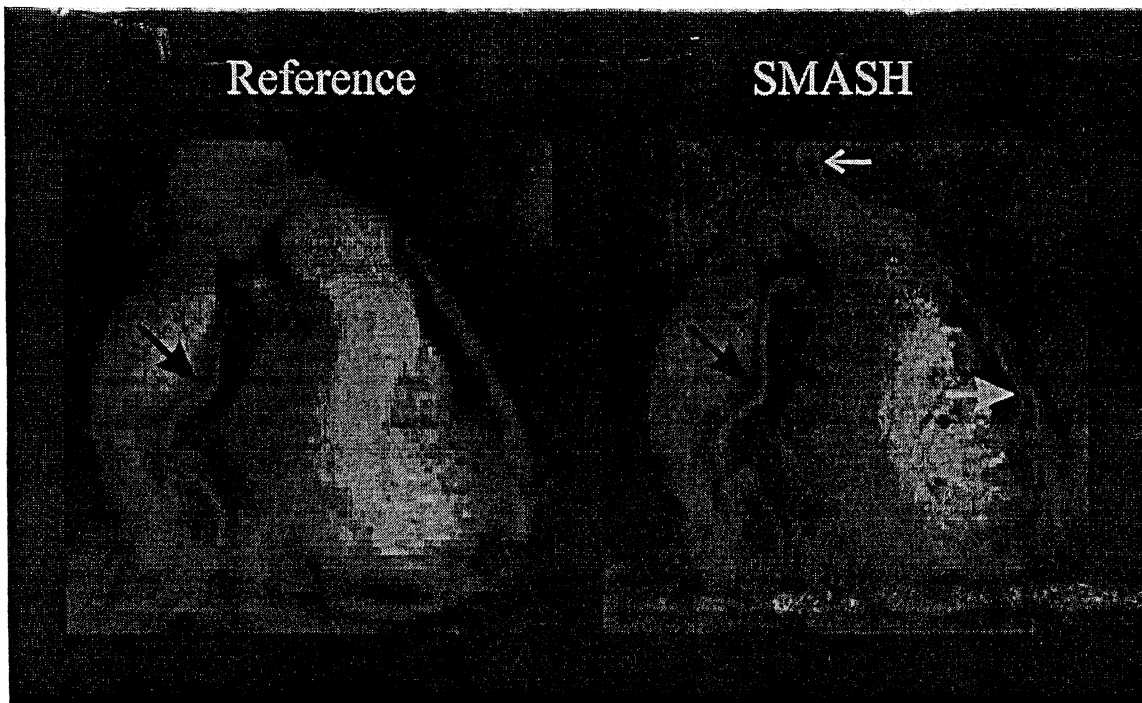
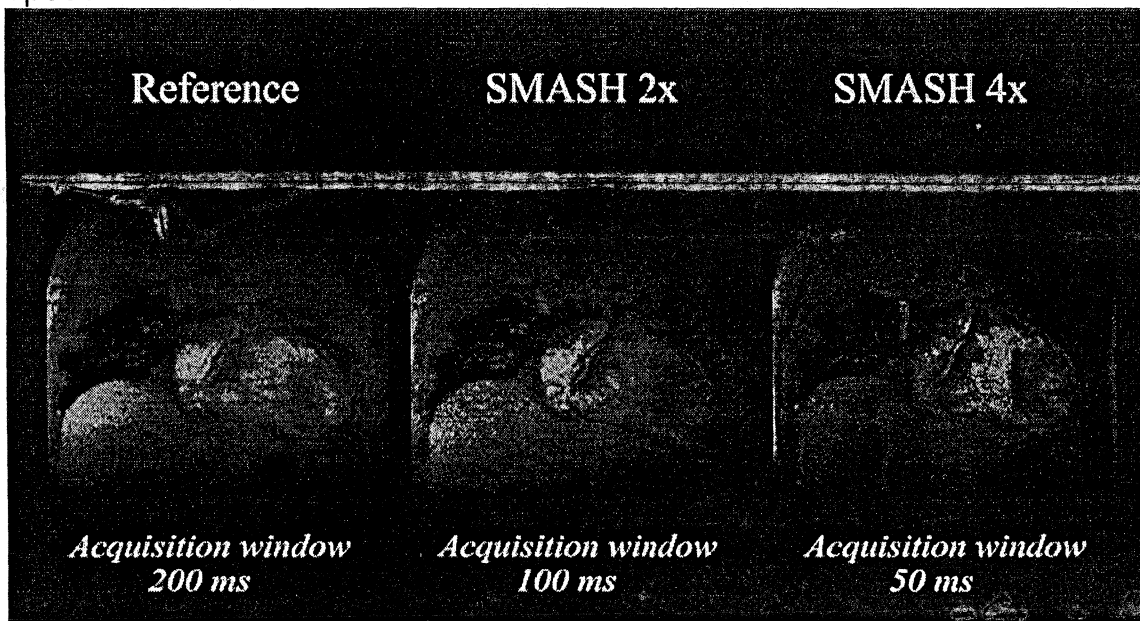


Figure 5 Schematic representation of the SMASH reconstruction procedure (left: k-space cartoon, right: corresponding phantom images). (A) Acquisition of data with reduced phase encoding. (B) Formation of shifted data sets using spatial harmonic combinations. (C) Interleaving of shifted data sets to generate a full signal matrix, corresponding to an image with full FOV



Coronal cardiac images showing the use of SMASH for increased spatial resolution



Coronal cardiac images from a 3D data set demonstrating the use of SMASH to increase temporal resolution at fixed spatial resolution

VD-AUTO-SMASH Imaging

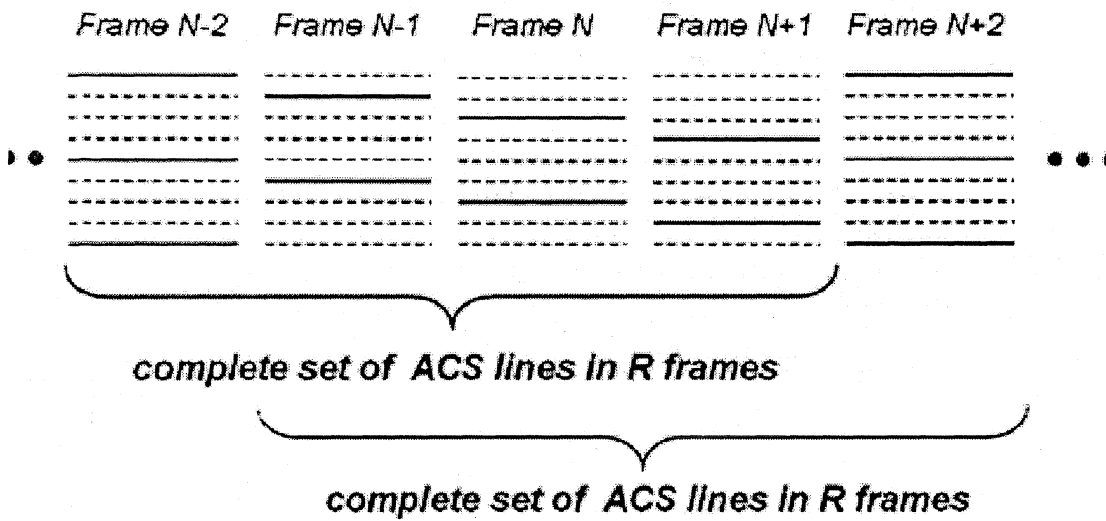
This method uses a VD k -space sampling approach and shows the ability to improve the image quality without significantly

increasing the total scan time. This new k -space adapted calibration approach is based on a k -space-dependent density function. In this scheme, fully sampled low-spatial frequency data are acquired up to a given cutoff-spatial frequency. Above this frequency, only sparse SMASH-type sampling is performed. On top of the VD approach, advanced fitting routines, which allow an improved extraction of coil-weighting factors in the presence of noise, are proposed. It is shown in simulations and in vivo cardiac images that the VD approach significantly increases the potential and flexibility of rapid imaging with AUTO-SMASH.

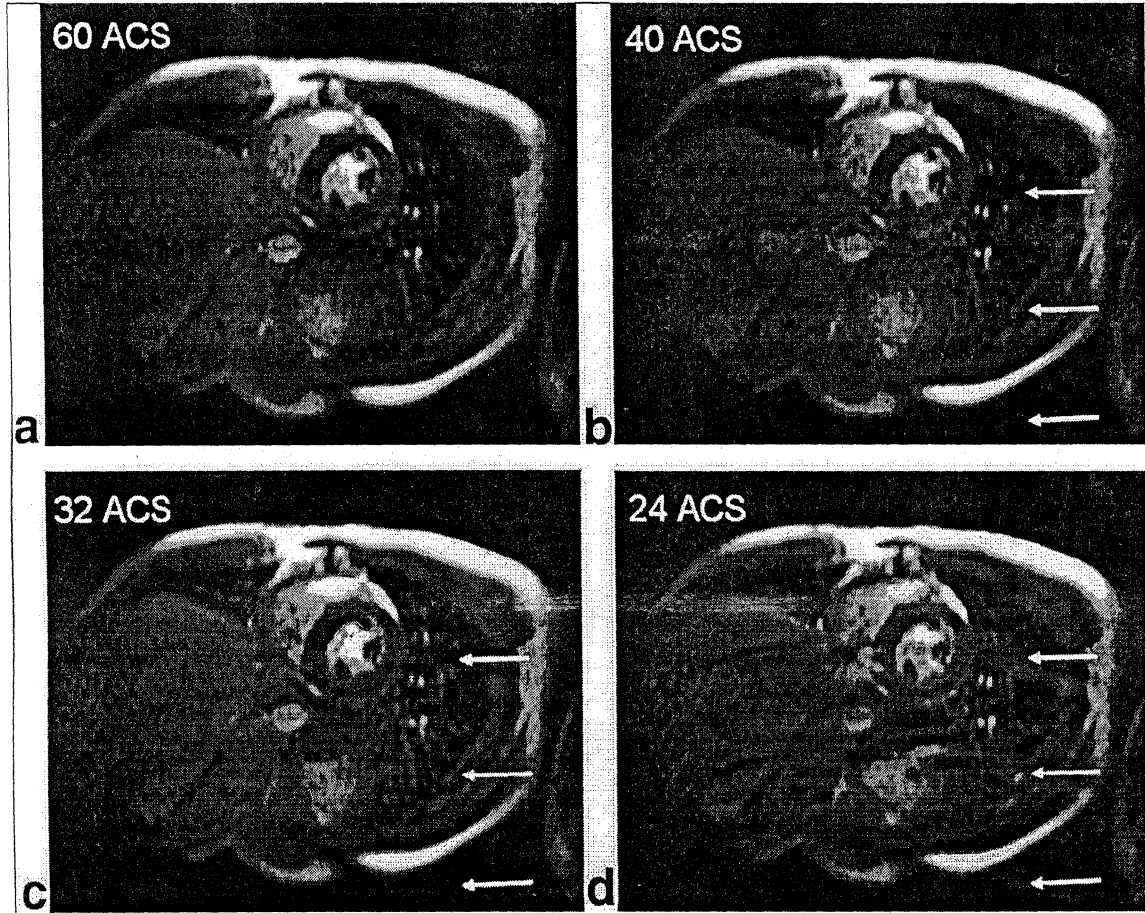
GRAPPA

Current parallel imaging techniques for accelerated imaging require a fully encoded reference data set to estimate the spatial coil sensitivity information needed for reconstruction. In dynamic parallel imaging a time-interleaved acquisition scheme can be used, which eliminates the need for separately acquiring additional reference data, since the signal from directly adjacent time frames can be merged to build a set of fully encoded full-resolution reference data for coil calibration. In this work, we demonstrate that a time-interleaved sampling scheme, in combination with autocalibrated GRAPPA

The GRAPPA parallel imaging reconstruction is performed in k -space by calculating the missing k -space lines in each coil in the array using a weighted sum of adjacent lines from all coils.



In general, more neighboring frames can be averaged to increase the SNR of the ACS (autocalibration signal) data, resulting in a potentially improved GRAPPA reconstruction.



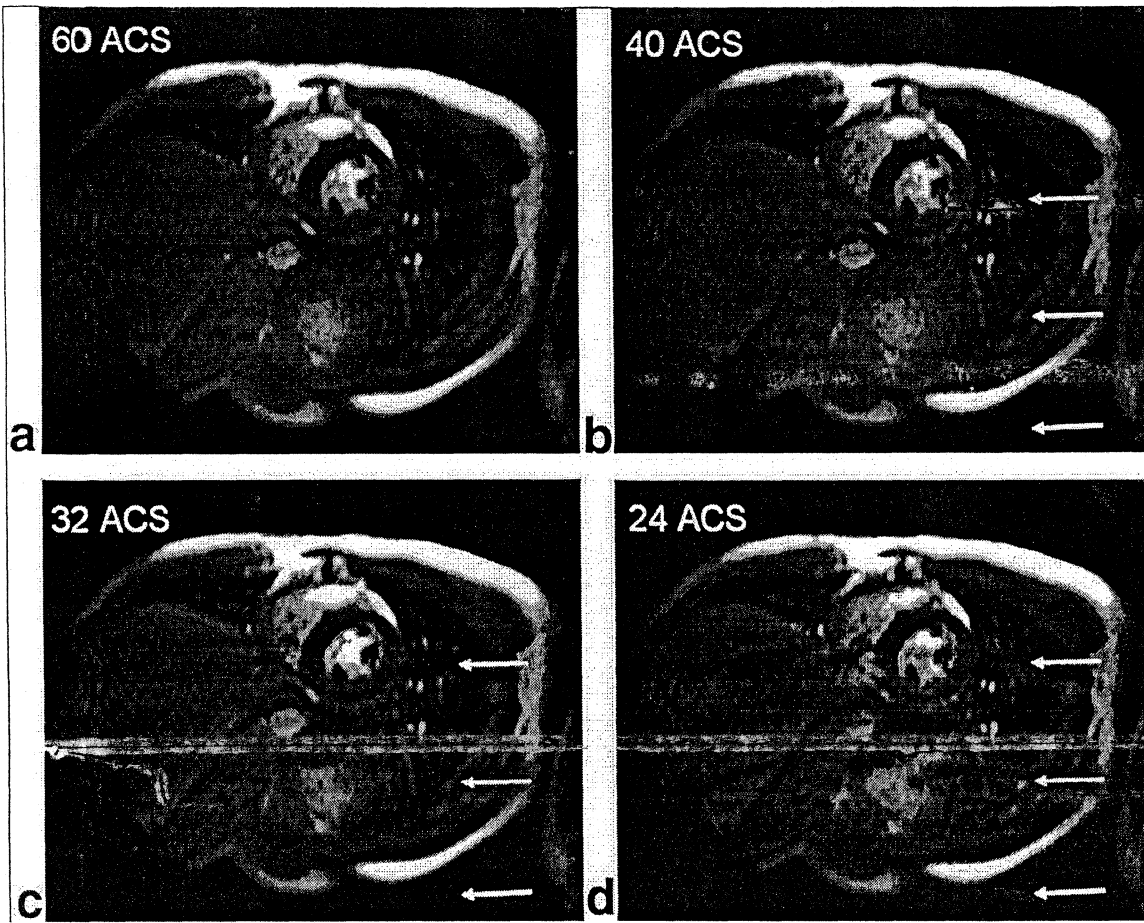


FIG. 4. Rate 4 undersampled example images after GRAPPA reconstruction using (a) 60 ACS lines (full spatial resolution), (b) 40 ACS lines, (c) 32 ACS lines, and (d) 24 ACS lines as reference data. The image quality improves significantly with increased number of ACS lines. Residual aliasing artifacts are visible even with 40 ACS lines as reference data.

MAGNETIZATION TRANSFER CONTRAST IMAGING

Magnetization transfer (MT) is a unique contrast mechanism in magnetic resonance (MR) imaging that has been known for the past decade. Magnetization transfer contrast (MTC) is most useful in two basic areas, improving image contrast and tissue characterization. MTC has also proven to be extremely useful in the reduction of

background signal in MR angiography and improves the appreciation of tissue enhancement by intravenous contrast agents.

Basis of Magnetization Transfer

The signal obtained in clinical MR imaging comes from mobile protons. These are largely present in free water with a smaller contribution from lipid. There are, in addition, a large number of protons in tissues contained in macromolecules or in water bound to these macromolecules. No signal is normally detected from these protons by standard clinical imaging techniques because they have a very short T2 value (of the order of 1 ms or less) and any transverse magnetization is rapidly dephased before data collection is possible. There is, however, a constant exchange of magnetization between the protons in these two pools, the free pool and the bound pool, which occurs by through-space dipole-dipole interactions or probably less importantly direct chemical exchange. By means of this interchange, the bound pool influences the signal obtained from the free pool even though the bound pool cannot be directly visualized. In MT imaging, the normal equilibrium between free and bound pools is perturbed and the resulting contrast is referred to as magnetization transfer contrast and indicates the exchange processes with the bound pool in that particular tissue. Wolff and Balaban, first produced in vivo images with MTC and described two-pool concept of magnetization transfer. They coined the terms free (H_f) and restricted (H_r) proton pools to describe the exchange compartments. These authors perturbed the normal equilibrium between free and bound pools by selectively saturating the bound pool (Fig. 1).

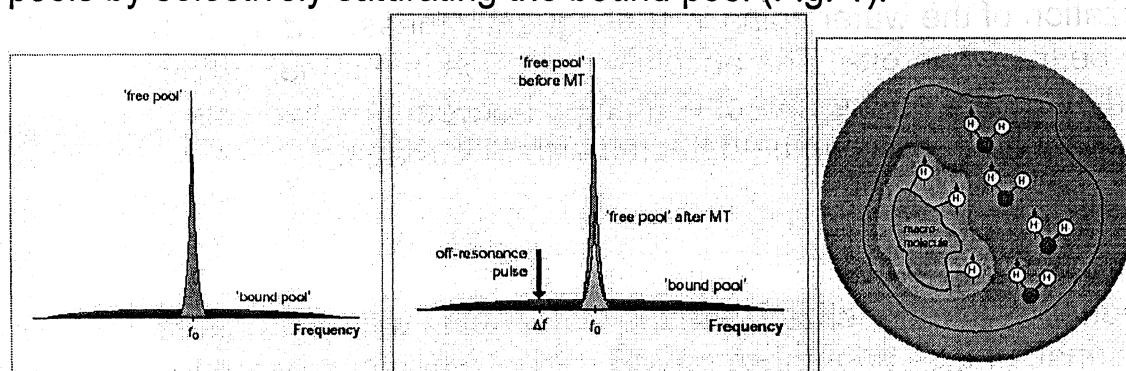
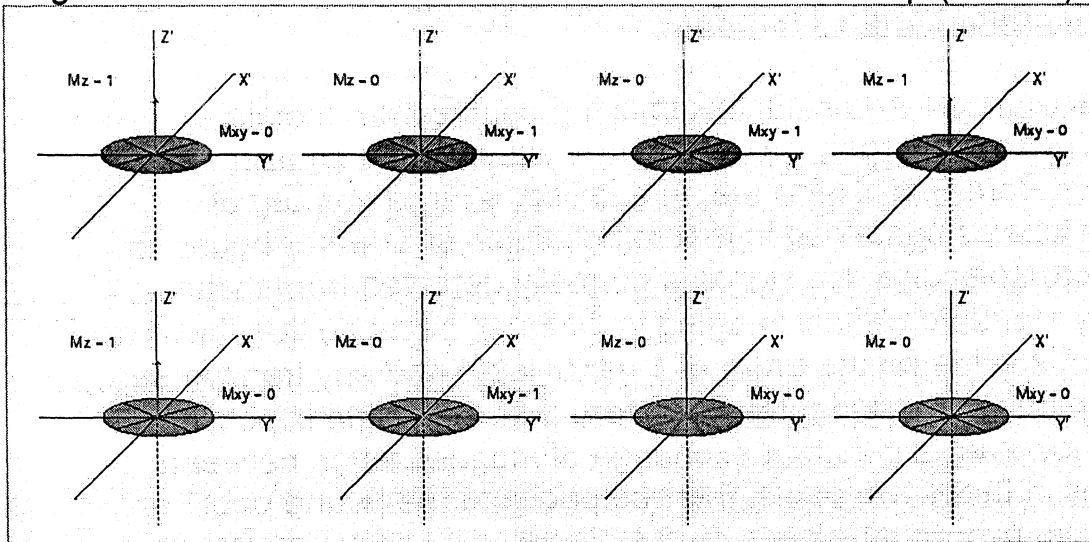


Figure: Schematic diagram of water and macromolecule proton linewidths (top), and the use of an off-resonance pulse to detect magnetization transfer in tissue where both lines overlap (bottom)



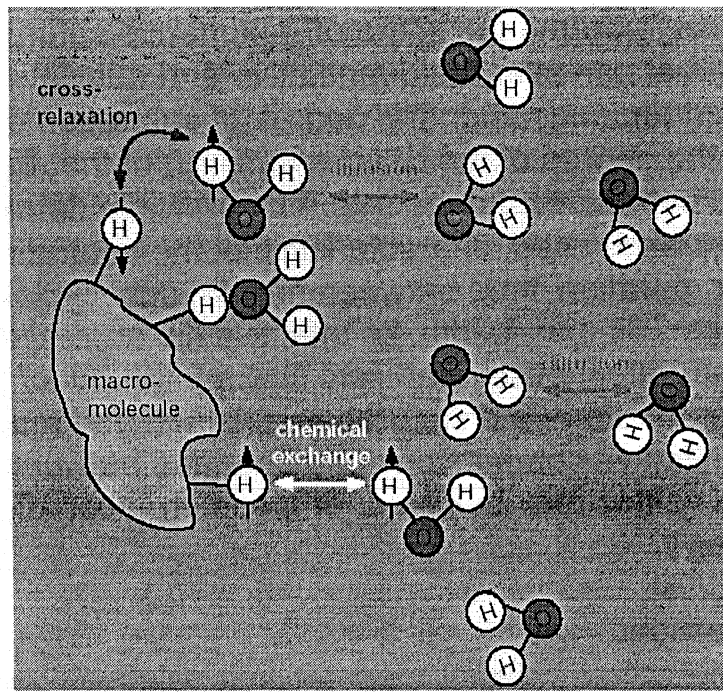
Selective saturation:

It is possible to saturate the macromolecular spins (i.e. to reduce their magnetization to zero) preferentially using radio frequency pulse. The restricted motion of protons results in a very short spin-spin relaxation rate (T_2). This results in a very broad absorption line shape (20-40 kHz) than the mobile spins (narrow water resonance peak of 15Hz), making them as much as 106 times more sensitive to an appropriately placed radio frequency pulse. The saturated macromolecular spins, with zero magnetization will exchange for water spins having magnetization of one. Thus, when the magnetization of the water spins is subsequently measured, it will be found to be less than one. The absolute reduction in magnetization will clearly depend, on the rate of exchange between the two spin populations, and hence can be detected with MR imaging.

MT Contrast:

The reduction of magnetization in the water spins will be manifest as less signal, or less brightness on MR image, when compared to the "control" image obtained without the MT preparation. The decrease will be larger in regions where exchange of magnetization is more efficient, determined by the relative proportion of the hydrogen

atoms of the two pools, their intrinsic relaxation times and the exchange rate. Although saturation is never perfectly selective in vivo, contrast between areas exhibiting varying degrees of MT effect is developed and superimposed upon the intrinsic contrast of the baseline image, be it proton-density-weighted(W) image, T1-W image, or some combination.



Saturation transfer techniques:

Three techniques for saturating the bound water have been studied: (a) off-resonance (radio frequency pulses applied at a frequency that is offset from the "free" water resonance) continuous wave excitation (3), (b) off-resonance shaped pulses, and (c) on-resonance binomial pulses[4,5]. The off-resonance shaped pulses are now the commonest saturation transfer techniques for clinical imaging.

Magnetization transfer ratio:

It is a quantitative measure of the MT effect on tissues. It is the degree of signal suppression of a given tissue compared with the conventional PD or T1-W image. The MT ratio (MTR) may be simply obtained by collecting a pair of identical images (PD or T1-W), one with and one without MT saturation.. For each region of interest (ROI), MTR is calculated from the two images using the formula:

$$\text{MTR} = \left\{ \frac{1 - M_s}{M_o} \right\} \times 100\%$$

Where M_o , M_s represent the signal intensity with the saturation pulse off and on, respectively.

Controlling parameters in saturation transfer:

There are several key parameters in the off-resonance saturation transfer technique that can alter the quantitative and qualitative MTC. The MTR measured depends upon the degree of saturation of the bound water, and the degree of the direct saturation of the signal of the mobile protons and the exchange rate. The principle determinants being the base sequence used, characteristics of the saturating pulse (pulse type, power, duration, time between pulses, duty cycle i.e. number of pulses per TR, bandwidth, its effective flip angle), offset frequency of the saturating pulse, and field strength [6]. All of these characteristics alter the degree of saturation obtained and must be known before quantitative comparisons can be made between clinical studies. Numerical values may be given, or the information displayed as a difference image.

Cross relaxation:

dipole-dipole interactions

Although the magnetization transfer mechanism is relatively well understood at the tissue level, much less detailed knowledge is available on a cellular and molecular level. Continued advances in modeling of magnetization transfer effects will improve the detailed picture of the mechanism underlying this phenomenon.

FMR IMAGING

Functional Magnetic Resonance Imaging (fMRI)

INTRODUCTION

The recent discovery that magnetic resonance imaging can be used to map changes in brain hemodynamics that correspond to mental operations extends traditional anatomical imaging to include maps of human brain function. The ability to observe both the structures and also which structures participate in specific functions is due to a new

technique called functional magnetic resonance imaging, fMRI, and provides high resolution, noninvasive reports of neural activity detected by a blood oxygen level dependent signal (Ogawa, et al, 1990 a and b, 1992, 1993; Belliveau, et al, 1990, 1991). This new ability to directly observe brain function opens an array of new opportunities to advance our understanding of brain organization, as well as a potential new standard for assessing neurological status and neurosurgical risk.

WHAT IS fMRI

Functional MRI is based on the increase in blood flow to the local vasculature that accompanies neural activity in the brain. This results in a corresponding local reduction in deoxyhemoglobin because the increase in blood flow occurs without an increase of similar magnitude in oxygen extraction (Roy and Sherrington, 1890; Plum, Posner & Troy, 1968; Posner, Plum & Poznak, 1969; Fox and Raichle, 1985). Since deoxyhemoglobin is paramagnetic, it alters the T2* weighted magnetic resonance image signal (Ogawa, et al, 1990a and b, 1992, 1993; Belliveau, et al, 1990, 1991; Turner, et al, 1991; Tank, et al, 1992). Thus, deoxyhemoglobin is sometimes referred to as an endogenous contrast enhancing agent, and serves as the source of the signal for fMRI. Using an appropriate imaging sequence, human cortical functions can be observed without the use of exogenous contrast enhancing agents on a clinical strength (1.5 T) scanner (Bandettini, et al, 1992, 1993; Kwong, et al, 1992; and Turner, et al, 1993; Schneider, et al, 1993). Functional activity of the brain determined from the magnetic resonance signal has confirmed known anatomically distinct processing areas in the visual cortex (Belliveau, et al, 1991; Ogawa, et al, 1992; Blamire, et al, 1992; Schneider, et al, 1993; Hirsch, et al, 1995), the motor cortex (Kim, et al, 1993a; Kim, et al, 1993b), and Broca's area of speech and language-related activities (Hinke, et al, 1993, Kim, et al, 1995). Further, a rapidly emerging body of literature documents corresponding findings between fMRI and conventional electrophysiological techniques to localize specific functions of the human brain (Atlas, et al, 1996; Puce, et al, 1995; Burgess, 1995; Detre, et al, 1995; George, et al, 1995; Ives, et al, 1993). Consequently, the number of medical and research centers with fMRI capabilities and investigational programs continues to escalate.

The main advantages to fMRI as a technique to image brain activity related to a specific task or sensory process include 1) the signal does not require injections of radioactive isotopes, 2) the total scan time required can be very short, i.e., on the order of 1.5 to 2.0 min per run (depending on the paradigm), and 3) the in-plane resolution of the functional image is generally about 1.5 x 1.5 mm although resolutions less than 1 mm are possible. To put these advantages in perspective, functional images obtained by the earlier method of positron emission tomography, PET, require injections of radioactive isotopes, multiple acquisitions, and, therefore, extended imaging times. Further, the expected resolution of PET images is much larger than the usual fMRI pixel size. Additionally, PET usually requires that multiple individual brain images are combined in order to obtain a reliable signal. Consequently, information on a single patient is compromised and limited to a finite number of imaging sessions. Although these limitations may serve many neuroscience applications, they are not optimally suitable to assist in a neurosurgical or treatment plan for a specific individual.

What are some common uses of fMRI?

fMRI is becoming the diagnostic method of choice for learning how a normal, diseased or injured brain is working, as well as for assessing the potential risks of surgery or other invasive treatment of the brain. The term "Functional MRI" (with a capital F) can also include other MRI techniques that are sensitive to physiologic changes (such as changes in water motion), whereas fMRI with a lower case f usually refers to the mapping of brain activity using MRI. As a group, these "Functional MRI" techniques appear to provide the most sensitive method currently available for identifying, investigating and monitoring brain tumors, strokes and certain chronic disorders of the nervous system such as multiple sclerosis. In addition, these methods appear to provide a useful means of documenting some brain abnormalities related to dementia or seizures.

In routine practice, fMRI studies are often used in planning brain surgery, since they can help physicians monitor normal brain function as well as any disturbed brain function. While research is still ongoing, it appears that fMRI can also help assess the effects of stroke, trauma or degenerative disease (such as Alzheimer's) on brain function.

How should I prepare for the procedure?

Since fMRI uses an MRI device, the standard preparations for an MRI procedure are necessary. For example, because the strong magnetic field used for MRI will pull on any ferromagnetic metal object implanted in the body, MRI staff will ask whether you have a heart pacemaker (or artificial heart valve), implanted port, infusion catheter (brand names Port-o-cath, Infusaport, Lifeport), intrauterine device (IUD) or any metal plates, pins, screws or surgical staples in your body. In most cases, surgical staples, plates, pins and screws pose no risk during MRI. Red dyes used in tattoos and permanent eyeliner may contain metallic iron oxide and could heat up during MRI, however this is rare. You will be asked if you have ever had a bullet or shrapnel in your body, or ever worked with metal. If there is any question of metal fragments, you may be asked to have an x-ray that will detect any such metal objects. Tooth fillings are not affected by the magnetic field, but they may distort images of the facial area or brain, so the radiologist should be made aware of them. The same is true of braces, which may make it hard to "tune" the MRI unit to your body. You will be asked to remove anything that might degrade MRI images of the head, including hairpins, jewelry, eyeglasses, hearing aids and any removable dental work.

The radiologist or an assistant may ask about drug allergies and whether head surgery has been done in the past. If you might be pregnant, this should be mentioned. Less than one in 20 patients who undergo MRI in an enclosed unit may feel confined or claustrophobic

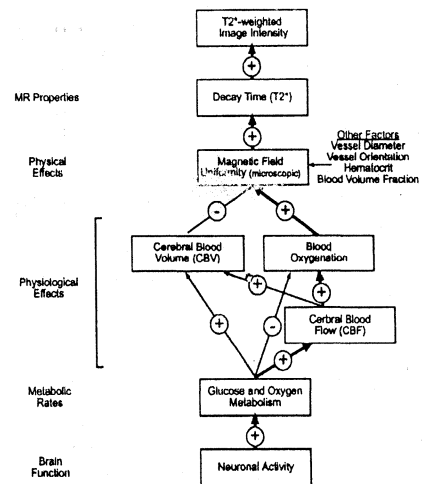
Task Procedure: Patients and subjects are positioned in the scanner as for a conventional scan, and plane lines are set based on conventional imaging methods. During a typical functional imaging series, 30 images are acquired in a 90 sec run where the initial and last 10 images are baseline conditions and the middle 10 images (30 secs) are acquired during a task. For example, in the case of a typical task designed to identify eloquent brain tissue involved in hand and finger movement, the patient taps fingers and thumb during the activity epoch. The beginning and end of this activity period is cued by a visual or auditory signal and occurs at images 10 and 20, respectively. Language, sensory, visual, auditory and other targeted functions are imaged in a similar manner. A task-induced signal change is illustrated in Fig. 1 for a sensory task involving tactile stimulation (touching) the left hand. The abscissa represents a 30

image acquisition run during a 90 sec period. The initial 10 pre-stimulation (baseline) images are followed by 10 activation images (left hand stimulation) and 10 post-stimulation images. Each 90 sec imaging series (illustrated by the intensity levels for one voxel) is actually repeated twice although only one series is illustrated. In this example the left hand stimulation results in right hemisphere activity and presumably represents the post central sulcus sensory strip.

BOLD-fMRI (Blood-Oxygen-Level-Dependent fMRI)

BOLD-fMRI is currently the most common fMRI technique (Table 2).

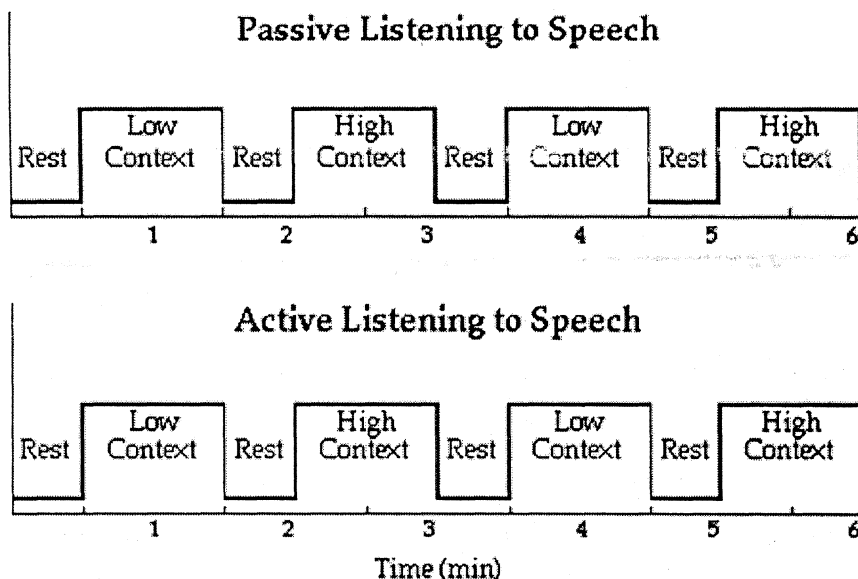
Here, the MRI scanner is tuned to resonate and image hydrogen atoms as in conventional MRI; however, T2*-weighted images are performed which take advantage of the fact that deoxygenated hemoglobin is magnetic whereas oxygenated hemoglobin is not⁶⁻⁸. Because of the magnetic properties of the unflipped magnetic deoxyhemoglobin molecule which causes rapid dephasing, T2* signal is retained longer in a region when it has more oxygenated blood compared to when there is less oxygenated blood. Thus, an area with more oxygenated blood will show up more intense on T2*-weighted images compared to when there is less oxygenated blood around.



With this technique, it is assumed that an area is relatively more active when it has more oxygenated blood compared to another point in time^{1,6}. This is based on the principle that when a brain region is being used, arterial oxygenated blood will redistribute and increase to this area. This principle has one caveat: there is a time lag of 3-6 seconds between when a brain region is activated and blood flow increases to it^{1,9,10}. During this time lag of 3-6 seconds, in fact, the activated areas experience a relative decrease in oxygenated blood as oxygen is extracted by the active regional neurons. Afterward, the amount of blood flowing to the area far outweighs the amount of oxygen that is extracted so that oxygenated blood is now higher. Although images can be acquired every 100 msec with echoplanar (a type of rapid acquisition) BOLD fMRI, this predictable but time varied delayed onset of the BOLD response limits the immediate

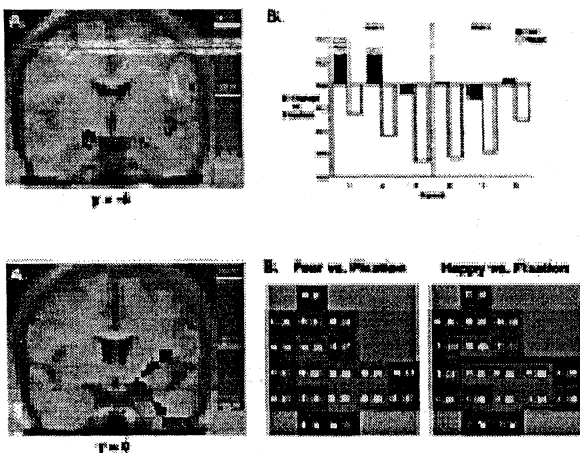
temporal resolution to several seconds instead of the 100 msec potential⁹. In the future, researchers may be able to improve the temporal resolution of fMRI by measuring the initial decrease in oxygenated blood with activation¹¹.

BOLD fMRI is a relative technique in that it must compare images taken during one mental state to another to create a meaningful picture. As images are acquired very rapidly (ie. a set of 15 coronal brain slices every 3 seconds is commonly done in our lab), one can acquire enough images to measure the relative differences between two states to perform a statistical analysis within a single individual. Ideally, these states would differ in only one aspect so that everything is controlled for except the behavior in question¹⁰. Breiter et al. (1996), for example, scanned Obsessive-Compulsive Disorder patients and healthy controls during activated (ie. holding dirty washcloth) versus rest states (ie. holding clean washcloth)¹² (see figure 2).



BOLD-fMRI paradigms generally have several periods of rest alternating with several periods of activation. Images are then compared over the entire activation to the rest periods (see figure 3). Images obtained over the first 3 to 6 seconds of each period are generally discarded due to the delay in hemodynamic response. Alternating paradigms are used because the signal intensity generated by the MRI scanner drifts with time.

With current technology, fMRI-BOLD is best used for studying processes that can be rapidly turned on and off like language, vision, movement, hearing, and memory⁹. The study of emotion is hampered by its slow and variable onset and its inability to be quickly reversed^{13,14}. Some have, however, succeeded in using this technique to study emotional processes^{12,15}. For example, Whalen et al. (1998)¹⁶ used a backwards masking procedure to present 3 alternating conditions to 10 subjects: (1) a baseline condition where subjects would see a “+” sign, (2) a “happy” (H) condition where subjects would see repeated presentations of 33 msec of a happy face followed by 167 msec of a neutral face, and (3) a “fear” (F) condition where subjects would see repeated presentations of 33 msec of a fearful face followed by 167 msec of a neutral face. Here, fearful and happy faces were presented in such a way that 8 of the 10 participants could not identify them. Despite this unconscious processing, subjects had relatively increased amygdala activation with fearful faces and relatively decreased amygdala activation with the happy faces (see Figure 4).



BOLD-fMRI is very sensitive to movement so that tasks are limited to those without head movement, including speaking. BOLD-fMRI is also limited in that artifacts are often present in brain regions that are close to air (i.e. sinuses). Thus, there are some problems in observing important emotional regions at the base of the brain like the orbitofrontal and medial temporal cortices. Another problem is that sometimes observed areas of activation may be located more in large draining veins rather than directly at a capillary bed near the site of neuronal activation⁶.

Currently, there are no indications for BOLD-fMRI in clinical psychiatry, although this technique holds considerable promise for unraveling the neuroanatomic basis of psychiatric disease. It may be of potential help in sorting out diagnostic heterogeneity and treatment planning in the future. Neurologists and neurosurgeons are beginning to use this technique clinically to noninvasively map language, motor, and memory function in patients undergoing neurosurgery²⁻⁶.

Bold is using gradient echoplanar sequences. In EPI, a gradient echo is formed in the broadening direction by applying an initial dephasing gradient, and then a series of 'blips' of the opposite polarity which form the echo. The time from the r.f. excitation pulse to the centre of the echo is defined as the echo time in EPI (see Figure 4.1). This is distinct from the repeated set of echoes formed by the switching of the read gradient.

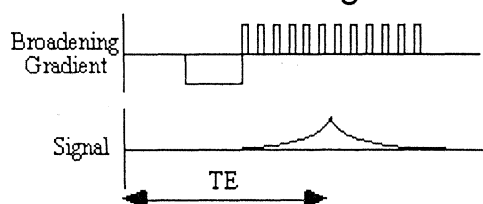
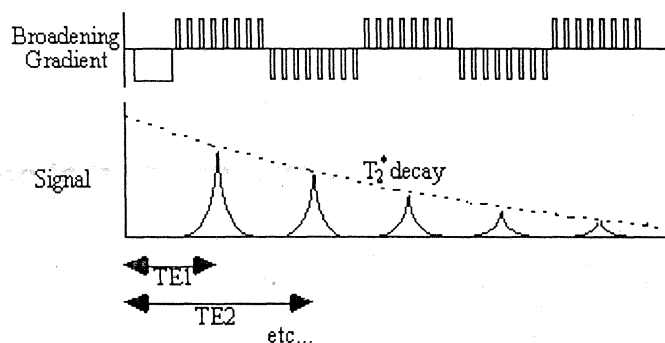


Figure 4.1. Formation of a Gradient Echo in EPI using the Broadening Gradient.

Having formed one gradient echo, it is possible to form other echoes simply by reversing the polarity of the blipped gradient. These echoes all form under the T_2^* decay envelope, and have different echo times (Figure 4.2). This is distinct from the multiple spin echo experiment described in section 2.4.3, in which the echoes form under the T_2 decay envelope.

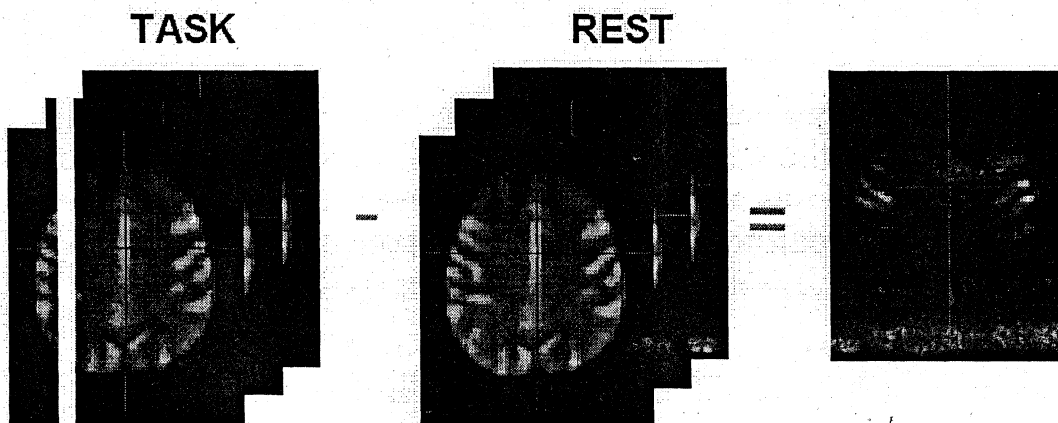


fmri paradigm

A large series of images are acquired rapidly while the subject performs at rest and activation. Several such volumes may be collected while the subject performs different tasks.

By correlating the signal time course in each voxel, it is possible to identify voxels that show signal change.

From acquired images to
“a clinically useful picture”

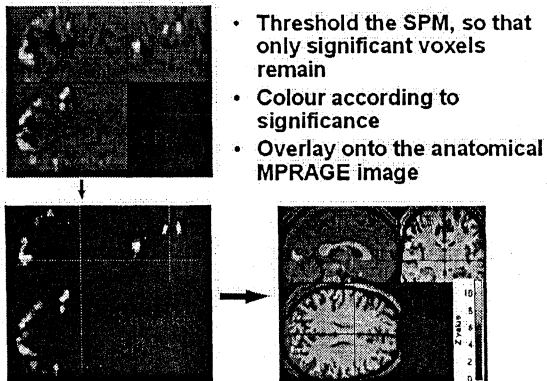


The t-test

The t-test calculates the difference in signal intensity between the two groups, rest and activation. Analyzing BOLD data - the calculation of t-

statistic for each voxel .One can take a threshold value of eg:- $p < .01$, this can pick out all voxels whose t value passes the threshold . These voxels are then displayed in colour over the anatomy images

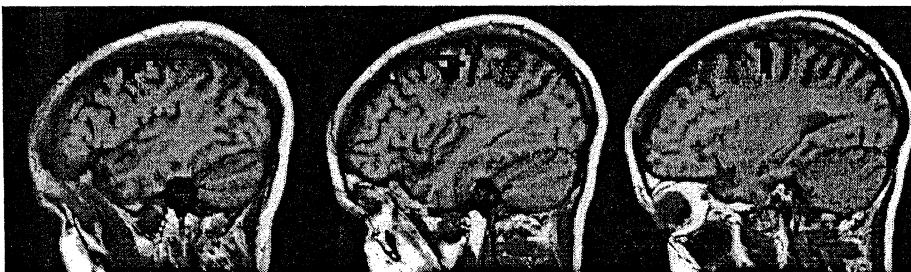
Visualisation

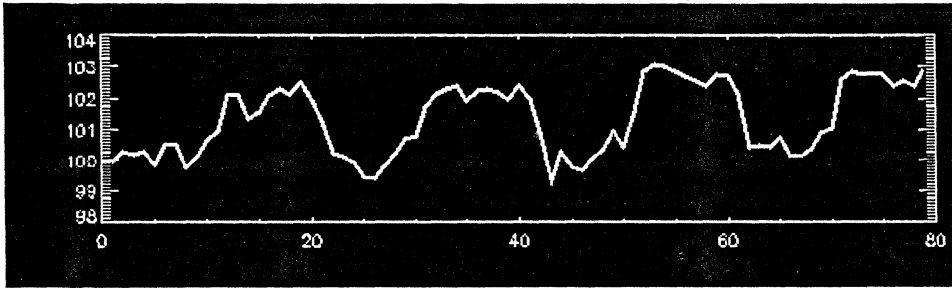


The kolmogorov –smirnov test

Same like t-test For each population (rest and activation) measured values are sorted in an ascending order .From the two distributions (P_{act} and P_{rest}) KS statistic is calculated as the maximum difference between the two cumulative distributions

Time Series and Activation Maps





Challenges in Functional FMRI

Sensitivity (Contrast-to-noise ratio)

BOLD signal change is ~1-2% at 1.5 T; signal-to-noise ratio in single-shot EPI images is ~100.

Physiological pulsations (cardiac and respiratory); Head motion; instrumental instability

Specificity

Location of activation – neurons or veins

Susceptibility artifacts

Suppression of Temporal Fluctuations

Head motion reduction

Head holder modified from a football helmet

Image realignment in data processing

Physiological pulsations

Correction using simultaneously recorded cardiac and respiratory signals

Ultra-fast imaging techniques

Single-shot echo-planar imaging (EPI)

Single-shot Spiral imaging

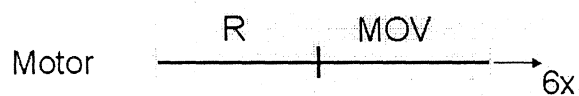
Post-processing

Denoising

fMRI paradigms

- Motor
- Language
- Other

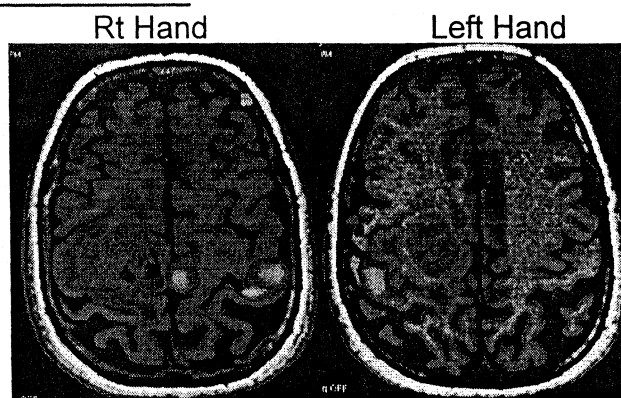
Motor Paradigm- 'BOX- CAR' Design



2 conditions are alternated:

- Rest (R)
- Movement (MOV)

Motor function



ARTEFACT IN BOLD :

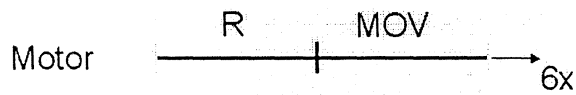
One type of artifact which is specific to EPI is the Nyquist. This arises because odd and even echoes are acquired under opposite read gradients, and the data requires reversal prior to image reconstruction. Inaccurate timing of the sampling relative to the switched gradient, temporal asymmetries in the analogue filter or inhomogeneities in the static field cause a modulation of alternate lines in k-space. This leads to a ghost image shifted by $N/2$ pixels in the phase encode direction. In fMRI, ghosting artifact can cause a number of problems. In the first place, the ghosting can spoil the look of the image, and ghosting of activated regions could lead to apparent 'activation' appearing outside the head. More serious effects occur if the artifact changes with time. Movement of the subjects head causes the interference fringes, of overlapping ghost and image, to change dramatically, with even small displacements having a large effect. Since subject motion is often stimulus correlated, particularly if the experimental paradigm involves movement, the changes of the interference pattern can show up in the statistical analysis as 'activation',

Diffusion-Weighted Imaging:

Diffusion Weighted imaging provides a technique for mapping proton contrast that reflects the microvascular water environment. Diffusion imaging makes use of the variability of "Brownian motion" of water molecules in brain tissue.

Brownian motion refers to the random movement of molecules. Water molecules are in constant motion, and the rate of movement or diffusion depends on the kinetic energy of the molecules and is temperature dependent. In biological tissues, diffusion is not truly random because tissue has structure. Cell membranes, vascular structures, and axon cylinders, for example, limit or restrict the amount

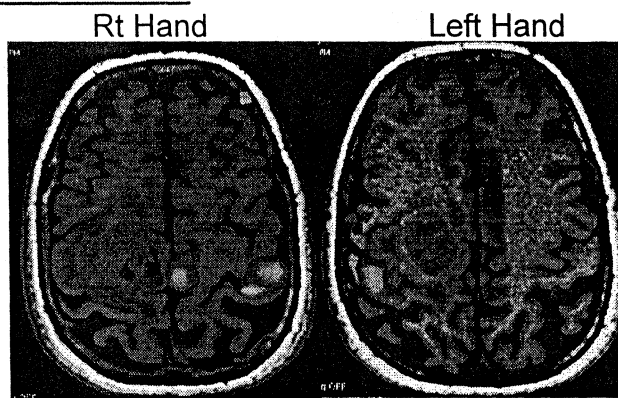
Motor Paradigm- 'BOX- CAR' Design



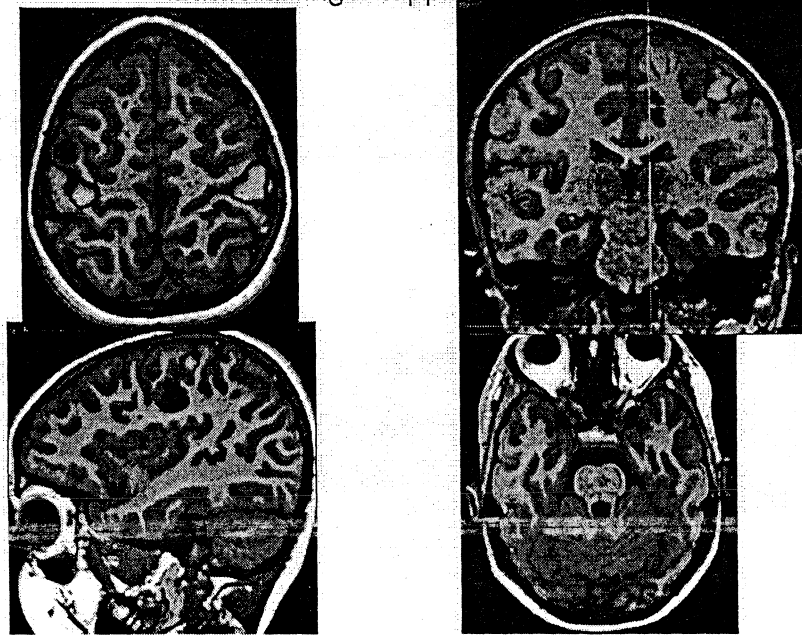
conditions are alternated:

- Rest (R)
- Movement (MOV)

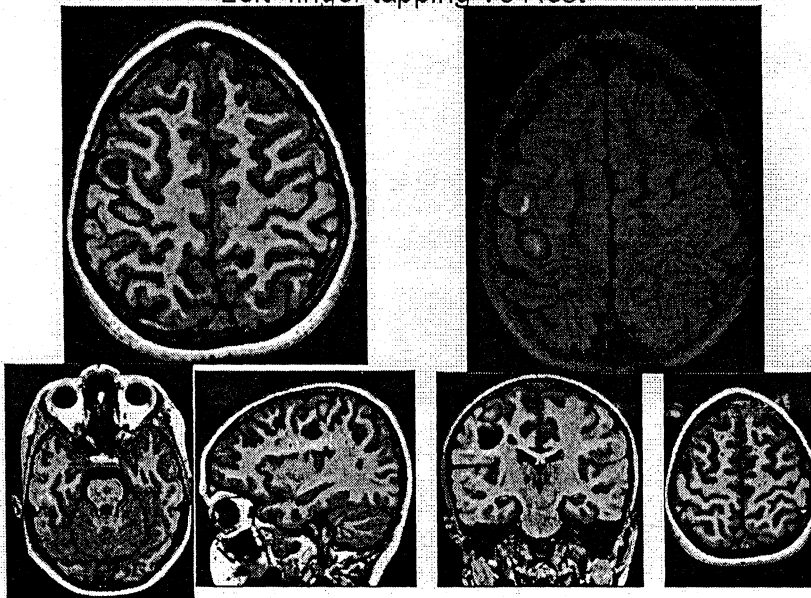
Motor function



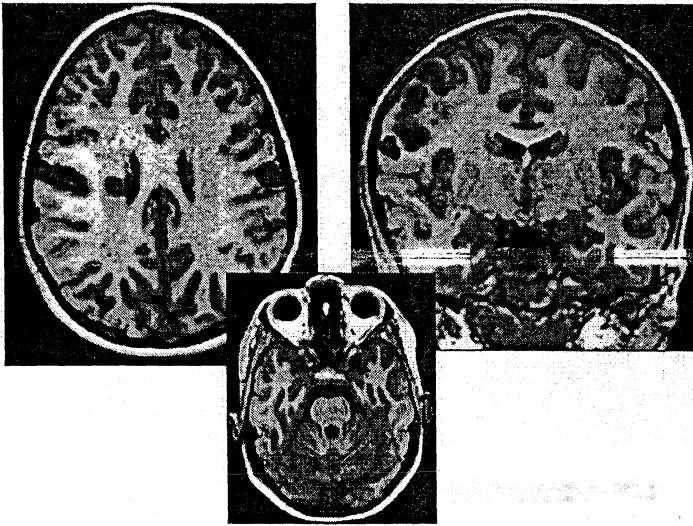
Bilateral finger tapping Vs Rest



Left finger tapping Vs Rest



Lip Pouting vs rest



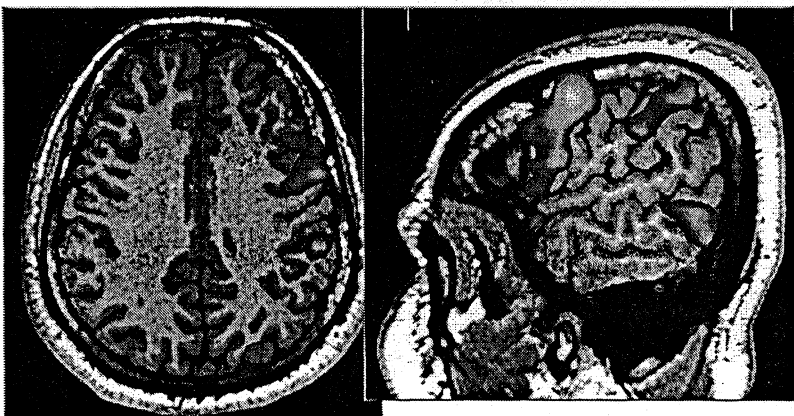
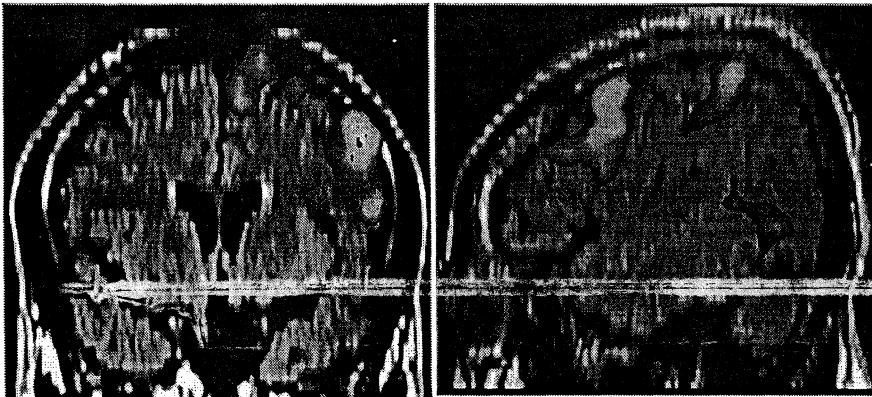
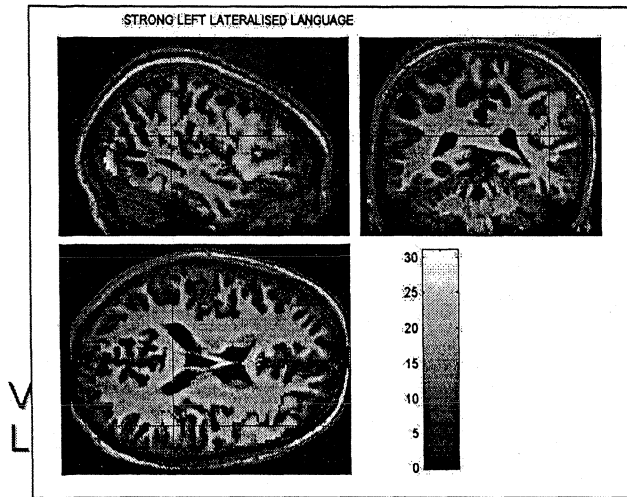
Bilateral leg motor vs Rest



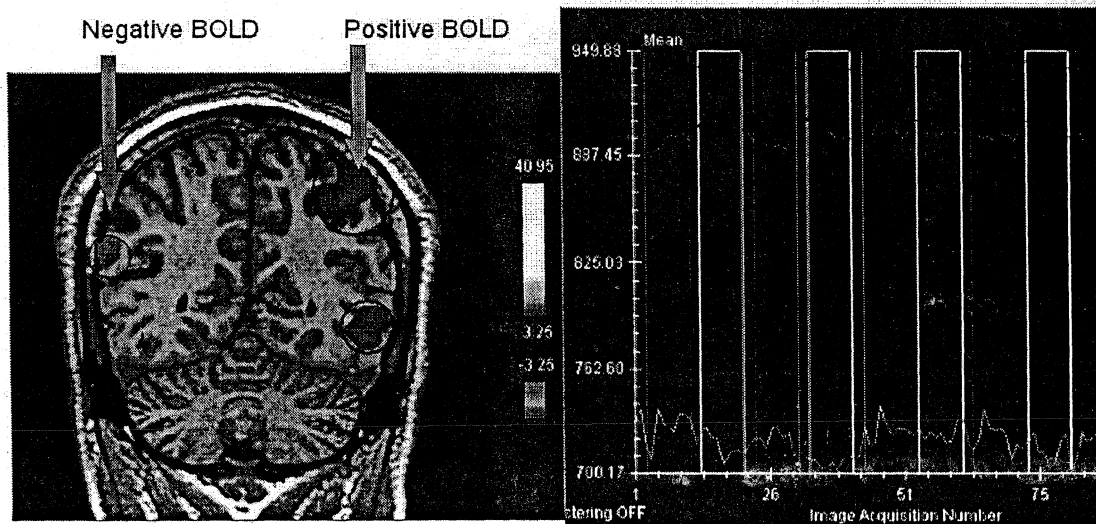
Language Paradigms

Verbal Fluency

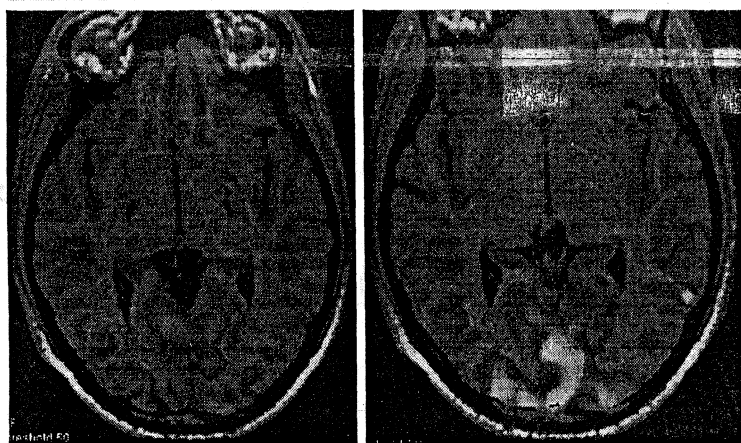
Semantic Decision
Language



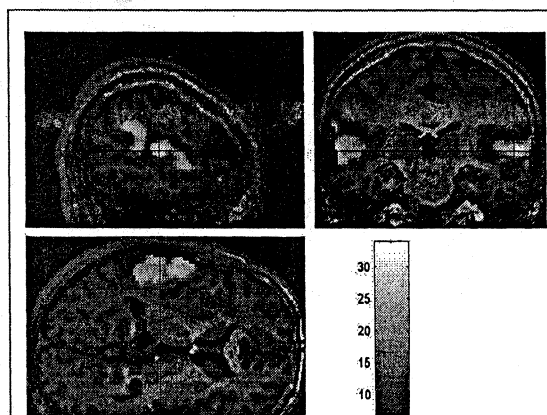
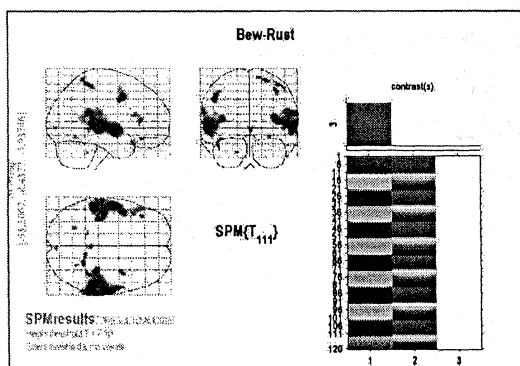
Verbal Fluency_ Inline BOLD
Left lateralized language



Visual



Auditory



ARTEFACT IN BOLD :

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of diffusion interactions

- Turbo STEAM
 - Conventional 1.5T system
 - Gradients - 9 mT/m
 - Single slice - 27 sec

diffusion weighted sensitive to of 1H in (brownian movement). The amount of water diffusion for a given pixel can be calculated and is called the apparent diffusion coefficient (ADC).

Also, chemical of water and macromolecules affect properties. Diffusion-imaging is very the random movement water molecules

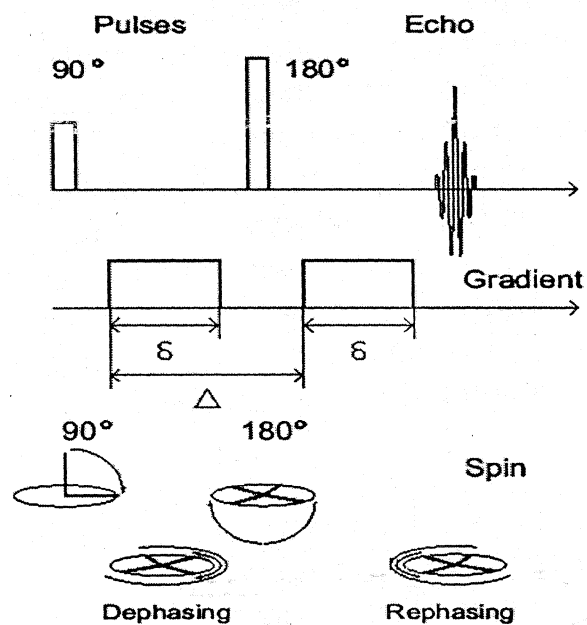
Areas with low ADC values (ie. low diffusion) appear more intense. ADC values are direction sensitive. For instance, if images are taken perpendicular to myelin fiber tracts like the optic chiasm, arcuate fasciculus, or corpus callosum, ADC values will be lower than if the images are taken along the length of these fibers. This is thought to be because there is little diffusion across myelin sheaths.

Thus, ADC direction sensitivity permits detection of myelination and may allow researchers to understand in greater detail myelin development in infants. On the other hand, this direction sensitivity hampers the study of diffusion in other processes as ADC values differ, depending on the imaging plane (axial, coronal, or sagittal). Gradients are applied in to three orthogonal plain of gradient (x,y and Z) and creating a average diffusion weighted image . There are now ways to calculate average ADC values incorporating all planes for each pixel, removing "artifacts" due to the direction of acquisition. Removing the directional diffusion sensitivity has been helpful in studying stroke.

A pair of strong gradient pulses are added to the pulse sequence. The first pulse dephases the spins, and the second pulse rephases the spins if no net movement occurs. The amount of diffusion weighting is determined by the strength of the diffusion gradients, the duration of the gradients, and the time between the gradient pulses

DIFFUSION TENSOR IMAGING:

The basic idea of diffusion tensor imaging is the same as for Phase Contrast Angiographic MRI. It is an anisotropic process with different molecular mobility in x, y and z directions, the diffusion constant "D" has to be replaced by a diffusion tensor



diffusion in white matter is extremely variable. The value of the diffusion co-efficient directly depends on the relative orientation of the fibers and the magnetic field gradients, which is known as "anisotropic diffusion." Water diffusion in gray matter does not exhibit

Although the individual axons and the surrounding myelin sheaths cannot be revealed with the limited spatial resolution of *in vivo* imaging, distinct bands of white matter fibers with parallel orientation may be distinguished from others running in different directions .

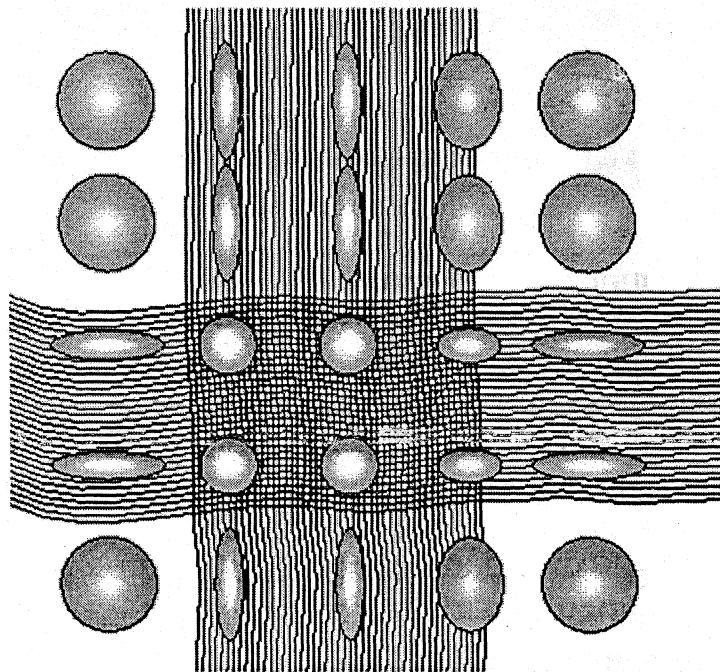
Calculation of the ADC

Pixel-by-pixel calculation of the ADC was performed by fitting the characteristic curve, $\log(S_0/S_b) = ADCb$, onto the DW image data, where S_0 is the signal intensity obtained without diffusion weighting; S_b , the signal intensity obtained with diffusion weighting; and b , the DW factor. Only those pixels that could be clearly differentiated from background noise were considered for calculation.

The trace of the diffusion tensor, Trace (\mathbf{D}), was calculated by using the equation,

$$ADC_{av} = \frac{1}{3} \text{Trace}(\mathbf{D}) = \frac{1}{3} \sum_{j=x,y,z} ADC_j \quad (2)$$

where x , y , and z indicate the directions of diffusion weighting, and ADC_{av} , the average ADC. The calculation of b values was performed by numerical integration of the diffusion-sensitizing gradients



PERFUSION IMAGING

Two methods have been developed for measuring cerebral blood flow. Mainly two methods are using to perform perfusion imaging.

1. intravenous bolus tracking,

Relies on the intravenous (iv) injection of a magnetic compound such as a gadolinium-containing contrast agent and measuring its T2*-weighted signal as it perfuses through the brain over a short time period of time. Areas perfused with the magnetic compound show less signal intensity as the compound creates a magnetic inhomogeneity that decreases the T2* signal. The magnetic compound may be injected once during the control and once during the activation task and relative differences in blood flow between the two states may be determined to develop a perfusion image alternatively, one can measure changes in blood flow over time after a single injection to generate a perfusion map.

Although gadolinium-based contrasts are not radioactive, the number of boluses that can be given to an individual is limited by the potential for kidney toxicity with repeated tracer administration. This technique also only generates a map of relative cerebral blood flow, not absolute flow as in the next technique. Arterial spin labelling is a T1-weighted noninvasive technique where intrinsic hydrogen atoms in arterial water outside of the slice of interest are magnetically tagged ("flipped") as they course through the blood and are then imaged as they enter the slice of interest

Perfusion Weighted Imaging

Several techniques have been developed for determination of perfusion. Both PET and SPECT are relevant for clinical assessment and often use diffusible tracers, which are ideal for measuring perfusion. Both techniques have drawbacks being invasive, using radioactively labeled tracers and having relative low spatial and temporal resolution. For measuring perfusion, MRI has the

advantages of high temporal and spatial resolution using no radioactivity, being almost noninvasive and offering combination studies using other MR techniques (diffusion, spectroscopy, angiography, structural imaging). MR perfusion imaging is divided into categories: (1) susceptibility based techniques either (a) using intravenous bolus injection of a paramagnetic contrast agent or (b) detecting changes in the endogenous paramagnetic substance deoxyhemoglobin (blood oxygen level dependent, BOLD), and (2) arterial spin labeling of protons in blood.

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The most commonly used of these techniques is (1a): contrast agent bolus tracking also known as dynamic susceptibility contrast magnetic resonance imaging (DSC-MRI). DSC-MRI was applied in our studies and will be further described.

2.7.1. Contrast agents

Opposite to radioactive tracers, MR contrast agents are considered safe. Their effect is due to paramagnetic properties causing shortening in T1 and T2 relaxation (further described below).

MR contrast agents containing gadolinium-chelates are approved for use in humans. Gadolinium is a lanthanide metal being paramagnetic property due to seven unpaired electrons. Due to toxicity gadolinium is chelated to different compounds: Gadolinium-DTPA (Dithylen-Triamin- Penta-Acetiacid) is known as gadopentate and sold as Magnevist® (Schering), DTPA-BMA (DTPA-bismethylamide) known as gadodiamide and sold as Omniscan®.

Vascular effects

The effect of MR contrast agents on T1 relaxation is caused by so-called dipole-dipole interactions i.e. on direct interaction of protons in the water molecules with the dipole moment of the unpaired electrons of the paramagnetic contrast agent.

Gadolinium-compounds in the brain are intravascular when the blood brain barrier is intact. T1 shortening produces signal enhancement in the blood volume, that contributes about 5% of the total brain volume. Susceptibility is a measure of the ability of a compound to become magnetized when subject to a magnetic field thus increasing or decreasing the applied external magnetic field. The intravascular paramagnetic contrast agents cause susceptibility changes and increase the local

external magnetic field. A magnetic field gradient between the lumen of the vessel and the surrounding tissue is induced (figure 15), thereby increasing the T_2^* relaxation rate and dephasing. This effect on T_2^* relaxation extends beyond the blood vessel, thus generating relative large and distant signal changes. The method is therefore more suitable for MRI perfusion measurements, than the dipole-dipole effect changing T_1 relaxation in the vessel only. The signal change caused by a paramagnetic compound also depends on the distance that the water molecules diffuse compared to the field inhomogeneity created by the paramagnetic agent, i.e., how many different magnetic fields the spins experience during the duration of the pulse sequence (figure 15). Around a larger vessel, water molecules will only experience the same "static" magnetic field because the distance traveled by diffusion is small in relation to the characteristic distance of field inhomogeneity. Dephasing due to static inhomogeneity leading to signal loss will occur in a gradient-echo experiment but not in a spin-echo experiment. Around smaller vessels, the water molecules will experience varying magnetic field gradients due to diffusion (around the vessel and around the neighboring vessels), this will lead to dephasing and signal loss in a gradient-echo as well as in a spin-echo experiment. Using gradient-echo sequences, the T_2^* effects from both large and small vessels will therefore be detected. Using spin-echo sequence the T_2^* effects from the large vessels will be suppressed. However, gradient-echo is often used in studies due to the better SNR, (greater signal loss than with spin-echo sequences), when using the same amount of MR contrast agent. The susceptibility effect will decrease when the BBB is damaged because local field inhomogeneity decreases. The dipole-dipole effects shortening T_1 will cause a greater signal increase in the tissue. The MRI sequence can be optimized so T_1 effects are decreased and post-processing of the signal time curve has also been proposed (108), (109). Use of paramagnetic contrast compounds causing small T_1 effects, e.g. dysprosium, can minimize effects from leakage of the BBB, although not available for human studies (110). Perfusion is confined to the capillary bed and using a spin echo sequence will select the relevant compartment for a perfusion measurement, (111). Spin echo images have better contrast-to-noise (CNR) than gradient echo images and are without severe susceptibility artifacts in the inferior frontal and temporal lobes. On the other hand, information on hemodynamics in the large

vessels is abolished. Also, more contrast agent is needed in bolus tracking using a spin echo sequence because $\Delta R2^*$ is smaller than $\Delta R2$.

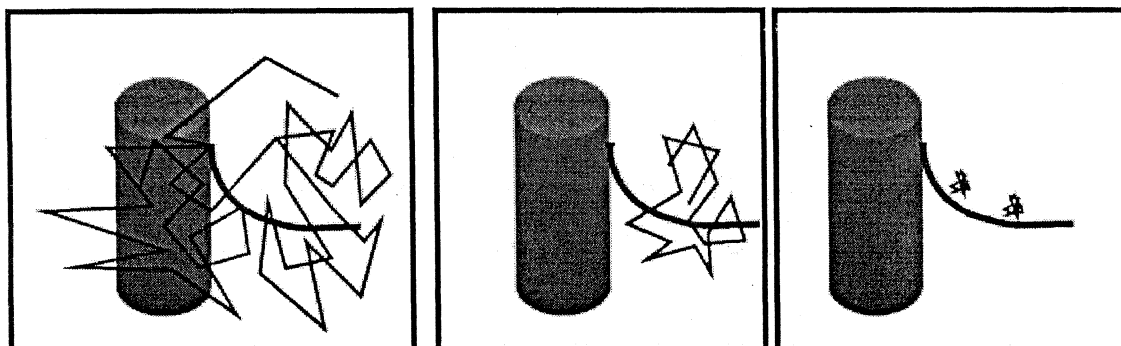
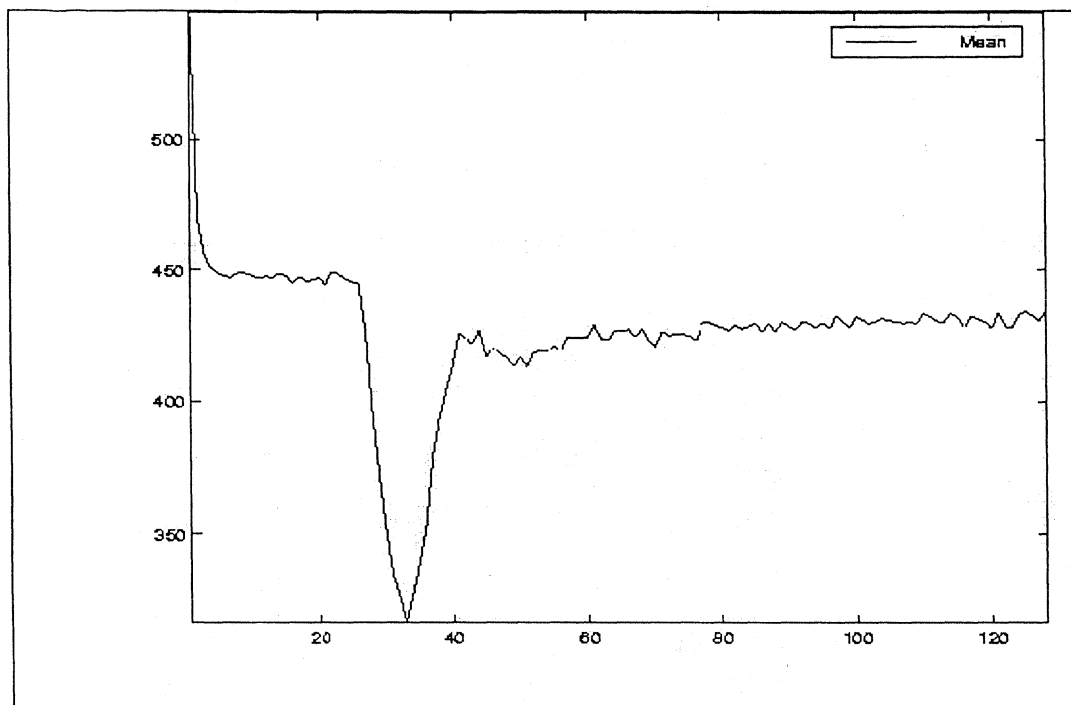


Figure 15. Vessel size and susceptibility effects

From left to right: small vessel, intermediate vessel, large vessel

2.7.3. Contrast agent bolus tracking

When tracking a bolus injection of paramagnetic contrast agent by fast imaging, the MRI signal will drop transiently, due to the susceptibility effect when the gadolinium-compound passes through the detecting plane (figure 16). The bolus of 0.1-0.2 mmol/kg is injected via the antecubital vein (3-5 ml/sec), using a power injector. With the availability of EPI, it has become possible in the last five years to obtain the passage with high temporal (1-2 seconds) and high spatial resolution (~ 1 mm) by multislice imaging (6-11 slices). 10-40 frames of baseline images should be acquired before the bolus is injected increasing SNR ratio when calculating $\Delta R2^*$ (see formula [2.20]) (111). The TE should be long (60-75 ms) increasing the susceptibility sensitivity (111).



60

Figure 16. Contrast agent bolus tracking *signal-time curve*
Tracer kinetics

Quantification of perfusion is done using the central volume theorem (112) based on certain assumptions describing a systems response to an injection of a tracer: Flow is constant during observation and the tracer is inert with no metabolism or retention.

The central volume theorem states that: CBF is the perfusion, CBV is the blood volume and MTT is the mean transit time. The central volume theorem is a general equation for all kind of tracers (diffusible, Intravascular or intermediate). The model used when calculating perfusion in DSC-MRI is based on tracer kinetics for nondiffusible intravascular tracers (113). When a bolus of contrast agent is injected, the concentration $C_{voi}(t)$ of the tracer in a volume (VOI) can be described in terms of

$$CBF = CBV/MTT \quad [2.14]$$

CBF is the perfusion, CBV is the blood volume and MTT is the mean transit time.

The central volume theorem is a general equation for all kind of tracers (diffusible, Intravascular or intermediate). The model used when calculating perfusion in DSC-MRI is based on tracer kinetics for nondiffusible intravascular tracers (113). When a bolus of contrast agent is injected, the concentration $C_{voi}(t)$ of the tracer in a volume (VOI) can be described in terms of $h(t)$, the *frequency distribution function*, describing the distribution of transit times through the VOI following an ideal instantaneous bolus.

• $R(t)$, the *residue impulse response function* i.e., the fraction of the bolus still present in the VOI at time t following a ideal instantaneous bolus. $R(t)$ and $h(t)$ are related

by

$$R(t) = \left[1 - \int_0^t h(\tau) d\tau \right] = 1 - H(t) \quad [2.15]$$

$$H(t) = \int_0^t h(\tau) d\tau \quad [2.16]$$

[2.16], represents the accumulated fraction of the bolus that has left the VOI. Formula [2.15]

state that after a instantaneous bolus, $R(t)$ is the fraction of the bolus in the VOI at time - t .

• *AIF*, the *arterial input function*, $Ca(t)$, the concentration of contrast agent in the supplying vessel to the VOI.

From these definitions $C_{voi}(t)$ is calculated as:

$$C_{voi}(t) = \rho / k_s \cdot CBF_{voi} \cdot (Ca(t) \otimes R(t)) \quad [2.17]$$

$$= \rho / k_s \cdot CBF_{voi} \cdot \int_0^t Ca(\tau) \cdot R(t-\tau) d\tau \quad [2.18]$$

where

- CBF_{voi} is the perfusion in VOI,
- Δ is the density of the tissue,
- kh is correcting for differences in hemotocrit in capillaries and large vessels.

Formula [2.17] states that C_{voi} can be expressed as a convolution of the residue impulse response function, $R(t)$, and the arterial input function, $Ca(t)$. To calculate CBF_{voi} , the arterial input function $Ca(t)$ and the $C_{voi}(t)$ therefore must be deconvolved determining the residue response function $R(t)$ and CBF_{voi} . Deconvolution is theoretically demanding with inherent uncertainties in the estimated $R(t)$, because the signal time curve in the artery and the tissue is almost identical when using intravascular tracers. However, attempts of calculating the rCBF have shown good agreement with values from PET measurements (114), (115). An accurate estimate of the arterial input function is necessary but can be difficult to obtain due to partial volume effects and bad SNR in EPI. Also it is assumed in the model that the bolus curve in the VOI is not dispersed or delayed compared to the AIF , which is certainly not valid in ischaemic regions (116). The CBV is expressed as:

$$CBV = \frac{k_h}{\rho} \cdot \frac{\int C_{voi}(t) dt}{\int Ca(t) dt} \quad [2.19]$$

Relative CBV can be estimated without knowledge of the AIF , assuming the same AIF to all parts of the tissue. This approach has therefore often been applied in human studies. For intravascular tracers used in DSC-MRI, the time concentration curve obtained in the VOI, also depend on the vascular architecture. The first moment of the tissue concentration curve differs from the true MTT (117). Calculating CBF by the central volume theorem [2.14], with MTT calculated as the first moment of the tissue curve, is therefore wrong

Concentration dependency

The concentration in the VOI C_{voi} used in the calculation of the hemodynamics is related to the change in $T2^*$ relaxation:

$$\Delta\left(\frac{1}{T2^*}\right) = \Delta R2^* = k \cdot C_{voi} \quad [2.20]$$

k is the relaxivity constant. Assuming the linear relationship [2.20] between the concentration of the gadolinium-compound, C_{voi} , and the change in $R2^*$. $R2^*$ is determined from the baseline signal S_0 , and $S_{VOI}(t)$, the signal in the VOI at time t (118), (107), (119). TE is the chosen echo time:

$$C_{voi}(t) = k \cdot \Delta R2^* = -\frac{k}{TE} \cdot \ln\left(\frac{S(t)}{S_0}\right) \quad [2.21]$$

[2.21], is valid for both the tissue and arterial input function, (however k may differ).

Perfusion parameter maps

With the technical and practical difficulty of absolute quantification of perfusion using DSCMRI, various parameters derived from the $R2^*$ time curve can be calculated. The time to $R2^*$ peak, (TTP), the bolus arrival time (BAT), the peak height, the area under the $R2^*$ curve (AUC) (which is related the CBV) and the first moment of the $R2^*$ time curve used as an approximation for the MTT. The interpretation, except for $AUC = CBV$ is not straightforward because there is no simple relation between these parameters and the rCBF. Also the parameters should be normalized, because they vary depending on the vascular architecture, cardiac output, injection

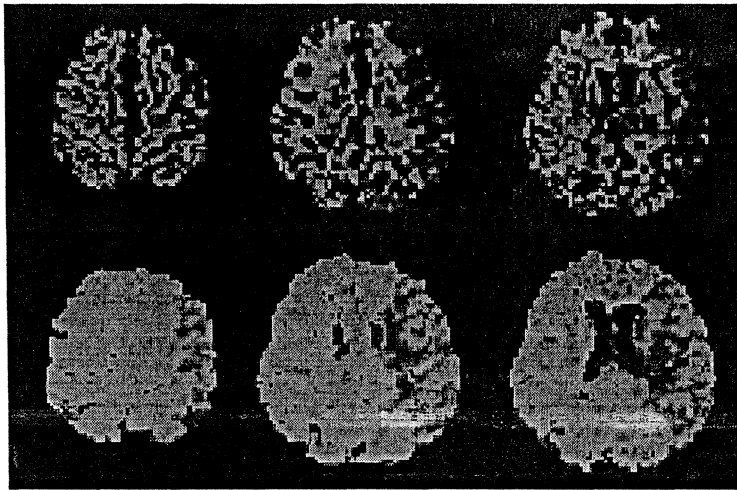
rate and dose. In ischemia, the normal contralateral side will often be used for normalization of the different parametric maps.

The effects of re-circulation can be eliminated either by considering only the first part of the signal-time curve or by fitting the curve to an assumed model function, most often a gammavariate function (120). The primary aim of fitting a gamma variate function to the data is to estimate the area under the first pass curve and to improve the SNR of the TTP estimate.

Due to the technical difficulties of deconvolution we used TTP and AUC maps in the study of acute stroke. However, TTP maps have some inherent differences compared to MTT and CBF maps in ischemic volumes. For instance, when considering an ischemic area caused by a stenosis or an occlusion, the CBV and CBF are decreased in the ischemic area itself, and the "true" calculated MTT will be unchanged. However, TTP in the case of stenosis will be prolonged in the whole area distal to the stenosis. If collateral blood supply is involved, the TTP will be prolonged in the collateral vessels, because blood is by-passed from the occluded vessel through the collaterals. In both cases TTP will overestimate the ischemic area on TTP maps. Generally, it can be shown that TTP does not depend on the perfusion but only on the actual input function and configuration of the residue impulse response function. However, it is not known from empirical data, if these parameters in ischemic tissue actually have configurations giving changes in TTP occurring parallel to changes in CBF.

CBV

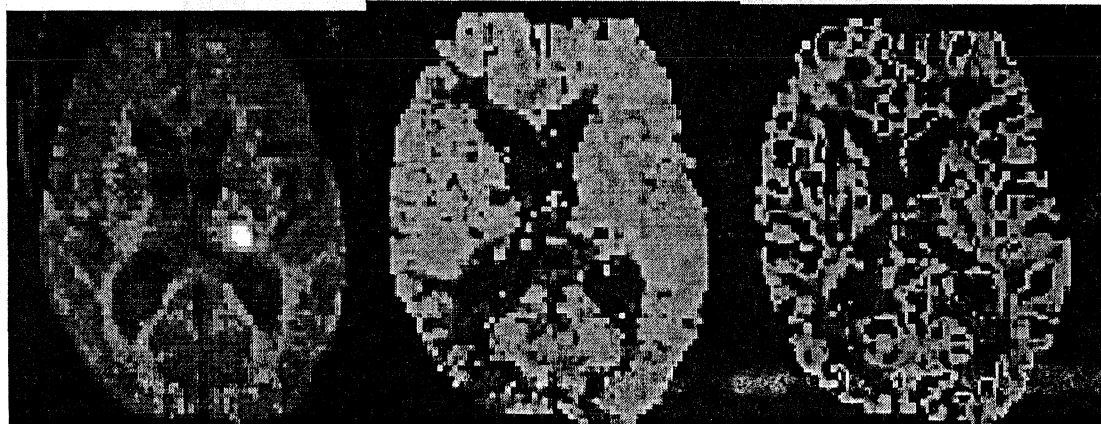
TTP

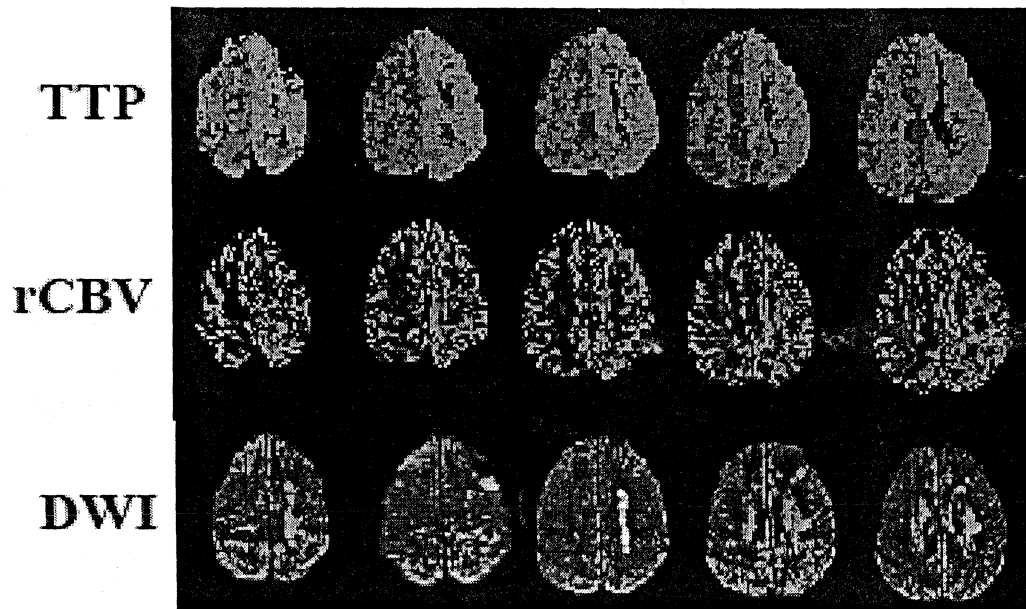


DWI

TTP

rCBV





2. Arterial spin-labelling.

The non-invasive nature of **arterial spin labelling (ASL)** has opened a unique window into human brain function and **perfusion** physiology. High spatial and temporal resolution makes the technique very appealing not only for the diagnosis of vascular diseases, but also in basic neuroscience where the aim is to develop a more comprehensive picture of the physiological events accompanying neuronal activation. The complete non-invasiveness of ASL makes it very suitable for **perfusion studies** of healthy volunteers and in patient groups requiring repetitive follow-ups. This is especially important in patients with particular conditions, such as kidney failure, or in paediatric populations where the use of radioactive tracers or exogenous contrast agents may be restricted.

Quantitative perfusion

In order to measure tissue perfusion, one needs to follow the course of blood flow through the organ, and for this there exist two methods, one based on freely diffusible and the other on intravascular tracers. As their names suggest, freely diffusible tracers can leave the

intravascular space without restriction and be distributed throughout the entire tissue volume, whereas intravascular tracers remain in the vasculature, which constitutes only a fraction of the full volume.

Basic arterial spin labeling

The overall goal of all existing ASL techniques is to produce a flow-sensitized image or "labelled" image and a "control" image in which the static tissue signals are identical, but where the magnetization of the inflowing blood differs. By adding a delay between **labelling** and image acquisition, called inversion delay (TI), the labelled blood **spins** are allowed to reach the capillaries where they exchange with tissue water and thereby give rise to the **perfusion** signal. The signal difference, which is only 0.5–1.5% of the full signal, depends on many parameters such as the flow, T_1 of blood and tissue, as well as the time it takes blood to travel from the **labelling** to the imaging region. Multiple repetitions are needed for ensuring sufficient signal-to-noise, and a model of the **perfusion** signal is usually used in order to quantify the **perfusion**.

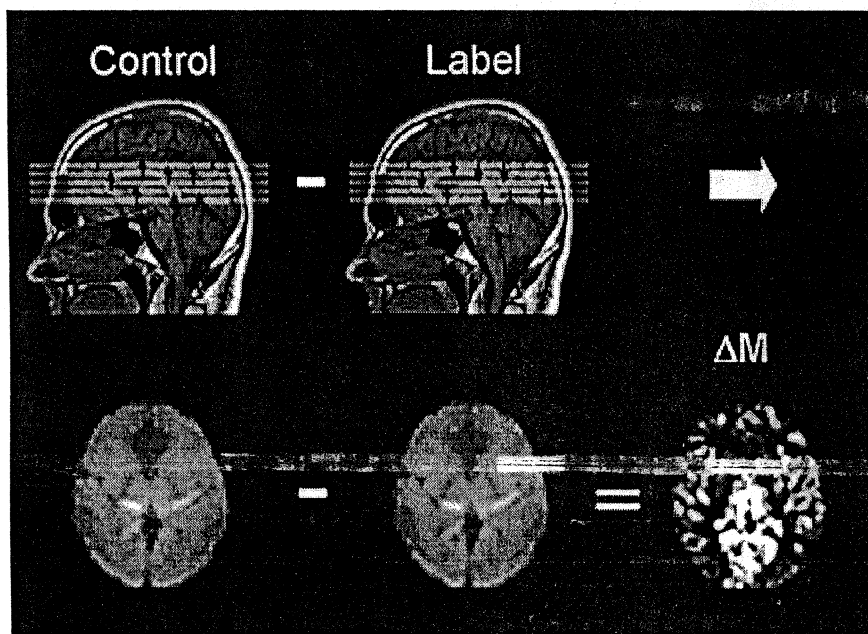


Figure 1 Schematic description of a **perfusion** weighted image (ΔM) obtained by subtraction of the labelled images from the control images.

There exist two main classes of ASL techniques: continuous ASL (CASL) and pulsed ASL (PASL). In CASL, the supplying blood is continuously labelled below the imaging slab, until the tissue magnetization reaches a steady state (Figure 2a+). The PASL approach labels a thick slab of arterial blood at a single instance in time, and the imaging is performed after a time long enough for that spatially labelled blood to reach the tissue and exchange at the region of interest (Figure 2b+). Both methods need a control experiment in order to visualize and quantify the perfusion.

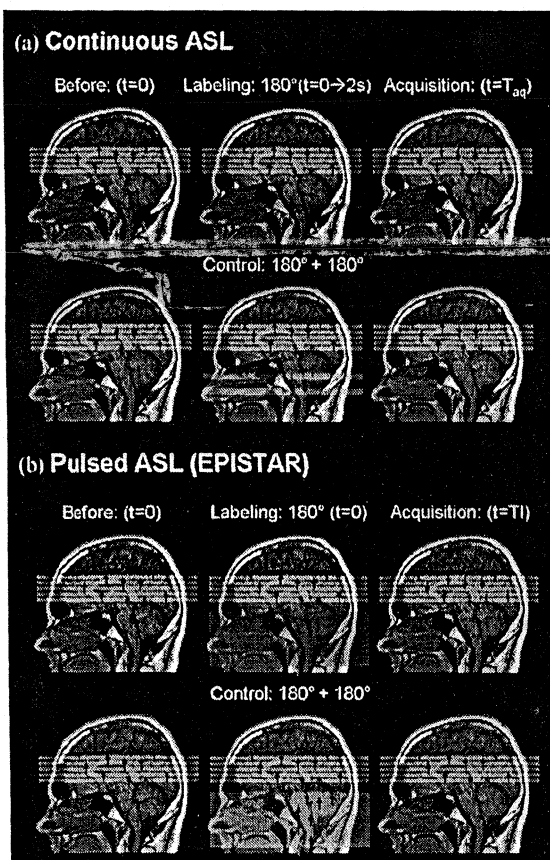


Figure 2. (a) Continuous arterial spin labelling (ASL) multislice experiment, using double adiabatic inversion for the control experiment, where labels get inverted during the passage of the first plane and returned to equilibrium during the subsequent passage of the second plane. (b) The EPISTAR pulsed ASL sequence, which labels everything at once and uses two $180^\circ + 180^\circ = 0^\circ$ pulses for the control images

Continuous arterial spin labelling

The original ASL method proposed by Williams et al in 1992 used a

continuous flow-driven adiabatic inversion scheme, a method that was previously used for angiography. This type of adiabatic inversion of the **arterial** magnetization is realized using a 2–4 s continuous radiofrequency (RF) pulse while applying a magnetic field gradient in the flow direction. The moving **arterial spins** will therefore experience a slow variation of the resonance frequency, which will result in their inversion, while static (tissue) **spins** will just be saturated. Typically, the inversion "slice" will be selected just proximal to the circle of Willis, near the medullospinal junction or at the level of the common carotid, and the **spins** in blood that flows through this plane will be inverted. The inversion efficiency α , which is important for further quantification, depends on factors like the mean velocity of the blood, angulations of the vessels to the plane and the selected RF amplitude and gradient strength. Typical **labelling** efficiency is in the range of 80–95%.

Among the confounding factors of these long lasting inversion pulses are magnetization transfer (MT) effects. When using a single coil for **labelling** and imaging, the off-resonance **labelling** pulse (with respect to imaging slice) will act as a powerful MT pulse in a way similar to an MT-weighted technique. The resulting saturation effect of the macromolecular pool will result in a reduced signal of the free water pool from the tissue of interest. This is a very important issue, as the **perfusion** weighted images are calculated by subtracting a labelled from a control acquisition, and if this MT effect is present only during the **labelling** scheme, it will lead to overestimated **perfusion**. In the first implementation, these MT-effects were compensated for by applying a distal **labelling** during the control experiment. This produces identical saturation effects but, due to the applied gradients during **labelling**, this is unfortunately valid only for a single slice. For multislice acquisition, Alsop et al proposed the use of two closely spaced inversion planes, also called double adiabatic inversion (DAI). In the control experiment, the magnetization gets inverted while traversing the first plane and returns theoretically to its original state during the passage through the second plane (the CASL experiment is shown in Figure 2a*). Double inversion is achieved by applying a sinusoidal modulation of the RF waveform. Global control of the MT-effects is obtained by matching the RF power and the location of the planes. Another method was also proposed, called simultaneously

proximal and distal RF irradiation (SPDI) for multislice acquisition [23]. In this later scheme, the RF power on the control scan is distributed evenly on both sides of the acquisition volume. However, a big limitation of these approaches is the doubled RF deposition, resulting in higher specific absorption rates (SAR).

Pulsed arterial spin labelling

In 1994, Edelman et al [31] proposed the first pulsed ASL scheme. Contrary to CASL, the **labelling** is performed once in a 10–15 cm slab proximal to the image slices. For the PASL sequences, MT-effects have to be considered as well, although these are much smaller compared with CASL. In this first version of the "Echo-Planar MR Imaging and Signal Targeting with Alternating Radio frequency" (EPSTAR) sequence, inversion was performed distal to the image slice during the control experiment to induce identical MT effects in both cases. Again, this truly compensates for a single slice only and, therefore, the sequence was modified for multislice acquisition using a single 180° adiabatic pulse for the label experiment and two $180^\circ + 180^\circ$ pulses of half the power for the control experiment at the same proximal location

ASL perfusion quantification

Having acquired the data using either technique, the subtracted control-label images will be **perfusion** weighted (Figure 1*). The relationship between the ΔM signal and the actual CBF depends mainly on proton density and T_1 relaxation rates of tissue and inflowing blood, and their respective differences. In addition, the label transit time from the inversion slab to the observed region in the images is also an important factor. Traditionally, quantitative CBF estimation is carried out using the tracer clearance theory originally proposed by Kety and Schmidt

The modified Bloch equation, including the flow dependent exchange term, becomes:

$$\frac{dM_t(t)}{dt} = \frac{M_{t,0} - M_t(t)}{T_{1t}} + \text{CBF} \left(M_a(t) - \frac{M_t(t)}{\lambda} \right) \quad (3)$$

where M_t , $M_{t,0}$ and M_a are the tissue-, equilibrium- and **arterial**-magnetizations, respectively, λ is the blood–brain partition coefficient, and T_{1t} is the longitudinal relaxation rate of the tissue. In the original quantification model, further assumptions about uniform plug flow and equal T_1 relaxation of both tissue and **arterial** blood were made

Buxton et al proposed a general kinetic model where all the above mentioned parameters can be taken into account. Here, the magnetization difference between labelled and control measurements is described using the convolution integral in a way similar to Equation (2):

$$\Delta M = 2 \cdot M_{a,0} \cdot \text{CBF} \cdot \int_0^t c(\tau) \cdot r(t-\tau) \cdot m(t-\tau) d\tau \quad (4)$$

where $M_{0,a}$ is the equilibrium magnetization in a blood filled **arterial** voxel, $c(t)$ is the delivery function or fractional **arterial** input function (AIF). The residue function $r(t-\tau)$ describes the washout of labelled **spins** from a voxel, and $m(t-\tau)$ includes the longitudinal magnetization relaxation effects. The possibility to choose a particular **arterial** input function and to consider a certain exchange mechanism (single or multicompartment) allows greater flexibility for data analysis of both CASL and PASL experiments.

Comparison with other imaging modalities

At this point, it should be kept in mind that related issues such as the

ones highlighted in the previous paragraphs apply to other **perfusion** modalities like DSC-MRI, CT-**perfusion** and positron emission tomography (PET) as well. Dynamic susceptibility contrast using tracers such as gadolinium-DTPA for instance, suffers from the fact that the relationship between measured signal and contrast is non-linear and depends on parameters such as the field strength, shim of the magnet and the constitution of the vessels. In addition, correct scaling of the AIF is influenced by partial volume effects in the voxels from where it is measured. Altogether, this makes quantification troublesome, and in the general clinical practice only a relative **perfusion** measure is possible. Similar problems exist in CT-**perfusion**, mainly related to the extraction of the global AIF. PET can be said to be a more "pure" method for measuring **perfusion** due to the use of free diffusible tracers and experimental durations that ensure a steady state. However, a relatively low resolution of typically 6–10 mm introduces partial volume effects, *i.e.* a mixture of grey-matter, white-matter and CSF will be present in almost all voxels, making direct comparisons to the abovementioned methods questionable.

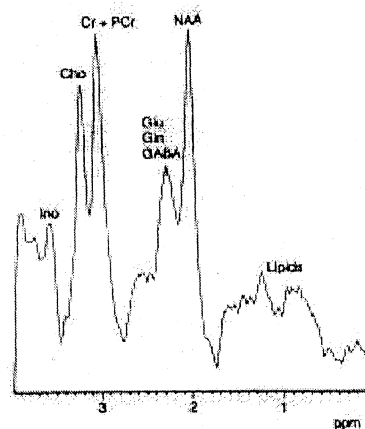
MR SPECTROSCOPY

HISTORY

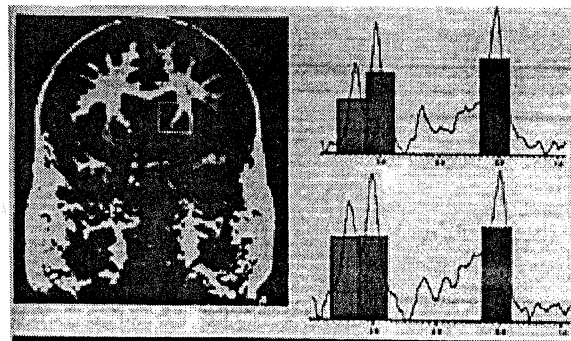
Purcell et al³ and Bloch et al⁴ first elucidated the principles of nuclear magnetic resonance in 1946. The technique of spectroscopy is widely applied in chemistry for the analysis of compounds in solution, and is a powerful tool for determining the structure of biological macromolecules. Similarly, MR spectroscopy can be used to identify important molecules in living tissue. Protons often are used for MR spectroscopy because of their high natural abundance and high nuclear magnetic sensitivity.⁵ Despite the huge number of biomolecules in tissue, relatively few are identifiable *in vivo* because only freely mobile compounds that are present in substantial concentrations give enough signals to be detected. The concentrations of metabolites of interest are in the millimolar range; water protons are a thousand times as common.⁶ For this reason,

water resonance has to be suppressed so that the other molecules can be detected. The diagnostically resolvable hydrogen MR spectra may be obtained using clinical instruments (1.5 T or greater) and routinely used surface

MRI spectroscopy (MRS) offers the capability of using MRI to non-invasively study tissue biochemistry. The hydrogen atom in water is the main one that is flipped (resonated). In MRS, either ^1H atoms in other molecules or other atoms such as ^{31}P , ^{23}Na , K , ^{19}F , or Li are flipped. Within a given brain region called a voxel, information on these molecules is usually presented as a spectrograph with precession frequency on the x-axis revealing the identity of a compound and intensity on the y-axis, which helps quantify the amount of a substance.



The quantity of a substance is related to the area under its spectrographic peak; the larger the area, the more of a substance that is present.



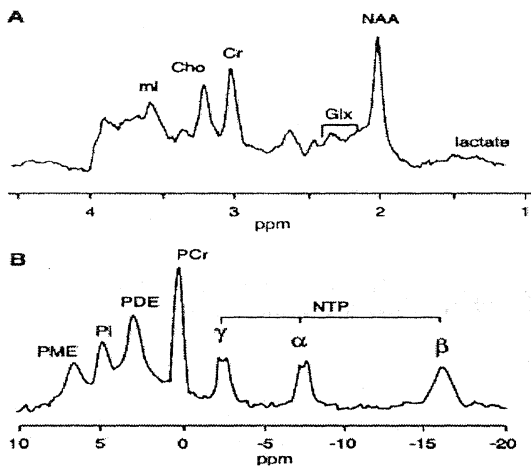
The reason why several molecules can be identified and quantified within a single scan is that the resonant magnetic pulse has a bandwidth over a narrow precession frequency range so that it can flip several molecules at once. The signal intensity at each of these precession frequencies can then be identified using a complicated mathematical procedure called a Fourier transform. For a given precession frequency (or spectrographic peak of a given molecule), information can also be presented spatially as metabolic maps which are created with similar principles to the ^1H atom in water spatial map in conventional MRI³³. The spatial resolution of these maps is generally less than that of conventional MRI as the substance concentration is much less than that of water. Consequently, the minimum area needed to obtain a visible signal is greater. The two most widely used MRS techniques involve either viewing ^1H atoms in molecules other than water or ^{31}P -containing molecules³¹. In ^1H MRS, the water signal must first be suppressed as it is much greater than the signal from other ^1H -containing compounds and has overlapping spectroscopic peaks with these compounds¹. Compounds that can be resolved with ^1H -MRS include

Proton MR Image

MR images contain anatomical information based on the distribution of protons and the relative proton relaxation rates in various tissues
MR images are based on proton signals from water and fat

MR Spectrum

MR spectroscopy determines the presence of certain chemical compounds. Stress, functional disorders, or diseases can cause the metabolite concentration to vary. Metabolite concentrations are low, generating ~10,000 times less signal intensity than the water signal



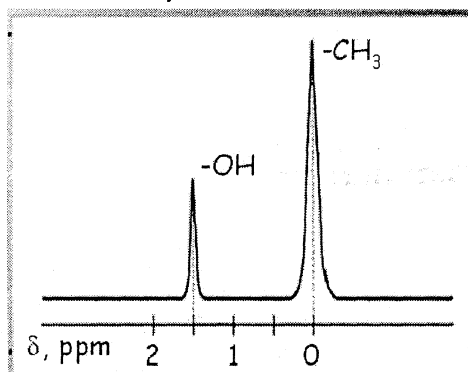
Chemical Shift

The electron cloud around each nuclei shields the external magnetic field. Because of differences in electron shielding, identical nuclei resonate at different frequencies. The resonance frequency in the presence of shielding s is expressed as:

$$\omega = (1 - s)gB_0$$

Where g is the gyromagnetic ratio and B_0 is the external magnetic field strength

¹H MR spectra



The frequency shift increases with field strength. For example, shift difference between water and fat

$(\omega_{\text{water}} - \omega_{\text{fat}})$ at 1.5 T is 255 Hz at 3.0 T is 510 Hz
 $(\omega_{\text{water}} - \omega_{\text{fat}}) 10^6 / \gamma B_0$, in ppm units
 $\Delta\omega_{\text{water-fat}}$ is 3.5 ppm independent of field strength

By convention. Signals of weakly shielded nuclei with higher frequency are on the left. Signals of more heavily shielded nuclei with lower frequency are on the right. Chemical shift of water is set to 4.7 ppm at body temperature.

Spatial Localization

Surface Coil Localization

Simple surface coil acquisition
 Depth Resolved Surface Coil Spectroscopy, DRESS

Single Volume Localization

Image Selected *In Vivo* Spectroscopy, ISIS
 Point Resolved Spectroscopy, PRESS
 Stimulated Echo Acquisition Mode, STEAM

Multiple Volume Acquisition

Chemical Shift Imaging, CSI

Surface Coil Acquisition

A simple loop of wire and associated circuit tuned to the desired frequency are placed directly over the tissue of interest to obtain spectra.



A surface coil

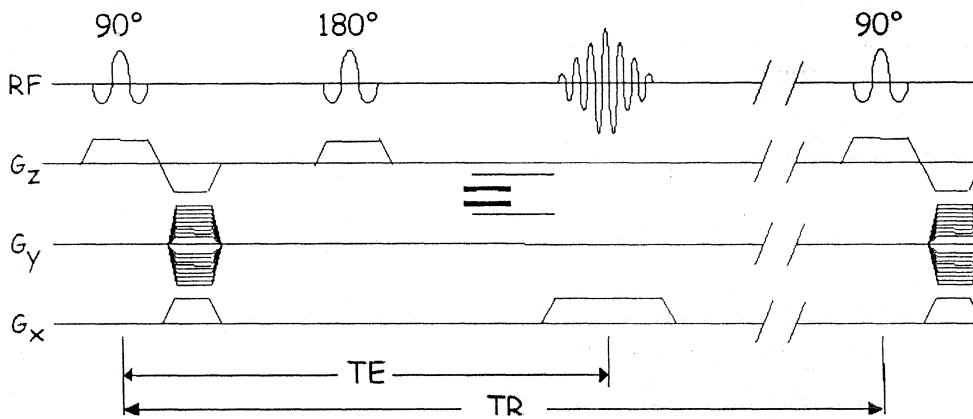
Advantages

Easy to build and does not require specialized pulse sequence
Superb SNR and filling factor

Disadvantages

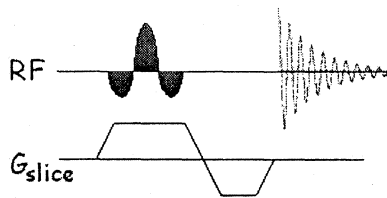
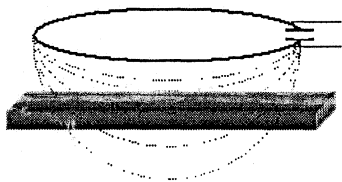
Must be close to region of interest
Changing ROI is difficult
Inhomogeneous RF field

Spin Echo Imaging Sequence



Depth Resolved Surface Coil Spectroscopy, DRESS

A disk-shaped slice is excited parallel to the surface coil with a frequency selective RF pulse in the presence of a gradient.



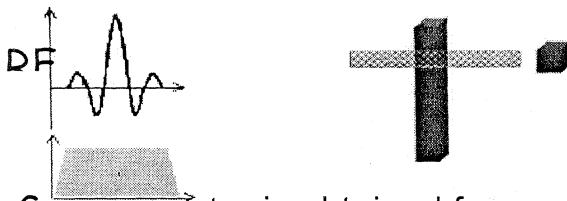
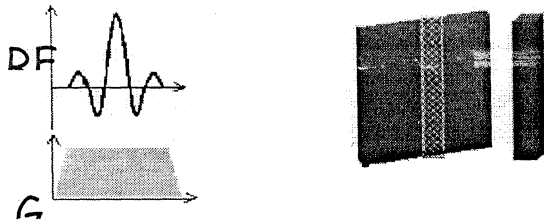
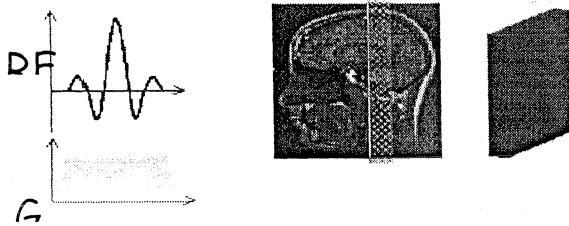
Advantages

Relatively simple
 Suppresses signal from superficial tissue
 Multi-slice acquisition, SLIT-DRESS

Disadvantages

T2 loss
 Partial Localization

Single Volume Localization



Localized spectra is obtained from a single volume of interest (VOI)
 Localization is achieved by sequential selection of three orthogonal slices. The size and location of VOI can be easily controlled
 Anatomic ^1H images are used for localizing the VOI

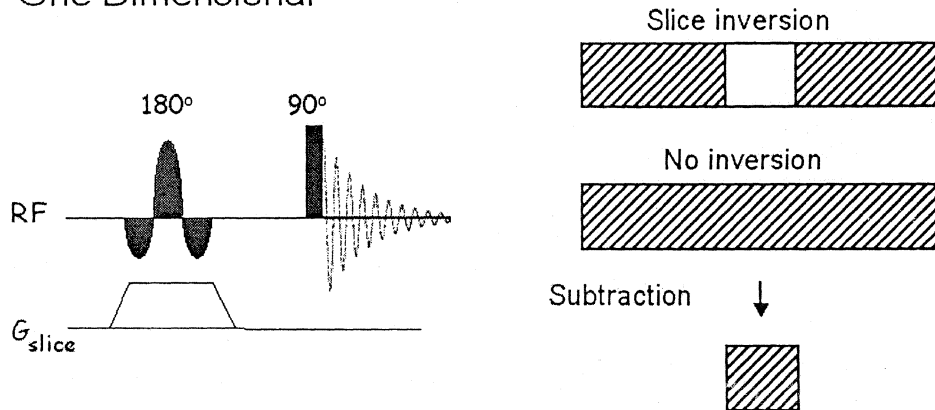
Single Volume Localization

Image selected *in vivo* spectroscopy, ISIS
 Point resolved spectroscopy, PRESS

Stimulated echo acquisition mode, STEAM

Image Selected *In Vivo* Spectroscopy ISIS

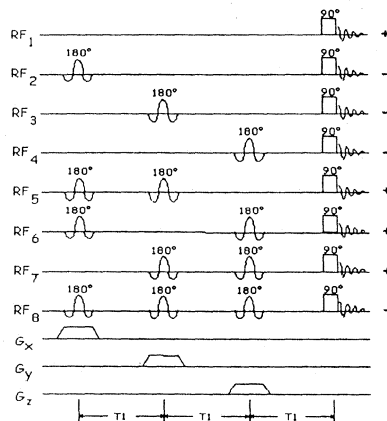
One Dimensional



Two acquisitions with and without inversion of a selected slice are obtained and subtracted

3D ISIS

A set of eight pulse sequences with one, two, or three slice selective inversion pulses are used. The signal is localized to a VOI by adding signals from sequences 1, 5, 6, and 7 and subtracting signals from 2, 3, 4, and 8.



Advantages

No T_2 loss – ^{31}P MRS

Less sensitive to gradient imperfections

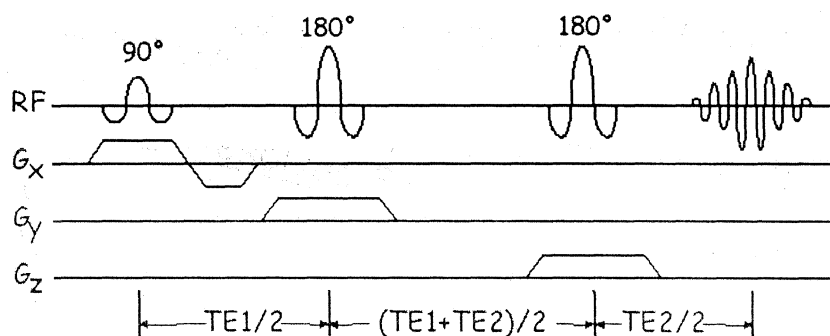
Can be used with a surface coil

Disadvantages

Dynamic range
Subtraction error due to motion

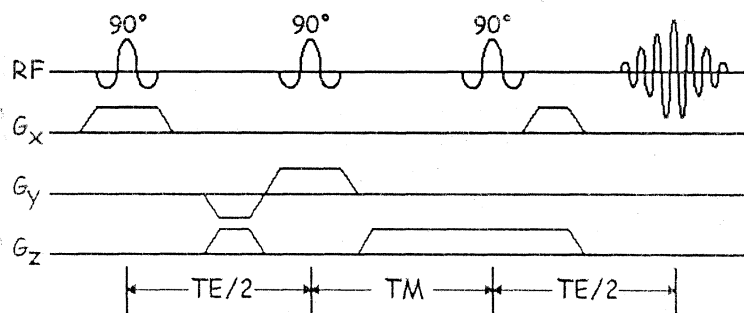
Point Resolved Spectroscopy, PRESS

A slice-selective 90° pulse is followed by two slice-selective 180° refocusing pulses. Achieves localization within a single acquisition
Suitable for signals with long T_2 – ^1H MRS



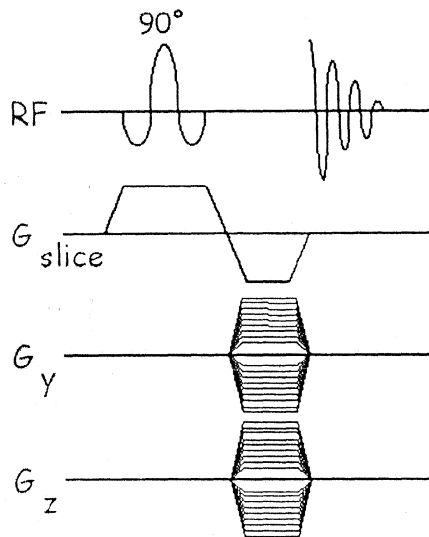
Stimulated Echo Acquisition Mode, STEAM

Three slice-selective 90° pulses form a stimulated echo from a single voxel. Achieves localization within a single acquisition. Only half of the available signal is obtained. Can achieve shorter TE than PRESS



Chemical Shift Imaging

Multiple localized spectra are obtained simultaneously from a set of voxels spanning the region of interest. No gradient is applied during data collection, so spectral information is preserved.



A type of MR spectroscopy imaging, in which spatially localized spectra are obtained using this technique. Although the name suggests that the data will be in the form of an image, the term CSI is applied more generally to any technique in which spectroscopic data are acquired simultaneously from a large volume of the sample and then decoded into a number of spatially resolved spectra. Spatial resolution can be in one, two or three dimensions, but full three-dimensional CSI demands very long acquisition times. The most usual implementation is a slice-selected two-dimensional acquisition. The data can be presented either as a set of spectra representing the chemical composition of each spatially-resolved voxel, or as an image in which the signal intensity depends on the concentration of an individual chemical species. In the latter case separate images can be produced for each chemical that is detected in the individual spectra.

Although most CSI techniques dispense with a read gradient in order completely to preserve the spectroscopic information, an alternative

approach is to employ frequency-selective pulses so that only a single spectral component is excited. In this way, conventional imaging sequences may be employed to produce an image of a particular chemical compound

Multinuclear MR Spectroscopy

Important Nuclei for Biomedical MR

Nucleus	Spin	γ , MHz/T	Natural Abundance	Relative Sensitivity
^1H	1/2	42.576	99.985	100
^2H	1	6.536	0.015	0.96
^3He	1/2	32.433	.00013	44
^{13}C	1/2	10.705	1.108	1.6
^{17}O	3/2	5.772	0.037	2.9
^{19}F	1/2	40.055	100	83.4
^{23}Na	3/2	11.262	100	9.3
^{31}P	1/2	17.236	100	6.6
^{39}K	3/2	1.987	93.08	.05

^1H – Neurotransmitters, amino acids, membrane constituents

^2H – Perfusion, drug metabolism, tissue and cartilage structure.

^{13}C – Glycogen, metabolic rates, substrate preference, drug metabolism, etc.

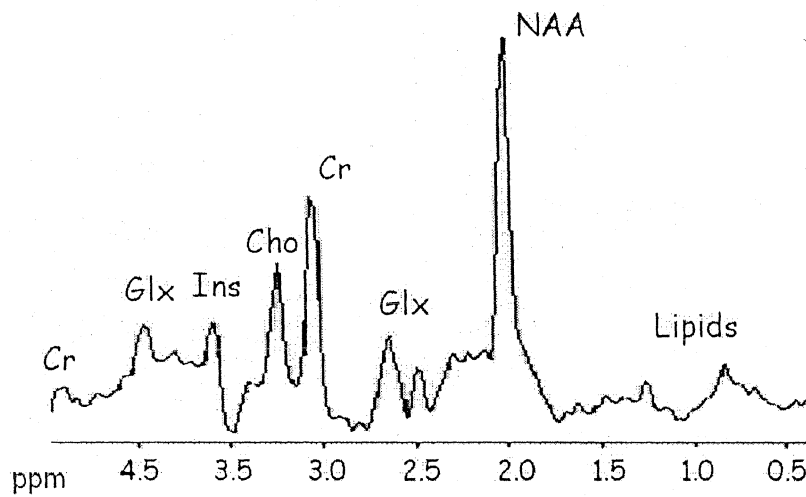
^{19}F – Drug metabolism, pH, Ca^{2+} and other metal ion Concentration, pO_2 , temperature, etc

^{23}Na – Transmembrane Na^+ gradient, tissue and cartilage structure.

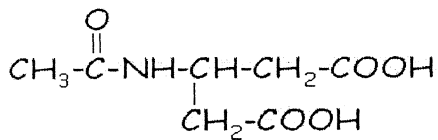
^{31}P – Cellular energetic, membrane constituents, pH_i , $[\text{Mg}^{2+}]$, kinetics of creatine kinase and ATP hydrolysis.

^1H MR Spectroscopy

^1H MR Spectra of the Brain Short TE



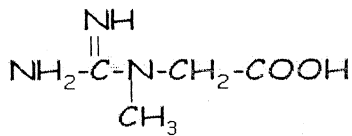
Important ^1H Signals

N-Acetyl aspartate (NAA)

2.02, CH₃
2.52, CH₂
2.70, CH₂
4.40, CH

- NAA is a neuronal marker and indicates density and viability of neurons.

- It is decreased in glioma, ischemia and degenerative diseases.

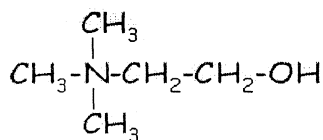
Creatine (Cr), phosphocreatine (PCr)

3.04, CH₃
3.93, CH₂

- Cr is a marker of aerobic energy metabolism

- Cr signal is constant even with pathologic changes and may be used as a control value

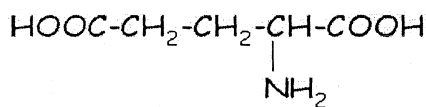
- However, isolated cases of Cr deficiency may occur in children

Choline (Cho), choline compounds

3.24, CH₃
3.56, CH₂
4.07, CH₂

- Cho compounds are involved in phospholipid metabolism of cell membrane.

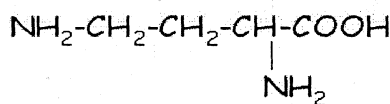
- Increase Cho mark tumor tissue or multiple sclerosis plaques

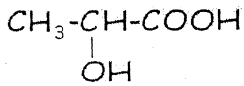
Glutamate (Glu), glutamine (Gln)

2.1, CH₂
2.4, CH₂
3.7, CH

- Glu is a neurotransmitter, Gln a regulator of Glu metabolism

- It is hardly possible to detect their signals separately. The signals are jointly designated "Glx".

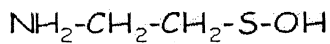


Lactate (Lac)

1.33, CH₃
4.12, CH

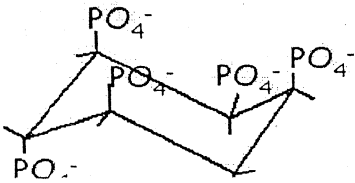
•Lactate is the final product of glycolysis

•It can be detected in ischemic/hypoxic tissue and tumors indicating lack of oxygen

Taurine (Tau)

3.27, NCH₂
3.44, SCH₂

•Cells examination indicates taurine synthesis in astrocytes

Myo-inositol (Ins)

3.56, CH

•Ins marks glia cells in brain

•It is decreased in hepatic encephalopathy and elevated in Alzheimer's disease.

The macromolecules from lipids resonate mainly at 0.8-1.5 ppm and thus may obscure other resonances such as lactate. Lipids are commonly due to contamination from fat included in the voxel or from extraneous unintentional contamination. In tumors, the presence of lipids mostly correlates with necrosis. Radiation necrosis generally contains a large amount of lipids.

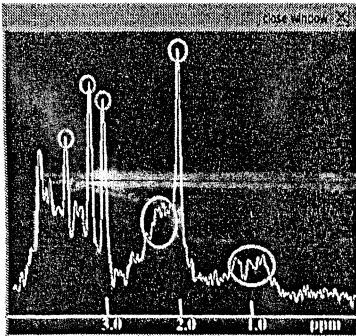
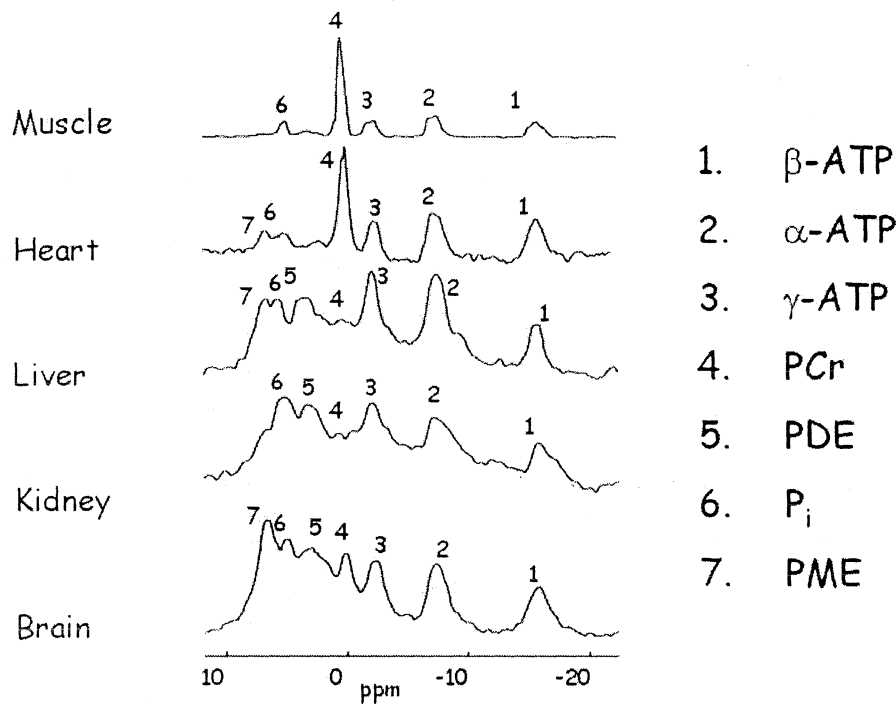


Table 1. METABOLITE PEAKS VISIBLE IN BRAIN AND BRAIN TUMORS

Chemical Structure	Peak Location (ppm)*	Peak Form
Ac methyl	1.92	Singlet
Cholines trimethyl	3.21	Singlet
Creatines CH	3.93	Singlet
Creatines methyl	3.04	Singlet
Gly CH	3.56	Singlet
NAA methyl	2.02	Singlet
NAAG methyl	2.05	Singlet
Suc methylene	2.42	Singlet
Ala	1.43	Doublet
Asp	3.90	Multiplet
Asp	2.81	Multiplet
Asp	2.63	Multiplet
Cholines	3.52	Triplet
GABA	3.01-3.02	Multiplet
GABA	2.30	Multiplet
GABA	1.91	Multiplet
GLN	3.78	Multiplet
GLN	2.46	Multiplet
GLN	2.74	Multiplet
GLU	3.76	Multiplet
GLU	2.35	Multiplet
GLU	2.11	Multiplet
Lac	1.33	Doublet
Myo-inositol	3.63	Triplet
Myo-inositol	3.54	Multiplet
Myo-inositol	3.29	Triplet
NAA	2.70	Multiplet
NAA	2.50	Multiplet
Tau N-CH2	3.27	Triplet
Tau S-CH2	3.43	Triplet
Threonine	3.59	Doublet
Threonine	1.33	Doublet
Valine	3.61	Doublet
Valine	2.27	Multiplet
Valine	1.05	Doublet
Valine	0.99	Doublet

*Resonances at ppm < 1.0 are not listed.

^{31}P MR Spectroscopy



Adenosine triphosphate (ATP)

-16.5	β -ATP
-7.8	γ -ATP
-2.7	α -ATP

ATP is the energy currency in living systems
 β - and γ -ATP have contributions from ADP, NAD and NADH
 β -ATP is uncontaminated and used for quantification

Phosphocreatine (PCr)

0	PCr
---	-----

PCr is used for storing energy and converting ADP to ATP
 It is absent in liver, kidney and red cells
 It is used as an internal reference for chemical shift

Inorganic Phosphate (P_i)3.7 to 5.7 P_i

- P_i is generated from hydrolysis of ATP and increased in compromised tissue
- Its chemical shift is sensitive to pH

Phosphomonoester (PME)

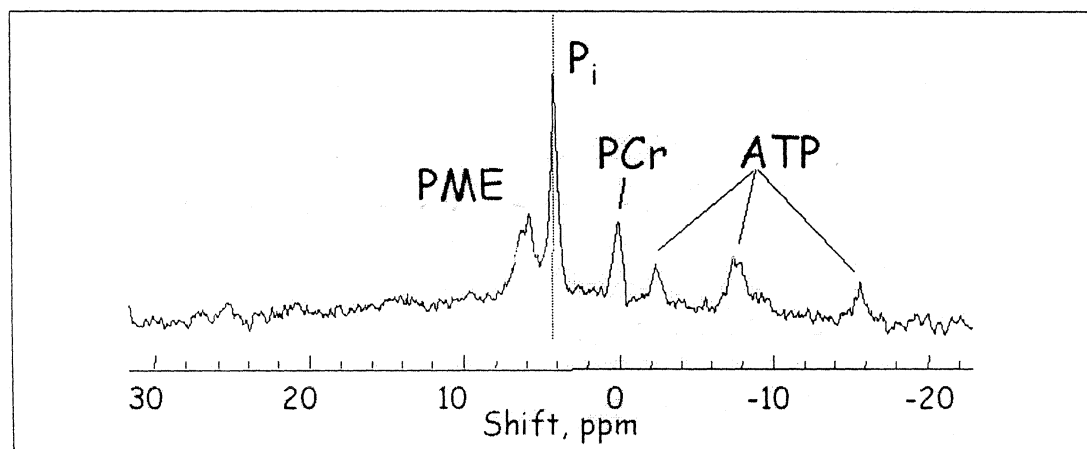
5.6 to 8.1 PME

- PME signal contains contribution from membrane constituents and glucose-6-phosphate and glycerol-3-phosphate.
- It is elevated in tumors

Phosphodiester (PDE)

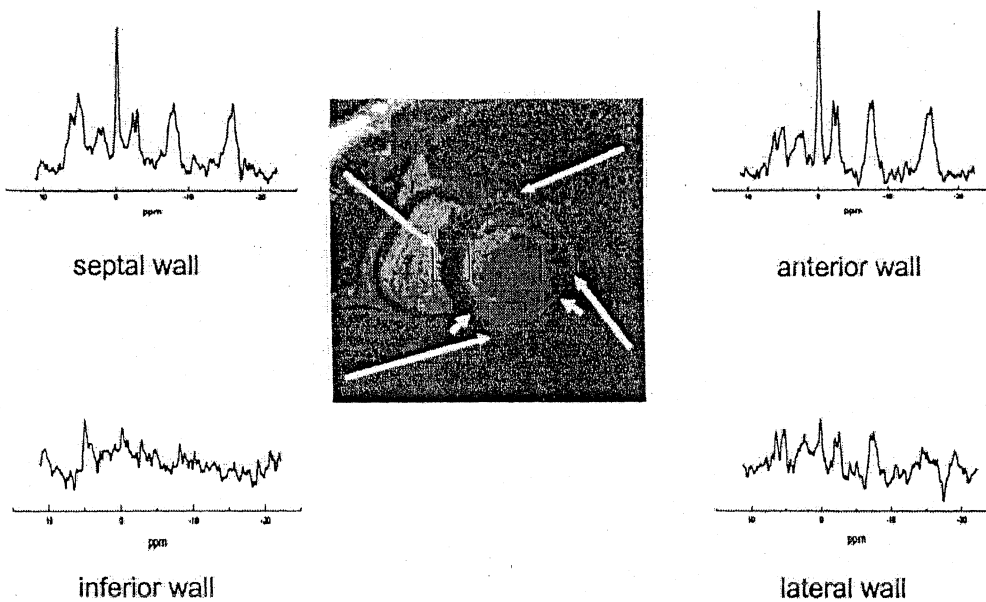
0.6 to 3.7 PDE

- PME signal contains contribution from membrane constituents

Measurement of pH by ^{31}P MRS

$$\text{pH} = \text{p}k_a + \log \left[\frac{\delta_{\text{obs}} - \delta_{\text{H}_2\text{PO}_4^-}}{\delta_{\text{HPO}_4^{2-}} - \delta_{\text{obs}}} \right]$$

Detection of myocardial infarctions by ^{31}P -MR spectroscopy



Beer et al., J Magn Reson Imaging. 2004;20:798-802.

^{23}Na MR Spectroscopy and Imaging

Biological Importance of Sodium

Sodium and other ions are inhomogeneously distributed across the cell membrane. A transmembrane sodium gradient reflects a dynamic equilibrium between Na^+ - K^+ ATPase versus passive or mediated flux. The sodium gradient may be altered in certain diseased states.

^{23}Na is the second most sensitive nucleus for biomedical NMR.

Intra- and extracellular sodium resonate at the same frequency.

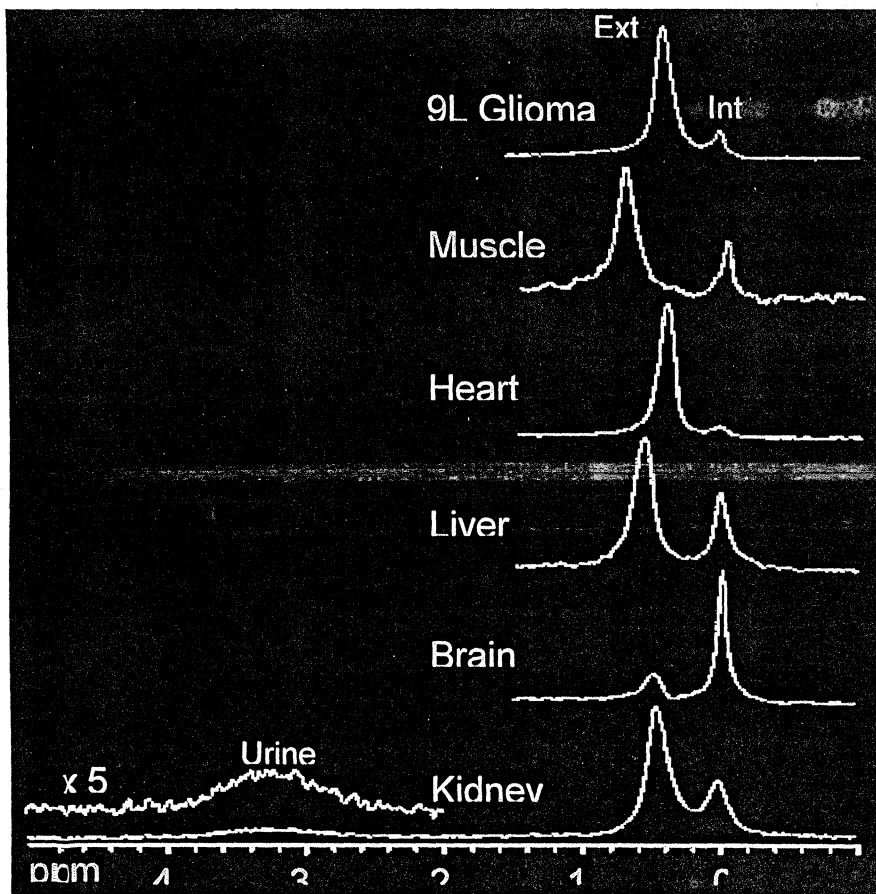
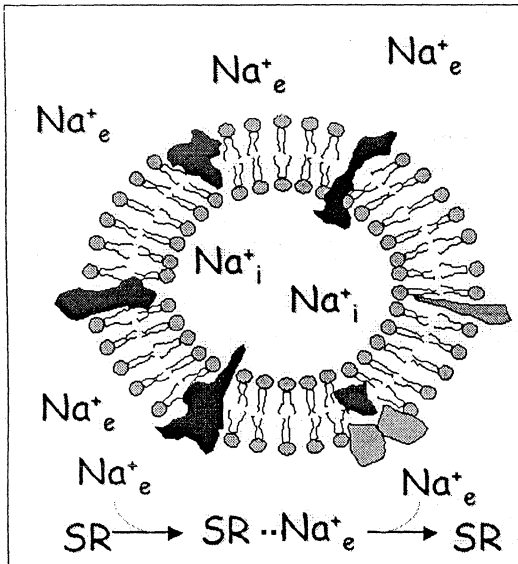
Two approaches to distinguish between different sodium pools:

Paramagnetic Shift Reagents

Multiple Quantum Filters

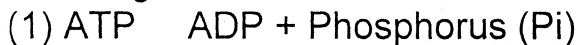
^{23}Na Shift Reagents

SRs are membrane impermeable negatively charged chelates of a lanthanide metal ion. They interact with extracellular Na^+ , causing its signal to be shifted away from intracellular Na^+ .



- (1) N-acetylaspartate (NAA) which is thought to be a neuronal marker that decreases in processes where neurons die;
- (2) lactate which is a product of anaerobic metabolism and may indicate hypoxia;
- (3) excitatory neurotransmitters glutamate and aspartate;
- (4) the inhibitory neurotransmitter gamma-aminobutyric acid (GABA) ;
- (5) cytosolic choline which includes primarily mobile molecules involved in phospholipid membrane metabolism but also small amounts of the neurotransmitter acetylcholine and its precursor choline;
- (6) myoinositol which is important in phospholipid metabolism and intracellular second messenger systems; and
- (7) creatine molecules such as creatine and phosphocreatine which usually have relatively constant concentrations throughout the brain and are often used as relative reference molecules (ie. you may see NAA concentration reported as the ratio NAA/creatine in the literature).

Phosphorus (^{31}P) MRS allows the quantification of ATP metabolism, intracellular pH, and phospholipid metabolism. ATP metabolism quantification is possible because ATP is involved in the following reactions :



where the relative concentrations of ATP and Pi can be determined with ^{31}P MRS

and



where the relative concentrations of phosphocreatine and ATP can also be determined with ^{31}P MRS .

pH can be measured because $\text{H}_2\text{PO}_4^- \rightleftharpoons \text{H}^+ + \text{HPO}_4^{2-}$, and the resonance frequency for H_2PO_4^- is different from HPO_4^{2-} . As the shift between these two molecules is so rapid, they present as one spectrographic peak. However, when one of these compounds is present at its equilibrium, the peak is shifted closer to that compound's true precession frequency, allowing changes in pH to be measured by its position ³⁴. Mobile phospholipids, including phosphomonoesters (PME - putative cell membrane building blocks) and phosphodiesteres (PDE - putative cell membrane breakdown products) can also be measured, supplying information on phospholipid membrane metabolism.

MRS can be used to identify regional biochemical abnormalities.

For example, ^{31}P -MRS studies of euthymic bipolar patients have revealed decreased frontal lobe PME (cell membrane building blocks) compared with healthy controls. However, when bipolar patients become either manic or depressed, their PMEs increase. These findings appear to be unrelated to medication treatment. The finding of decreased frontal PMEs in euthymic bipolars has also been demonstrated in schizophrenia and speculatively accounts for the finding of decreased frontal lobe metabolism in both of these disorders. The schizophrenia finding also appears to be medication-independent³².

MRS may also be of future help in the differential diagnosis of certain psychiatric diseases such as dementia. In normal aging, there is a decrease in PMEs and increase in PDEs^{36,37}. In early Alzheimer's Dementia, there appears to be increased PMEs which may help distinguish it from normal aging³⁸. Researchers have also found increased myoinositol and decreased NAA levels in Alzheimer's Dementia compared with healthy controls^{39,40}. Some believe that a decrease in NAA coupled with an increased myoinositol level helps in differentiating probable Alzheimer's Dementia from healthy age-matched controls as well as other dementias (usually decreased NAA but normal myoinositol levels)³⁹.

With MRS, changes in metabolic activity can be measured over time within an individual scanning session. For instance, Dager et al. (1995) used ^1H MRS to measure changes in lactate concentration with controlled hyperventilation in panic disorder patients and healthy controls⁴¹. MRS can also be used to measure changes in metabolic activity between sessions, such as before and after medication treatment. For example, Satlin et al. (1997) used ^1H MRS to measure midparietal lobe cytosol choline levels in 12 Alzheimer's subjects before and after treatment with Xanomeline, an M1 selective cholinergic agonist, or placebo⁴². Additionally, MRS can be used to measure drug levels of certain psychotropic drugs. The magnetic elements ^7Li and ^{19}F do not naturally occur in the human body but they are found in psychotropic drugs; lithium for ^7Li and fluoxetine and stelazine for ^{19}F . For example, studies have consistently found that the brain concentrations of lithium are about 0.5 that of serum Li levels and correlate with treatment response.

For psychiatry, MRS is a research tool at this time. In neurology and neurosurgery, however, MRS is starting to be used in the

characterization of tumor, stroke, and epileptogenic tissue and in presurgical planning^{26,34}.

Limitations

MRS is restricted to studying mobile magnetic compounds. As neurochemical receptors are not usually mobile, they cannot be measured with MRS. Thus, receptor-ligand studies are still the domain of SPECT and PET. Another problem with MRS is that due to the low concentrations of many of the imaged substances, larger areas than with water are needed to obtain detectable signals. Larger volume units imaged over longer periods are thus used with this technique, limiting both temporal and spatial resolution compared with conventional MRI, and BOLD-fMRI. However, stronger magnetic fields which can spread out precession frequencies over a wider range may improve this resolution (ie. a magnetic field twice as strong will double the difference between two substances' precession frequencies, thus increasing resolution). Stronger magnetic fields may also allow detection of compounds that are currently considered to be in too low a concentration to be seen with current MRS equipment.

THE SEQUENTIAL STEPS OF AN MRs EXAMINATION

PATIENT POSITIONING

GLOBAL SHIMMING

ACQUISITION OF MR IMAGE FOR LOCALIZATION

SELECTION OF MRS MEASUREMENT SEUENCES & PARAMETERS

SELECTION OF AVOLUME OF INTEREST

LOCALIZED SHIMMING

OPTIMIZATION OF WATER SUPPRESSION

MRS DATA COLLECTION

MRS DATA PROCESSING AND DISPLAY

PATIENT POSITIONING

Patient is positioned in the standard head coil Head should be firmly supported to prevents the involuntary. Line broadening ,incorrect localization ,loss of signal.SVS- it causes to collect signal from out side the volume. CSI-Phase differences,ghosting on MR images ISIS-Motion between the multiple measurements leads to incomplete cancellation unwanted signals

GLOBAL SHIMMING

Optimizing the MF homogeneity It is doing at the Beginning of an imaging or MRS. It is Fast and it ensure good quality images & spectra

It provides coil tuning transmitter frequency & power adjustments

An alternate approach –store shim sets as data files that may be recalled and used when needed

POSITION OF THE VOLUME

Voxel is placed over the abnormal tissue. The size of the voxel is adjusted in order to minimize the normal tissue .Voxel should avoid the scalp, bone sinuses, blood vessels ,csf ,calcification & necrotic area. Minimize the mixture of gray & white matter

LOCAL SHIM

Optimize the MF homogeneity over the selected volume. Good local shim produce a narrower metabolite peak better spectral resolution ,improved SNR. Selected MR sequences and it use the water signal. The procedures consist of varying shim currents, one at a time until the narrowest water peak is produced. FWHM– 4-10 HZ, 8 –27 CM³

WATER SUPPRESSION

Large water signal make it difficult to observe weak signal from metabolites

CHES- Frequency selective 90° pulse , followed by dephasing gradient

100-single CHES

DRYSTEAM

WET– 4 frequency selective RF pulse with different flip angle

It produce better water suppression than CHES

WEFT

MEASUREMENT

3-6 min –SVS

12min-CSI

DATA PROCESSING & DISPLAY OF SPECTRA

Automatically or Manually

Time domain signal into frequency domain signal

The spectrum is typically displayed with water peak -4.7ppm
Chemical shift of water with respect to TMS
The major metabolites peaks between $0-4.3\text{ppm}$

PROCESSING OF THE MR SIGNAL

Time domain signal –ADC offset correction ,zero filling, apodisation
,correction from residual eddy currents ,removal of the water peak
Following the FT—Phase correction ,base line correction ,curve
fitting

PHASE CORRECTION

Separate pure absorption & pure dispersion mode into the real and
imaginary part of the complex spectrum
Corrected with constant or zero order phase factor

BASE LINE CORRECTION

Base line of the MR spectrum will be flat until the resonance peak is
reached.Delay between the RF-excitation and beginning of the
collection period produces a rolling baseline.Un subtracted water
will cause sloping of the base line

PEAK IDENTIFICATION

Determination of the peak position and characteristics
Verification of the presence or absence of given metabolites

PEAK EDITING

Separating overlapping peaks for identification

PEAK AREA CALCULATION

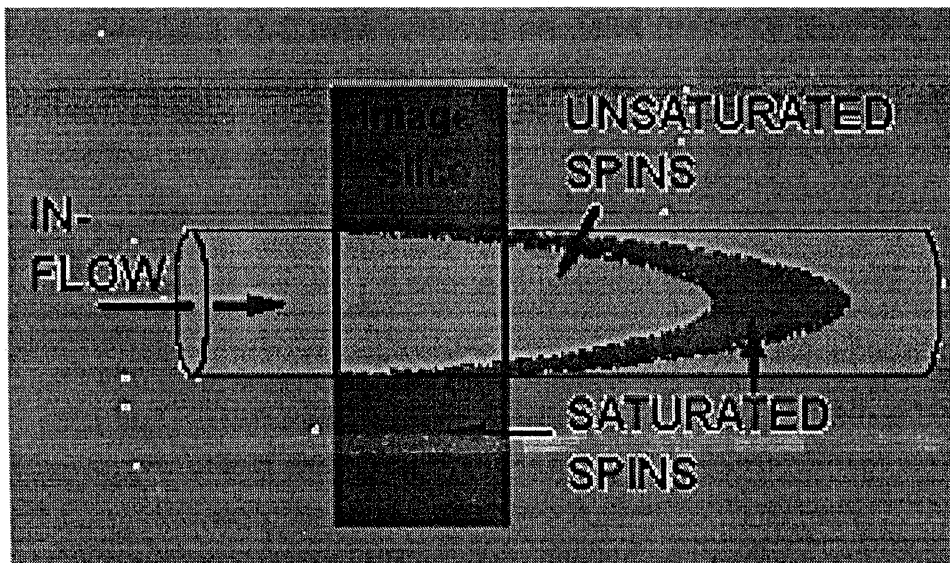
Area under a peak is directly proportional to the number of spins
contributing to the peak

MR ANGIOGRAPHY

Time-of-flight angiography (TOF), phase-contrast angiography (PCA) and contrast enhanced angiography (CEA) are the three common MR angiographic techniques. Flow related enhancement is the basis of time-of-flight angiography.

1. Time-of-flight MR angiography (TOF)
2. Phase contrast MR angiography (PCA)
3. Contrast enhanced MR Angiography (CEA)

Flow-related Enhancement



Flow-related enhancement is an enhancement of flowing blood seen on gradient echo pulse sequences as well as in entry slices of multislice spin echo sequences. This enhancement is a result of inflow of unsaturated (completely relaxed) spins into a slice plane or imaging volume between RF excitations. Stationary spins within the imaging volume will undergo incomplete T1 relaxation between RF excitations resulting in less signal following the next RF pulse when compared to inflowing, completely relaxed spins in flowing blood. The distance that the unsaturated blood can extend into an imaging

volume and therefore the degree of enhancement is proportional to the TR and the velocity of the blood. The use of gradient motion nulling (flow compensation) improves the flow-related enhancement in gradient echo sequences.

1. Time-of-Flight MR Angiography

Time-of-flight (TOF) angiography is based on the phenomenon of flow-related enhancement of spins entering into an imaging slice. As a result of being unsaturated, these spins give more signal than surrounding stationary spins. With 2-D TOF, multiple thin imaging slices are acquired with a flow-compensated gradient-echo sequence. These images can be combined by using a technique of reconstruction such as maximum intensity projection (MIP), to obtain a 3-D image of the vessels analogous to conventional angiography

With 3-D TOF, a volume of images is obtained simultaneously by phase-encoding in the slice-select direction. An angiographic appearance can be generated using MIP, as is done with 2-D TOF. Several 3-D TOF volumes can be combined to visualize longer segments of vessels. 3-D TOF angiography will allow greater spatial resolution in the slice-select direction than 2-D TOF; however, with thick volumes and slow flowing blood, loss of signal is seen with the 3-D TOF method.

Multiple Overlapped Thin Slab Acquisitions (MOTSA).

3D volume is subjected to multiple RF excitations (as you go from proximal to distal) it causes saturation of slow flowing spins. That can be avoided by using overlapping slabs (MOTSA), this with the expense of increased imaging time. While 3D slabs are stacked together a dark line crosses the image called Venetian blind artifact. This can be avoided by giving 30 – 50 % overlap.

2. Phase-contrast MR Angiography

Spins that are moving in the same direction as a magnetic field gradient develop a phase shift that is proportional to the velocity of the spins. This is the basis of phase-contrast angiography. In the simplest phase-contrast pulse sequence, bipolar gradients (two

gradients with equal magnitude but opposite direction) are used to encode the velocity of the spins. Stationary spins undergo no net change in phase after the two gradients are applied. Moving spins will experience a different magnitude of the second gradient compared to the first, because of its different spatial position. This results in a net phase shift. This information can be used directly to determine the velocity of the spins. Alternatively, the image can be subtracted from one acquired without the velocity encoding gradients to obtain an angiogram.

3. Contrast enhanced MR Angiography:

Principles of 3D Gadolinium-enhanced MR Angiography

Gadolinium is one of the rare earth elements in the transition group IIIb of the periodic table. Actually, it is not rare at all, but a rather common element found throughout the earth's crust. It has eight unpaired electrons in its outer shell, which causes its paramagnetic effects. Gadolinium by itself can cause heavy metal poisoning. However, when bound to a chelator, it is safe for intravenous injection, yet remains paramagnetic. It shortens the T1 of blood in the region of the gadolinium molecule according to the following equation (1):

$$1/T1 = 1/1,200 + (R1 \times [Gd]),$$

where R1 is the T1 relaxivity of the gadolinium chelate, [Gd] is the gadolinium concentration in the blood, and 1,200 is the blood T1 (in msec) without gadolinium.

Effect of [Gd] on Blood T1

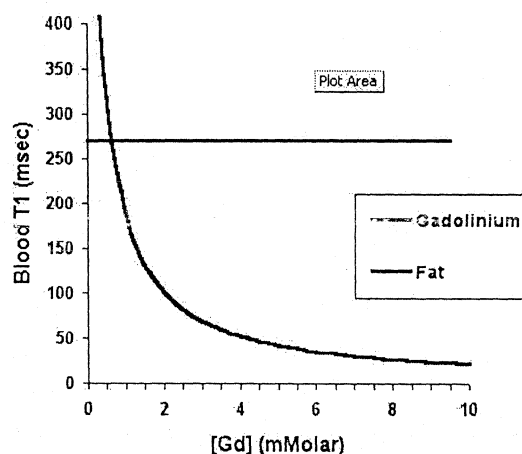


Figure 1. Effect of gadolinium concentration on blood T1. (The graph was calculated with Microsoft Excel 97 [Microsoft, Redmond, Wash] by using the above formula.)

The blood [Gd] must be greater than 1.0 mmol/L for T1 to be less than 270 msec, which is the T1 of fat at 1.5 T (1, 5). This is the property which is important for increasing the MR signal intensity of blood on contrast-enhanced SPGR images. Note that this T1 shortening effect is maximized by using gadolinium chelates with the highest relaxivity and by having a high gadolinium concentration in the blood.

K Space

The most important effect is related to how MR data are mapped in k space. K space, or Fourier space, does not map to the image pixel by pixel. Rather, the information within k space determines spatial frequency features of the image. The low spatial frequency information, in the center of k space, dominates image contrast, while the higher spatial frequency data, at the periphery of k space, determines image detail. To obtain an arterial-phase image in which arteries are bright and veins are dark, it is essential that the central k-space data (ie, the low spatial frequency data) are acquired while the gadolinium concentration in the arteries is high but relatively lower in the veins. The presence of contrast agent is not as important for acquisition of peripheral k-space data. This trick allows a relatively long MR acquisition to achieve the image contrast associated with a brief window of time. That brief window of time is the instant when central k-space data are acquired. Therefore, it is critical to time the bolus for maximum arterial [Gd] during acquisition of central k-space data. With perfect bolus timing, high S/N arterial-phase images are possible with smaller doses of gadolinium.

Gadolinium Extraction

A second important effect is the extraction of gadolinium in the systemic capillary beds. This extraction results in venous blood tending to have a lower concentration of gadolinium relative to arterial blood, even for relatively long, sustained infusions lasting several minutes. This effect is not present in the cerebrovascular circulation because of the blood-brain barrier. Consequently, arterial-phase

imaging in the central nervous system is more difficult.

Infusion Rate and Cardiac Output

A third effect is the relationship between arterial gadolinium concentration, the infusion rate, and cardiac output as follows:

$$[\text{Gd}]_{\text{Arterial}} = \text{Injection Rate} / \text{Cardiac Output}$$

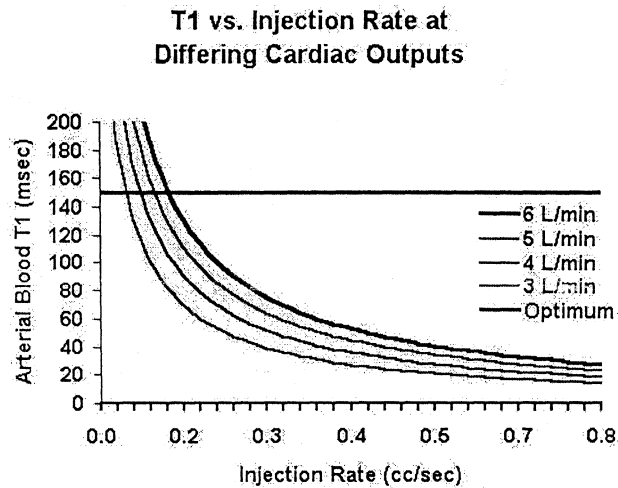


Figure 2. T1 versus injection rate at differing cardiac outputs.

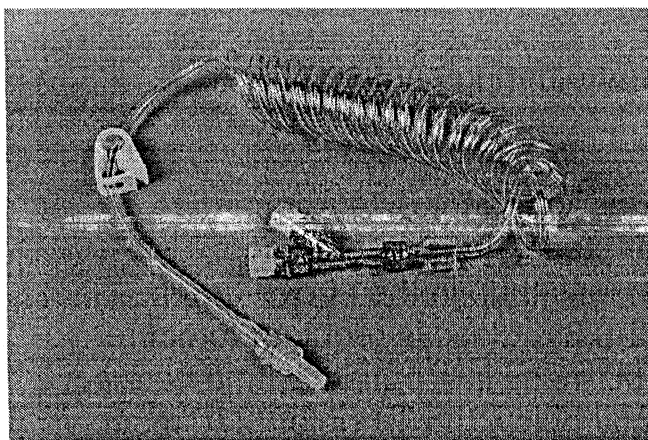
This graph is a computer model generated for different cardiac outputs. Note that this graph is for static, nonmoving blood. The actual T1 of moving blood is even less than in the above graph. Notice that an injection rate of 0.2-0.3 mL/sec is required to decrease the blood T1 value to less than 150 msec in a patient with a cardiac output of 5 L/min. This is sufficiently below the T1 of fat (270 msec at 1.5 T) so that only the gadolinium in blood produces high signal intensity on T1-weighted SPGR images.

Arterial [Gd] is maximized by relaxing the patient to reduce cardiac output and by using a high infusion rate. However, fast infusion rates lasting for at least half of the acquisition time require large doses of gadolinium. The dose can be kept to a reasonable level by scanning rapidly. Fast acquisitions (<45 seconds) are possible with high performance gradient systems that allow short repetition and echo times, without having to make the bandwidth too wide. Fast

acquisition has the additional benefit of making it possible for the cooperative patient to suspend breathing and to hold perfectly still.

Saline Flush and Intravenous Tubing

It is important to use intravenous (IV) line tubing that allows simultaneous attachment of separate syringes for the contrast agent and saline flush.



The developed at the University of Michigan, has one-way valves that allow automatic switching between the contrast agent injection and saline flush so that there will be one continuous bolus with no gaps. By using the same tubing set for all patients receiving dynamic contrast agent injection, the operator becomes familiar with performing the injections and especially with the resistance to injection. It is then easier to concentrate on correctly timing the bolus and instructing the patient to suspend breathing.

At least 20 mL of saline is recommended to adequately flush the

contrast agent through the IV tubing and arm vein. By starting with a 30-mL saline-filled syringe, it is possible to initially prime the SmartSet with 8 mL of saline, test the IV once or twice with a 1-mL saline injection, and still have 20 mL left for the dynamic flush.

Gadolinium Dose

When beginning, we recommend use of two bottles (42 mL) for the average-size patient and three bottles (63 mL) for patients weighing more than 100 kg. Once you learn how to time the contrast agent injections perfectly, you will find it is possible to reduce the dose and still obtain diagnostic images.

Contrast Agent Bolus Timing

Perfect contrast agent bolus timing is crucial to ensure that the maximum arterial [Gd] occurs during the middle of the acquisition, when central k-space data are acquired. It is also essential that [Gd] not change too rapidly, as this will create a "ringing" artifact. Minimizing the ringing artifact requires that the contrast agent infusion last for at least half the duration of the 3D image acquisition. Bolus timing is difficult because the time required for the contrast agent bolus to travel from the injection site (typically an antecubital vein) to the artery being imaged is highly variable. For renal arteries, it may be only 10 seconds in a young, healthy person with a central intravenous line, or it may be as long as 50 seconds in an older patient with congestive heart failure and an intravenous line in the hand or wrist.

Timing for Long Acquisitions

For long acquisitions, lasting more than 100 seconds, timing is easy because errors of 10-15 seconds are small relative to the total scan duration. Use sequential ordering of k space, so that the center of k space is collected during the middle of the acquisition. Sequential ordering tends to result in fewer artifacts. Begin injecting the gadolinium just after initiating imaging. Finish the injection just after the midpoint of the acquisition, being careful to maintain the

maximum injection rate for the approximately 10-30 seconds prior to the middle of the acquisition. This will ensure a maximum arterial [Gd] during the middle of the acquisition, when central k-space data are collected. To ensure full use of the entire dose of contrast agent, it is useful to flush the IV tubing with 20 mL of normal saline. This can be facilitated by using the SmartSet, which has ports for simultaneous attachment of contrast agent and saline syringes and valves for automatic switching between syringes. In this way, there is no delay between finishing the contrast agent injection and beginning the saline flush.

Timing for Fast (Breath-hold) Scans

For fast scans, less than 45 seconds in duration, contrast agent bolus timing is more critical and challenging. This is because bolus timing errors of 15 seconds can ruin a fast breath-hold scan. There are several approaches to determining the optimal bolus timing for these fast scans. The simplest, although least successful approach, is to guess on the basis of patient age, cardiac status, presence of aortic aneurysmal disease, and IV location. For a typical breath-hold scan duration of 35-45 seconds in a reasonably healthy patient with an IV site in the antecubital vein, a delay of approximately 10-12 seconds is appropriate. Therefore, in this scenario, begin the injection, and then 10 seconds later start imaging while the patient suspends breathing. If there is no convenient clock available to time this delay, take advantage of the natural rhythm of the patient's respiration. One deep breath in followed by a deep breath out takes approximately 4 seconds. Two breaths are eight seconds, followed by a deep breath in, 10 seconds, which represents the optimum delay between start of injection and beginning of scanning. If a patient is older and has a history of cardiac or aortic aneurysmal disease, add one or two extra breaths to the delay. Also, if the IV site is in the wrist, add an extra breath to the delay. Alternatively, if the patient is a marathon runner or you are injecting via a central line, it may be suitable to use only 1½ breaths of delay, or 6 seconds. A firm injection is necessary to keep the contrast agent bolus together. However, if the injection is too vigorous, it may cause rupture of the vein, with resulting extravasation of the contrast agent.

More reliable and precise techniques for determining the contrast

travel time are also available. These include using a test bolus to precisely measure the contrast travel time, using an automatic pulse sequence that monitors signal in the aorta and then initiates imaging after contrast is detected arriving in the aorta (Fluoroscopic triggering or MR SmartPrep), or imaging so rapidly that bolus timing is unimportant. A typical monitor signal graph is shown below.

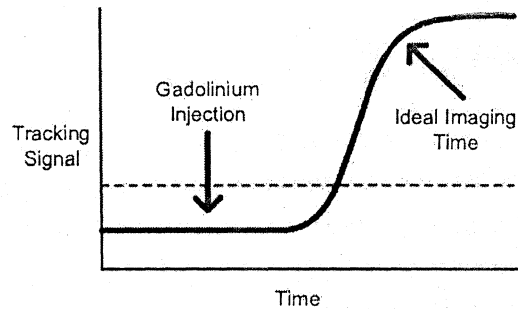


Figure 3. Tracking signal versus time.

Blue line - MR signal intensity in volume of interest.

Green line - Level at which signal is 100% above baseline.

Red arrows - Time of injection and optimal time for imaging.

Note that the centric ordering of k space for the triggered acquisitions may create artifacts if the contrast agent bolus is still arriving when the scan is started. Sometimes this artifact may be reduced with sequential ordering. This artifact may also be reduced by delaying 5-8 seconds after detecting the leading edge of the bolus to give the contrast agent time to flow in completely and reach the plateau phase of the bolus.

Postprocessing of MR Data

Substantial improvement in image quality, and especially image contrast, can be attained through postprocessing techniques.

Zero Padding

Image resolution can be increased with interpolation. One particularly useful interpolation scheme is known as zero padding. This involves filling out peripheral lines of k-space data with zeroes prior to performing the Fourier transform. Although no additional time is

required for data collection, the Fourier transform will reconstruct more images with a smaller spacing. For example, with two-fold zero padding, if the partition thickness is 3 mm, the Fourier transform will reconstruct additional images that also have a 3-mm slice thickness but at 1.5-mm spacing with 50% overlap. This helps eliminate volume averaging and creates smooth visualization of small vessels on the reformatted maximum intensity projection (MIP) images. If available, two-fold zero padding in the slice direction is recommended.

MR Digital Subtraction Angiography (DSA)

Image contrast can be improved by digital subtraction of precontrast image data from dynamic, arterial, or venous phase image data. This subtraction can be performed either slice-by-slice or prior to the Fourier transform by using a complex subtraction method. The improvement in contrast achieved with DSA may reduce the gadolinium dose required. However, there must be no change in the patient position between the precontrast and dynamic contrast-enhanced imaging. This requirement for no motion is easily met in the pelvis and legs, which can be sandbagged and strapped down. It is more difficult to achieve in the chest and abdomen, where respiratory, cardiac, and peristaltic motions are more difficult to avoid. Note that complex subtraction is generally performed automatically by the scanner before creating any of the images.

Multiplanar Reconstructions

Reformations and MIPs are essential for optimal assessment of vascular anatomy. Single-voxel-thick reformations and narrow subvolume MIPs show bifurcations and branch vessels in profile. This is important because atherosclerotic disease tends to be most severe at branch points. By creating subvolume MIPs of the 3D image data, these techniques help unfold tortuous vessels and eliminate the confusing overlap of vascular anatomy.

Creating an MIP Image

One approach to performing a subvolume MIP is to first load the entire 3D volume of arterial phase image data into the computer workstation 3D analysis program. Display a coronal MIP of the entire

volume, an axial reformation, and an oblique view. On the coronal view, move the dot (which tracks the location) cranially and caudally while watching the axial reconstruction window to find the renal arteries. Display this subvolume of sagittal data as an MIP. Make this oblique MIP thick enough to encompass most of the aorta. Be certain to align the axis of the subvolume MIP so that it is parallel to the origin of the vessel. Although the entire length of the vessel may not be seen on this image, it will be an accurate representation of the vessel's origin, with no overlap from the aorta. This may then be repeated by moving the tracker dot on the axial image and watching the oblique view to create a sagittal view of the celiac and superior mesenteric arteries. This will show the celiac and superior and inferior mesenteric arteries, as well as the anterior and posterior margins of the aorta to best advantage.

Time-Resolved Peripheral MRA

Time-resolved Gd-MRA techniques are useful when they are combined with 3D Gd-MRA. These are 3D TRICKS and 2D projection MRA of the calves and the feet. Time-resolved images shows the contrast arrival time to the calf and gives an idea about how fast the blood flow is. The bolus timing of subsequent 3D bolus chase Gd-MRA can be determined based on this information. If the contrast arrival time to the calf is less than 25 s, acquisition of 3D Gd-MRA without venous enhancement is almost impossible. In these fast flow patients, tourniquets (Smart Tourniquet) applied on the thighs are especially important for decreasing venous contamination.

Time-resolved 2D-projection MRA (calves and feet)

SPGR pulse sequence which acquires 30 sequential images of a thick slab (60-120 mm). Display rate of the images is 2 frames/s.

Position the coil over the area from above the knee to mid-tibia,
Landmark on mid-coil

First acquire the pre-contrast sagittal localizer images

On sagittal images prescribe a coronal thick slab for imaging the trifurcations of both legs

For contrast enhanced image acquisition, fill the tubing with Gd, and then attach to the angiocatheter. Begin administering 6 ml of Gd bolus by hand as fast as possible and simultaneously press the 'Scan' button. As soon as the Gd administration finishes, switch to 20

ml saline flush at the same rate. If you are using an injector; 3.5 ml/s Gd followed with 25 ml saline flush as 3.5 ml/s.

Move the coil down to image ankle and the feet. Position with cushion under knees so that foot will point more horizontally. The toes should be sticking out slightly beyond the end of the coil. Tape feet together to remind the patient to hold still.

First acquire coronal localizer images. Coronal T1 of ankle-foot is ready to run. No adjustment necessary.

Prescription of sagittal 2D projection MRA: On the coronal localizer (choose a slice 10-30 mm posterior to isocenter) set slab thickness to width of foot-ankle. Use "shift" to place one sagittal thick slab on each ankle-foot.

Use sufficient phases so that total scan time is > 1 minute. If patient is older or expected to have really slow flow, make sure there are sufficient slices to image for at least 90 seconds.

Using similar parameters and another bolus injection of 6 mL of gadolinium, acquire sagittal time-resolved ankle and foot images.

An image obtained before the contrast arrival (visually identified on unsubtracted images) was used as the mask for complex subtraction from the subsequent image data. This is done both calves and feet)

Reconstruction from remote terminal:

Cardiac MRI

Magnetic resonance imaging (MRI) of the heart and great vessels has improved substantially over the past decade, and it is entering the mainstream of diagnostic imaging. As commercial cardiac MRI systems become more available, demand will grow exponentially. The demand for cost-effective, noninvasive, and safer technology ensures that cardiac MRI will become a mainstay in cardiac imaging. Diagnosticians must embrace and master the techniques of cardiovascular imaging and image interpretation if the full potential of cardiac MRI is to be realized.

Cardiac MRI already is considered the procedure of choice in the evaluation of pericardial disease and intracardiac and pericardiac masses, for imaging the right ventricle and pulmonary vessels, and for assessing many forms of congenital heart disease, especially

after corrective surgery. and the assessment of ischemic heart disease

Complexity of cardiac anatomy

The cardiac anatomy is complex, and cardiac structures have different appearances depending on the imaging plane and other options selected. The most useful imaging planes are those parallel and perpendicular to the cardiac axes. The imager needs to be aware that these planes are not based on external landmarks. Obtaining images in these double-oblique planes requires the use of multiple localizing sequences and knowledge of the target anatomy.

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Choice of pulse sequences

The choice of pulse sequences and parameters that affect the appearance of the images obtained with a given sequence greatly depends on the clinical question. The limitations and advantages inherent in particular imaging methods such as spin-echo (SE) or gradient-echo (GE) techniques must be considered in the selection of pulse sequences. GE uses a single excitation pulse to produce imaging signals, whereas SE uses both 90° and 180° pulses. Cardiac morphology and anatomy are appreciated better with SE images, which show higher contrast. Contrast is related to signal intensity differences, which are chiefly determined by the proton density, T1

and T2 relaxation time differences, flow and motion effects, and magnetic susceptibility, as they are related to the pulse sequence applied. In general, tissue characteristics and pathologic changes are generally more pronounced on SE images. Signal intensity on GE images largely reflects differences in proton density; these images inherently show lower tissue contrast. As a result, SE is the preferred method for imaging cardiac masses and diseases of the pericardium.

Blood pool appearance

The appearance of the blood pool is a factor in selecting the acquisition protocol. In general, blood flows too quickly through an imaging section for it to be exposed to both the 90° and 180° pulses. Because of this, SE images can fail to depict structures that move out of the imaging plane. Because signal generation at SE imaging depends on refocused spins (protons stimulated by a 90° excitation pulse and refocused by a 180° inversion pulse), the blood pool, by virtue of not being exposed to both pulses, appears as a region of signal void; the result is termed black-blood imaging. Darkness of the blood pool can be further promoted by applying a saturation signal and/or a double inversion recovery (DIR) technique.

Black-blood imaging improves conspicuity of the endocardial borders and helps demarcate intracardiac masses from normal structures. The improved delineation of vessel walls with SE also enables superior anatomic evaluation. GE, on the other hand, involves the application of only 1 section-selective excitation. After that excitation, blood contributes to the image even after it moves out of the imaging plane. Such contrast is termed bright-blood imaging. Complex branching vessels, such as the pulmonary veins or arteries, are better depicted with this technique. Another parameter that affects the appearance of the blood flow is the echo time (TE). With GE, the signal depends on gradient reversal, without the additional spin refocusing of SE. Spins that enter the section during acquisition produce heightened signal. The bright blood pool can be used to distinguish structures such as bronchi from vessels.

Spatial and temporal resolution

Spatial and temporal resolution

Spatial resolution is a concern with any imaging modality. Fortunately, the size of the major cardiac structures is well within the limit for MRI. Spatial resolution generally is determined by the field of view (FOV) and the number of phase-encoding steps. For comparable FOVs and matrix sizes, spatial resolution is equal with SE and GE techniques. Their major difference is with temporal resolution. Adequate depiction of the atrial septum or valve leaflets depends on the temporal resolution, as well as the spatial resolution. Regarding time resolution, GE imaging has advantages compared with SE imaging. The assessment of dynamic heart structures is performed with cine GE acquisitions because its temporal resolution is better. A trade-off exists between temporal resolution and spatial resolution. For typical static SE imaging, one may use a matrix of 256×256 and a FOV of 34 cm. To make a movie of the beating heart, one may select a GE technique and reduce the number of phase-encoding steps, for example, to 128. The spatial resolution decreases from 1.33×1.33 mm to 1.33×2.66 mm accordingly. That resolution is sufficient to depict cardiac wall motion and to calculate the ejection fraction, but higher resolution may be required if detailed evaluation of the cardiac valves is desired.

ECG gating

The most important factor in the acquisition of diagnostic images is the quality of the MRI system's ECG gating. If the gating system lacks consistency or cannot tolerate irregularities of the cardiac rhythm, cardiac images are severely impaired. Without dependable gating, many pulse sequences become useless, and if those sequences are required, the examination is unlikely to yield interpretable data. As a first step, a good ECG tracing must be obtained. Standard MRI-compatible electrodes are placed on the anterior or posterior chest wall of the patient. Depending on the gating system, typically 3 or 4 electrodes are used. The quality of the electrodes is of minor importance. Experience at the author's institution shows no difference in the performance among electrode pads on the basis of their cost. Electrode contact is important. Thorough cleansing of the contact area with alcohol and/or acetone is sufficient. Body hair should be removed in the area where the electrode is placed because it can

cause loss of contact. With older systems, adequate tracings required that electrodes be placed within 15 cm of each other, and the leads had to be straight and could not have loops. The advent of fiberoptic lead systems has simplified lead placement and have decreased the concern about looping or degradation of the ECG tracing. However, overlapping electrodes can significantly degrade the ECG tracing; this is a concern in small pediatric patients.

Acquisition protocols

MRI primarily is performed by using SE and GE techniques. As noted earlier, flowing blood creates a signal void on SE images. However, the complex pathway of blood flow through the heart results in disordered flow patterns. Focal areas of slow flow create signal, and the blood pool does not appear homogeneous. Depending on the imaging subject, the degree of inhomogeneity can degrade image quality and limit interpretation of the findings. To counteract this affect, two 180° presaturation pulses are used before the SE acquisition is initiated. This results in images on which the blood is truly black termed double inversion recovery (DIR) images. This technique provides superior delineation of the endocardial margin and vessel walls. Significantly disordered flow on DIR images suggests an abnormality, such as a septal defect or valvular disease. In this instance, a direct examination of the area is indicated. Because DIR is less sensitive to disturbed flow than other techniques, a GE sequence may be required for that evaluation. Any suspected flow abnormality on DIR images should be confirmed with GE images

DIR acquisition is ECG triggered with a recovery time (TR) equal to the R-R interval. Common imaging settings include the following:

- TE of 30.5 milliseconds
- Matrix size of 256 x 256
- FOV of 34 x 25 cm in adults, 24 x 18 cm in children, and 16 x 12 cm in neonates
- Number of excitations (NEX) of 1 in adults and children who can hold their breath, 2 in those who cannot, and 4 in neonates
- Section thickness of 8 mm with a gap of 0 mm in adults, usually 5 mm in children, and 3-5 mm in neonates

GE sequence is ECG triggered with a TR/TE of 9.4/5.1 and a matrix size of 256 x 128-256. The flip angle (FA) varies, but it usually is 15-30°. The section thickness, FOV, and NEX are adjusted for the patient's age, with the same considerations as those described for DIR. Vascular structures and connections sometimes are imaged with the white-blood technique to improve their conspicuity.

Cine GE sequences are used for functional assessment. Cine imaging involves multiple acquisitions or cardiac phases throughout a cardiac cycle in each imaging section. The result is a collection of images in each section that depict differing degrees of ventricular contraction. The images in each section can be stacked and displayed in a loop to permit visualization of the cardiac chamber and valvular motion. Cine GE settings are as follows:

- TR/TE of 9.4/5.1
- Matrix of 256 x 128
- FOV of 34 x 25 cm
- Section thickness of 8 mm
- NEX of 0.75
- FA of 15-30°

- Twenty cardiac phases: At least 16 image phases per cardiac cycle are needed to reliably depict end diastole (ED) and end systole (ES) for quantitative analysis.

When wall thickening and wall motion are assessed, significantly greater accuracy can be achieved by using a serial motion assessment with reference tracking (SMART) technique to account for long-axis and rotational movements.

Segmented k-space data acquisition is used widely. A specified number of phase-encoding steps are applied in each imaging phase per cardiac cycle to enable the efficient use of data collection time in multiple phases and sections. The temporal resolution depends on many factors and is discussed in the following section.

An important point to remember is that the FA affects contrast. Small variations in the FA can significantly change the brightness of the blood pool, the distinction of the myocardial wall, the appearance of

turbulent flow, and the imaging time. With the author's system, the acquisition time for an 8-mm-thick section increases from 5 seconds with a FA of 10 to approximately 16 seconds with a FA of 40°.

Sensitivity to blood-flow disturbances increases as the acquisition time increases. The loss of spin-phase coherence creates a blood-pool signal void. In addition, the brightness of flowing blood is linearly related to the FA. These 2 phenomena result in increased blood-pool contrast with increased FAs. With an FA greater than 30°, these factors can hamper image interpretation. The appearance of disordered flow can obscure the endocardial margin. Significant flow disturbances may lead to inaccurate qualitative and quantitative assessment, rendering the study useless.

With GE imaging, the contrast resolution of the soft tissue improves as the FA increases. With a FA smaller than 15°, the dominance of the proton density, along with the previously mentioned phenomena, make discrimination of the endocardial and epicardial margins difficult. Myocardial distinction is improved with FAs as large as 30° before increased scatter from effects of the blood pool cause interference.

Temporal resolution and imaging time

Functional evaluation of the cardiac chambers and internal structures requires good time resolution. Temporal resolution is a function of the TR, the heart rate, and the number of phase-encoding steps and cardiac-imaging phases. The number of phase-encoding steps per segment depends on the matrix and number of segments. In the standard protocol at the author's institution, a matrix with 128 phase-

encoding steps and 20 phases per cardiac cycle results in a k-space segment that contains 6 phase-encoding steps.

The temporal resolution is equal to the TR multiplied by the number of phase-encoding steps, or 9.4 milliseconds \times 128, which equals 1203 milliseconds per

imaging phase. However, the effective temporal resolution is equal to the TR multiplied by the number of phase-encoding steps per k-space segment, or 9.4 milliseconds \times 6, which equals 56.4 milliseconds.

This time is within the accepted temporal resolution needed to stop cardiac motion of 70-80 milliseconds, as reported in the literature.

One difficulty in cardiac imaging is related to the heart rate. The TR is gated to the R-R interval for cardiac-triggered imaging. For standard SE techniques, 1 phase-encoding step is acquired for each section per cardiac cycle. A series with a 256 x 256 matrix requires 256 heartbeats to complete. In a patient with a heart rate of 60 bpm and an R-R interval of 1000 milliseconds, image acquisition requires 4 minutes 16 seconds. With the use of cardiac magnets, SE sequences are faster. With the magnet at the author's institution, the DIR has an echo train length of 32, and an image is acquired in 8 heartbeats, or 8 seconds at a heart rate of 60 bpm. With this imaging time, 2 sections can be imaged per breath hold. Problems arise if the number of acquisitions must be increased, because the imaging time is lengthened, or if respiratory status is compromised, because the breath-holding time is decreased. In comparison, a heart rate of 100 bpm with an R-R interval of 600 milliseconds enables quicker imaging. In such conditions, an image can be acquired in less than 5 seconds, and more sections can be imaged per breath hold; thus, the examination is shortened. Conversely, breath-holding time is shorter, resulting in greater compliance.

Imaging protocols

The imaging protocol may include the following steps:

- Position the patient supine, with his or her feet first on the table.
- Place the ECG electrodes and leads.
- Position the cardiac radiofrequency coil over the anterior and posterior aspects of the chest. In children weighing less than 50 lb, the head coil is usually used.
- Begin imaging with a sagittal single-shot fast spin-echo (SS-FSE) sequence (TR/TE, 18,738/28.4; matrix, 256 x 128; FOV, 40 x 40; NEX 0.5; section thickness, 8 mm; spacing, 2 mm) with breath holding, if possible.
- Check the adequacy of cardiac coil placement to ensure coverage of the heart, and perform adjustments as necessary.

The sagittal series is used to configure a breath-hold coronal SS-FSE acquisition through the heart and great vessels

Ventricular analysis requires a true short-axis plane. This author's approach begins with the selection of an image from the coronal series that depicts both the cardiac apex and proximal ascending aorta. A few breath-hold GE or DIR sections are prescribed along an oblique axial plane parallel to a line from the cardiac apex through the middle of the aortic root. The result is a quasi 4-chamber view on which the interventricular septum can be localized. Cine imaging is performed from the apex to the base of the heart in a left parasagittal plane perpendicular to the interventricular septum. This provides short-axis images with excellent depiction of the aortic and pulmonary outflow tracts for ventricular analysis. The aortic and pulmonary valves can be evaluated in this fashion; image interpretation is limited with 8-mm sections but better with contiguous 5-mm sections.

Evaluation of the atrioventricular (AV) valves requires modification of the imaging plane. Breath-hold cine sections are prescribed along a doubly oblique plane that is parallel to a line connecting the cardiac apex and middle of the mitral valve and perpendicular to the septum. The result is a standard 4-chamber view. Cine GE imaging in this plane enables dynamic assessment of the tricuspid and mitral valves. Parasagittal images parallel to the AV junction can be used to measure the valve orifice area. Images obtained in a plane perpendicular to the long axis of the 4-chamber view are true short-axis views.

VENTRICULAR ANALYSIS

Many methods of ventricular volume calculation are used. Rehr et al, as well as Pearlman et al, found excellent correlation between findings at volumetric analysis with MRI and findings with ventricular casts (0.99 correlation, 4.9 mL standard error). Accuracy increases with the inclusion of long-axis measurements. Three-dimensional volumetric calculations are well correlated with ventriculographic findings and have low interstudy variability (<5%) compared with ventriculographic and echocardiographic results. The first step in

the calculation of ventricular volume is the selection of representative ED and ES cardiac-phase images. According to Semelka et al, either phase images that depict the largest and smallest ventricular volumes or the phase images obtained immediately before mitral valve closure (ie, ED) and opening (ie, ES) are chosen. Next, the right ventricle (RV) and left ventricle (LV) are traced along the endocardial margin on each section obtained in the selected ED and ES phases from the cardiac apex to the section just prior to one that depicts the mitral and tricuspid valves. The tracings should exclude trabeculations but may include papillary muscles inside the endocardial margin, as long as the procedure is performed in systole and diastole

Calculations can be made as follows:

- ED and ES volumes for each section are totaled to yield the RV and LV end-diastolic volume (EDV) and end-systolic volume (ESV).
- The stroke volume (SV) equals the EDV minus the ESV, or $SV = EDV - ESV$.
- The ejection fraction (EF) equals the SV divided by the EDV times 100, or $EF = (SV / EDV) \times 100$ to give a value reported as a percent.
- Cardiac output equals SV multiplied by the heart rate.

For myocardial mass assessment, the RV and LV epicardial borders are traced in ED. The interventricular septum is assigned to the LV and excluded from the RV tracing of the myocardial mass (see The volumes of all sections are added, and the corresponding EDV is subtracted to determine the myocardial volume. This result is then multiplied by the specific gravity of the myocardium (ie, 1.05 g/mL) to calculate the mass. This measurement is useful in the assessment of hypertrophy and to follow up the ventricular response to therapy.

The AV and ventriculoarterial valves also can be assessed with cine GE sequences. Valvular stenosis or regurgitation produces turbulent jets of signal void in the appropriate directions. Regarding the AV valves, regurgitation is graded according to echocardiographic criteria and is related to the distance to which the jet extends into the atrium. Grades of valvular stenosis are calculated more reproducibly. The

valve orifice areas can be measured and graded according to current standards

Table 1: Ventricular and Valvular Parameters*

Parameter	Men	Women
LV EDV (mL)	77-195	52-141
RV EDV (mL)	88-227	58-154
LV ESV (mL)	19-72	13-51
RV ESV (mL)	23-103	12-68
LV EF (%)	56-78	56-78
RV EF (%)	47-74	47-80
LV SV (mL)	51-133	33-97
RV SV (mL)	52-138	35-98
Cardiac output (L/min)	2.82-8.82	2.65-5.98
LV mass (g)	118-238	75-175
RV mass (g)	30-70	24-55
Septal mass (g)	40-82	26-58

*From Lorenz, 1999.

Typical areas for the aortic and mitral valves are 2.5-3.5 and 4-6 cm², respectively; areas of less than 0.8 and less than 1 cm², respectively, indicate severe stenosis. The values for aortic and mitral valve area apply to both males and females.

With cardiac MRI imaging, key points include the following:

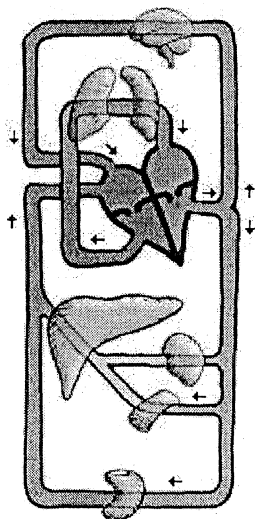
- Cardiac MRI is an accurate method for the evaluation of structure and function. Although it already is the procedure of choice in many clinical settings, its application continues to expand. The driving force, most likely, is the continuing shift toward diagnosis with noninvasive procedures.
- A complete understanding of the clinical question is needed for proper examination planning.
- Cardiac MRI findings have an excellent correlation with results with other means of functional assessment, with improved

interstudy variability. Calculations are easier to obtain and have fewer limitations compared with calculations with other modalities. In addition, MRI is changing the evaluation of coronary artery disease.

- Cine GE and contrast-enhanced imaging can be used to evaluate MI and predict functional recovery.
- Results of dynamic perfusion imaging at rest and with stress are promising in the evaluation of coronary artery disease.
- Space-time maps summarize the signal intensity changes in the myocardium over time, during transit of the bolus of contrast agent. These show impaired blood delivery as a delayed equitime, or the time to accumulate the LV signal, without a need to induce ischemia.
- Ongoing research and technologic advancements are sure to make MRI the leading cardiac imaging technique.

Planning Cardiac MRI

Anatomy

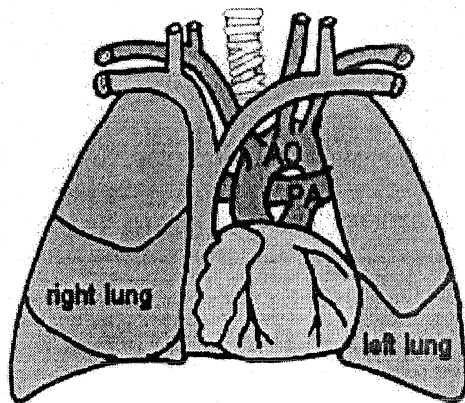


Circulation

Blood flows through the entire body. There are two circles. One circle brings the blood from the heart via the lungs back to the heart, the other circle brings the blood to all the other organs and back to the heart again. The main function of the heart is to pump the blood around. Under normal conditions the amount of blood pumped out each heartbeat is roughly 100 ml, for each circle. That equals 6 l every minute.

Position

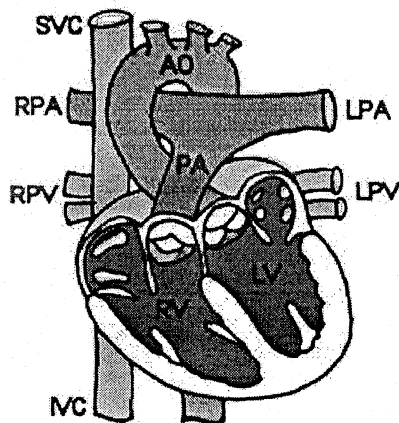
The heart is located in the chest, in between both lungs. Slightly rotated, the apex points towards the left side. On the top two large arteries are connected. The pulmonary artery brings the blood to both lungs, the aorta to the rest of the body.



Two sides

The heart is divided into two sides. The left heart receives fresh oxygenated blood from the left and right pulmonary veins (LPV and RPV) in the left atrium (LA). Via the Mitral valve the blood flows into the left ventricle (LV). When the heart contracts the Mitral valve closes and the blood is pushed via the aortic valve into the aorta (AO). When the

blood returns from the body it flows via the superior and inferior vena cava (SVC and IVC) into the right atrium (RA). Via the tricuspid valve the blood enters the right ventricle (RV). As soon as the heart contracts the tricuspid valve closes and blood flows via the pulmonary valve into the pulmonary artery (PA). At the bifurcation the blood flows via the left and right pulmonary artery (LPA and RPA) to the lungs.



Four chambers

Long axis view

Starting with a transverse plane through the centre of the heart, a long axis view (LA or LAX) can be defined. Position the slice through apex and centre of the Mitral valve as indicated in Fig 1A. This LA view shows the left ventricle over its full length from apex to base, including the Mitral valve and the left atrium. Note the pulmonary veins that come in from superior into the left atrium.

This view is sometimes also referred to as vertical long axis (VLA).

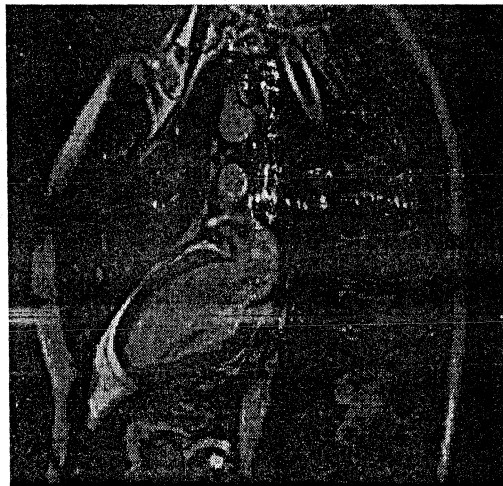
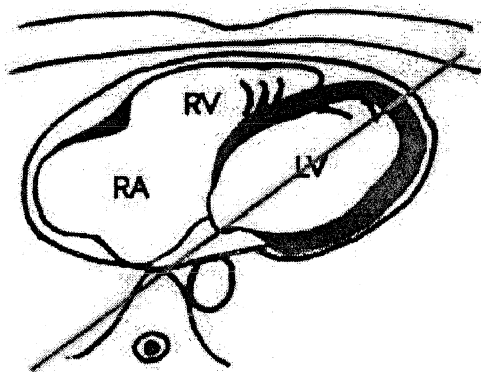


Fig 1A: Transverse view.



Fig 1B: Resulting LA view.

Semi four-chamber view

Some people like to define all cardiac views using two orthogonal images. This ensures that the resulting plane is nice orthogonal aligned in both directions. If however the initial transverse plane already clearly shows the apex and the Mitral valve I usually skip this step and continue straight with the short axis view. refer to this view as "semi" four-chamber because usually the AP angulation is not precisely right. In this example it's more a semi left ventricular outflow tract. I definitely require a short axis view first to ensure a true four-chamber view.

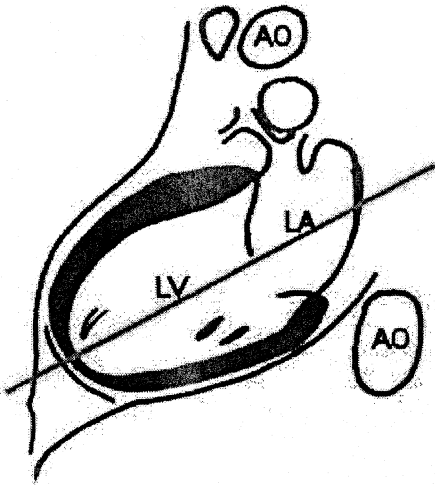


Fig 2A: LA view.

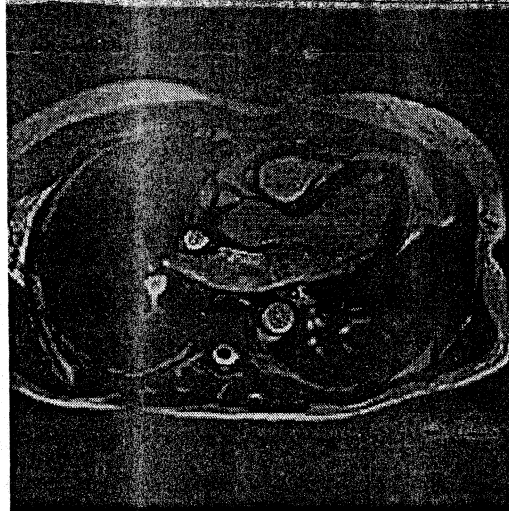


Fig 2B: Resulting (semi) 4CH view

Short axis view

Now that we have two orthogonal views of the left ventricle we're ready to define another cross-section of the heart: the short axis view (SA or SAX). There are several guidelines that can be followed to define a short axis view:

1. - Orthogonal to the line that connects the centre of the Mitral valve with the apex.
2. - Orthogonal to the septum. (Not seen on a LA view, but visible on semi 4CH)
3. - Parallel with the Mitral valve.

It doesn't matter much which rule to follow here. The accuracy of the volumetric analysis depends mostly on the decision which slice to include as the most basal one. For that reason I always choose to go parallel with the Mitral valve. Sometimes the previously described (semi) four-chamber view (Fig 4A) is used to verify the angulation in another orthogonal direction

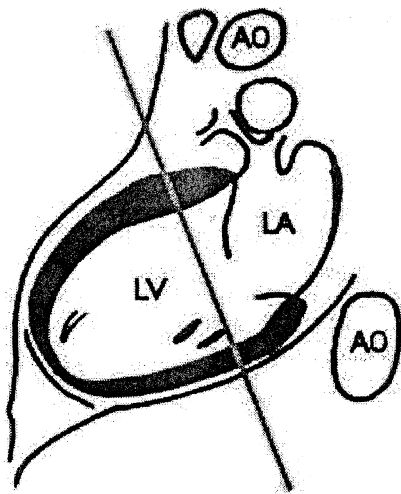


Fig 3A: LA view.

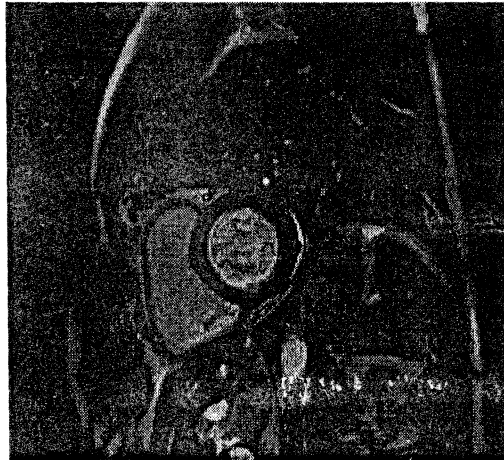

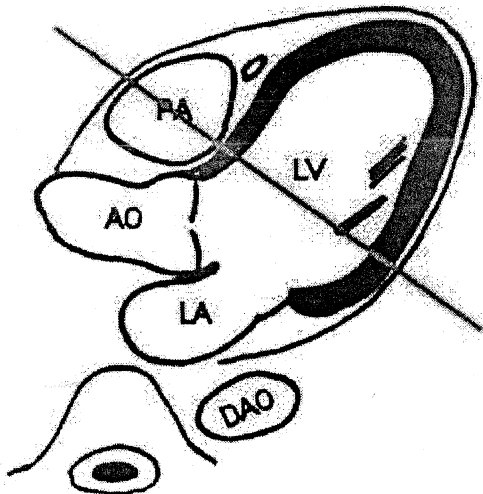

 Fig 3B: Resulting SA view.


Fig 3C: Semi 4CH view.

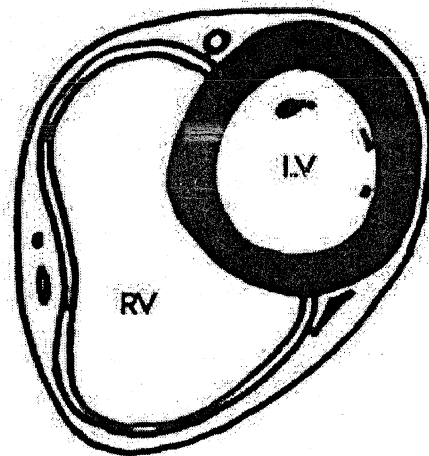


Fig 3D: Resulting SA view.

Four-chamber view

The SA view is used to define the four-chamber view (4CH) that shows us both ventricles and atria in a single plane. This view is sometimes also referred to as horizontal long axis view (HLA). Define a line through the centre of the left ventricle and the lower corner ("apex") of the right ventricle. Many times it's very difficult to distinguish the right ventricle from the right atrium. That is because the right ventricular wall is so thin, and the tricuspid valve is not easily recognized.

It's a good idea to verify the planning of the four-chamber view using a long axis view. Just to ensure that I am cutting nice through the apex and the center of the Mitral valve. Keep in mind that the heart is moving with the respiration in feed-head direction. Therefore you must make sure that both views were acquired in the same respiration phases. I strongly encourage you to acquire all views in expiration phases only because the position of the heart tends to vary less compared to positions acquired in inspiration.

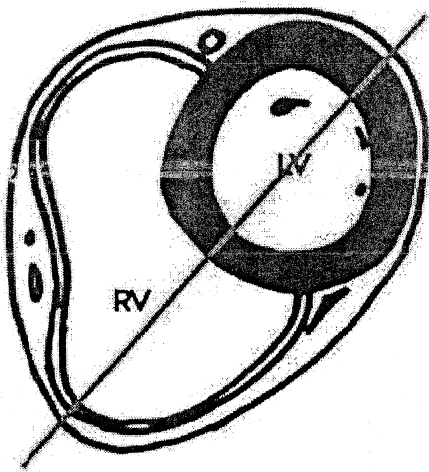


Fig 4A: SA view.



Fig 4B: Resulting 4CH view.

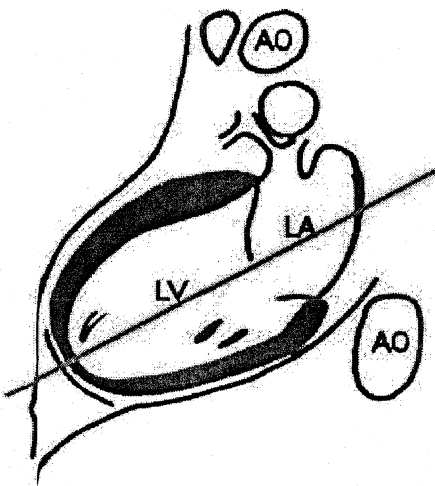


Fig 4C: LA view.

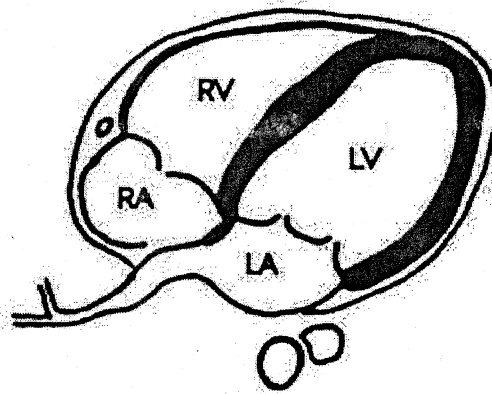


Fig 4D: Resulting 4CH view.

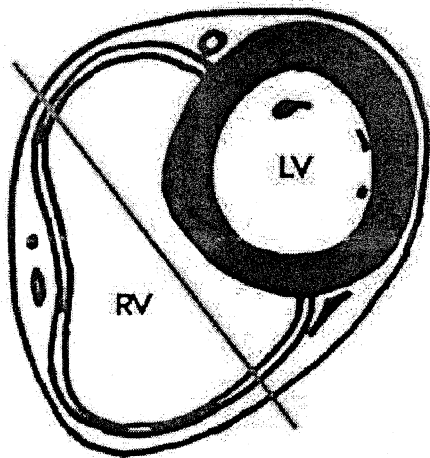


Fig 2S: SA view.

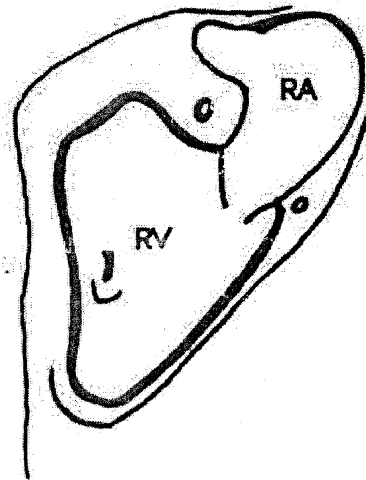


Fig 2D: Resulting R2CH view

Left ventricular outflow tract

There is some disagreement on the precise definition of the Left Ventricular Outflow Tract. Some sites refer to this view - that visualizes the left atrium, mitral valve, left ventricle, aortic valve and aorta - as the "three chamber view". This is a little bit strange because this view only shows two chambers: the left atrium and ventricle. Perhaps some do count the root of the aorta as third chamber. Nevertheless the idea of this view is to visualize nicely both the mitral valve and the aortic valve. The first landmark we use is the centre of the left ventricle. If more SA slices are available a basel slice could be selected that shows the root of the aorta. The LA view can be used to verify again that the slice cuts through the far end of the apex. Sometimes the plane is angled such that it follows the sagittal definition. The result is that those very steep angled views are presented as sagittal views which may lead to some confusion. Anterior is presented towards the left side of the image instead of towards the top as for transverse planes. Just "rotate & mirror" the images and you'll recognize them again as true LVOT views.

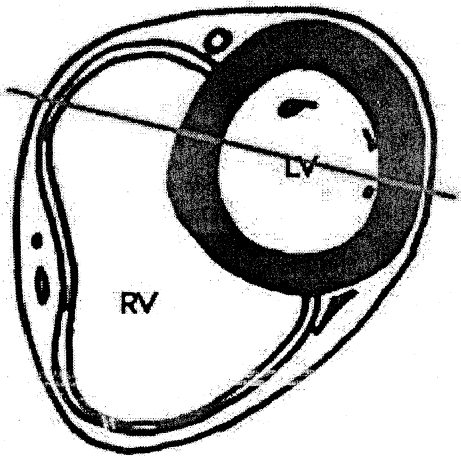


Fig 3A: SA view.

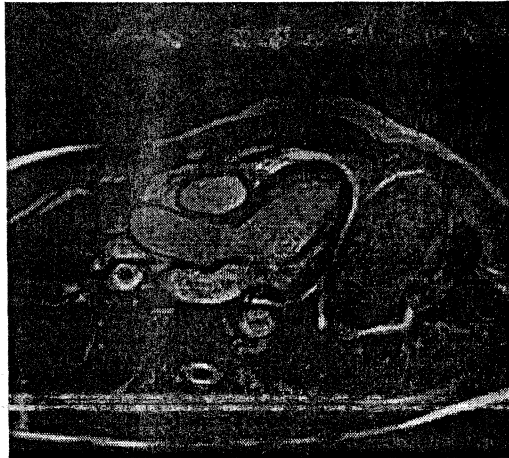


Fig 3B: Resulting LVOT view.

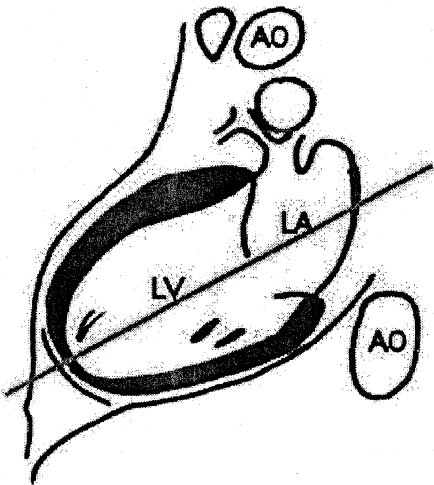


Fig 3C: LA view.

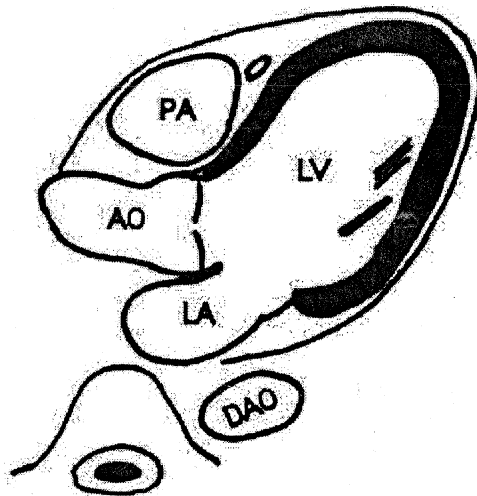
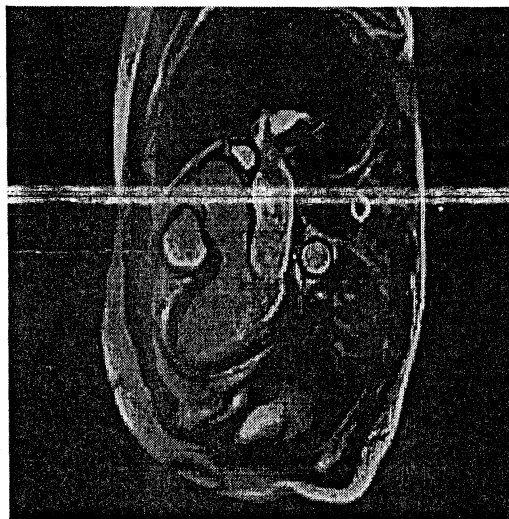


Fig 3D: Resulting LVOT view.



Inflow and outflow tracts

Fresh oxygenated blood comes from the pulmonary veins and flows into the left atrium. The P-wave initiates contraction of the atria. Both the tricuspid valve and mitral valve open and the blood is pushed further into the ventricles. When the R-peak is there the ventricles start to contract and the pressure inside the ventricle increase. Soon the tricuspid valve and mitral valve will close.

Left two chamber view

Very similar to the definition of our first view, the long axis view. The only difference is that this two chamber view is orthogonal to the four-chamber view. (The previously described LA view is perpendicular to the transverse view.) Just describe the line through the apex of the left ventricle and the centre of the mitral valve.

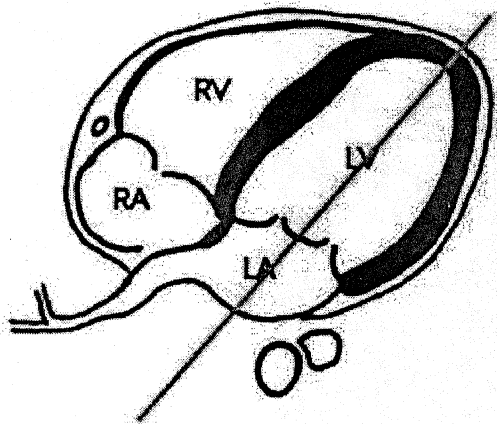


Fig 1A: 4CH view.



Fig 1B: Resulting L2CH view.

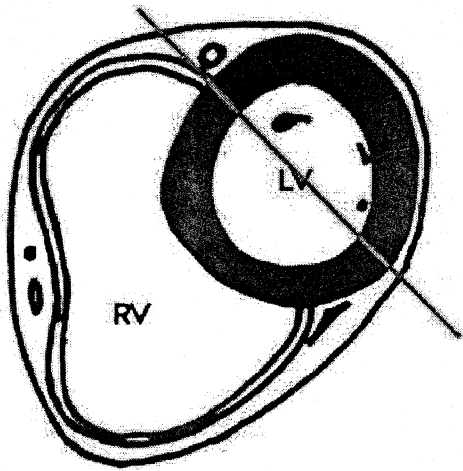


Fig 1C: SA view.

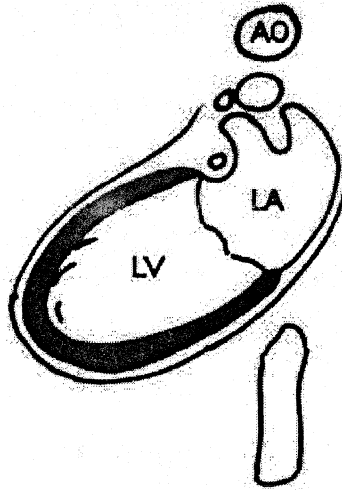


Fig 1D: Resulting L2CH view.

Right two chamber view

Now move the plane to the right side of the heart. Perhaps rotate it a little bit to get the plane parallel with the septum. Make sure to cut precisely through the centre of the tricuspid valve.

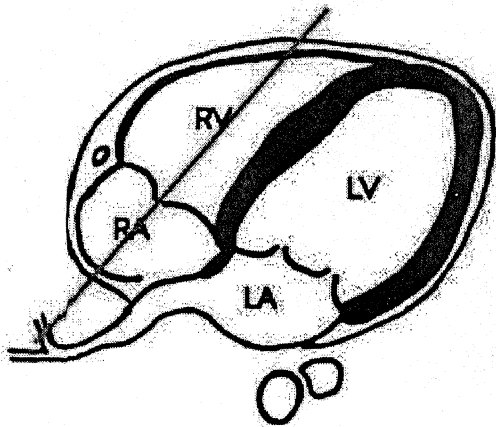
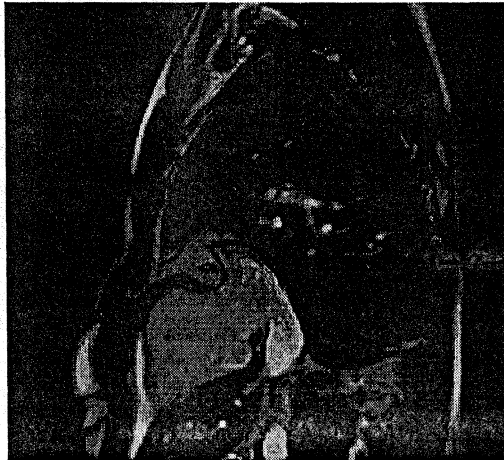




Fig 2A: 4CH view.


 Fig 2B: Resulting R2CH view.

 Fig 3F: Steep angled LVOT views might be presented as sagittal images...

Vertical left outflow tract

As said before there is some disagreement on what is meant with "left ventricular outflow tract". Therefore I prefer to the following view as **vertical** left outflow tract. Take the LVOT view as described before and define a line orthogonal to the aortic valve plane. Rotate it a little bit just to capture as much from the aorta as reasonable without loosing too much from the left ventricle ;-). And watch how nice you can see the aortic valve opening and closing...

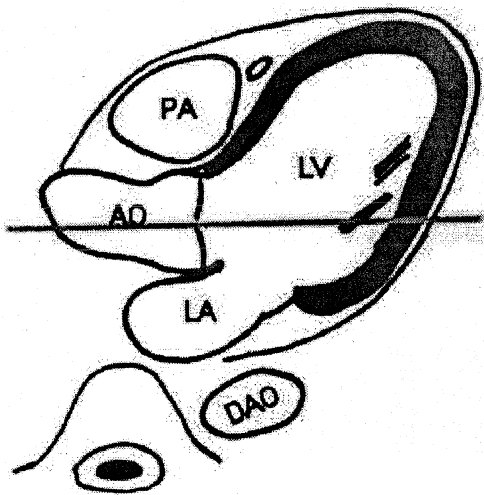



Fig 4A: LVOT view.



 Fig 4B: Resulting VLOT view

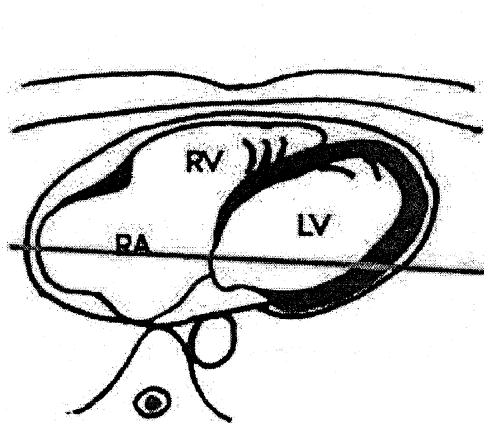


Fig 4A: Transverse view.

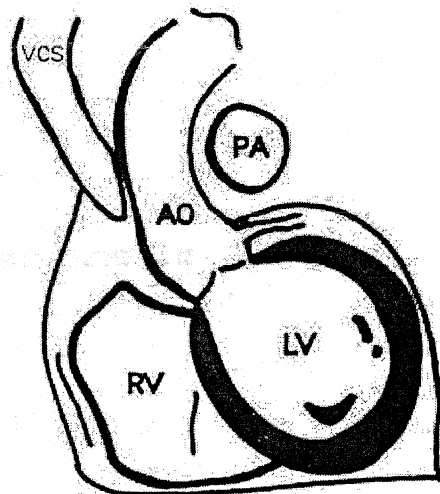


Fig 4B: Resulting VLOT view

Right ventricular outflow tract

The aim is to visualize the pulmonary valve, such that we can differentiate the left ventricle from pulmonary artery. The easiest way to define this plane is to start with just a sagittal plane. Browse through your transverse slices from a survey scan and look for the root of the pulmonary artery, it's not difficult to find. Place your slice through the center of the pulmonary artery.

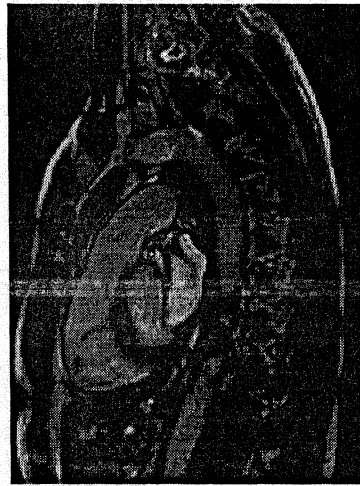
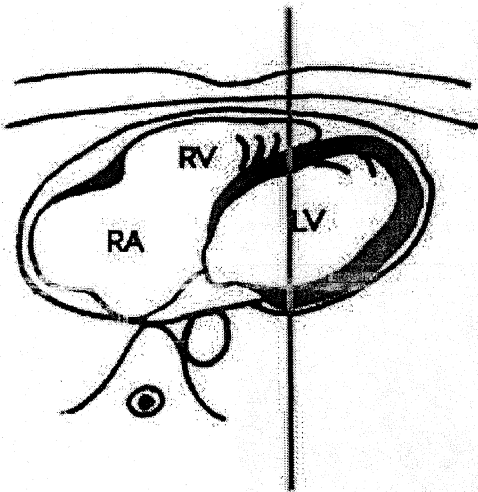


Fig 5A: Transverse view.



Fig 5B: Resulting RVOT view.

Aortic arch

If a "three-points-planscan" tool is available than that's the way to go! Just simply click three points: one in the **ascending aorta**, another one right in the **aortic arch**, and the last one in the **descending aorta**. If no such tools are available then select a straight transverse view acquired above the heart. One that shows both the ascending and descending aorta. Position the slice through both parts of the aorta. Select a vertical outflow tract and align the image plane along with the ascending aorta.

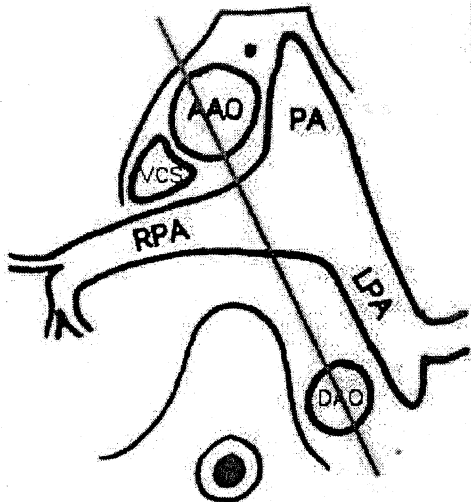


Fig 1A: 4CH view.

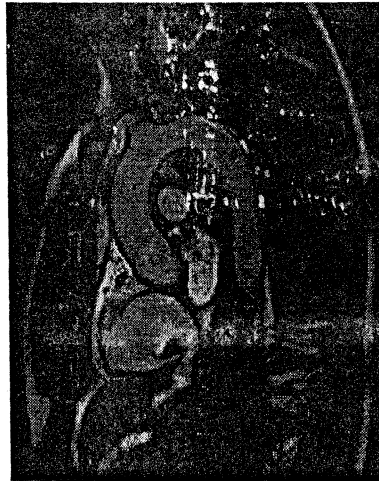

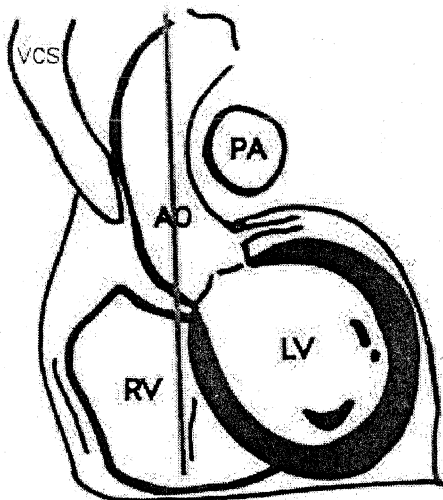

 Fig 1B: Resulting aortic arch.


Fig 1C: VLOT view.

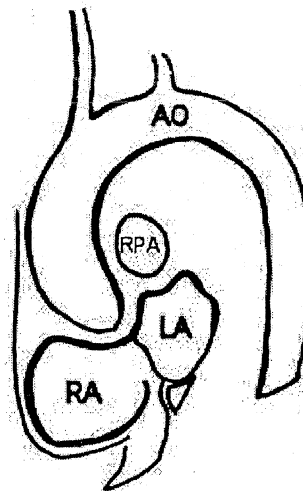


Fig 1D: Resulting aortic arch.

Pulmonary artery

It's exactly the same view as the RVOT: The aim is to visualize the pulmonary valve, such that we can differentiate the left ventricle from pulmonary artery. The easiest way to define this plane is to start with just a sagittal plane. Browse through your transverse slices from a survey scan and look for the root of the pulmonary artery, it's not difficult to find. Place your slice through the centre of the pulmonary artery.

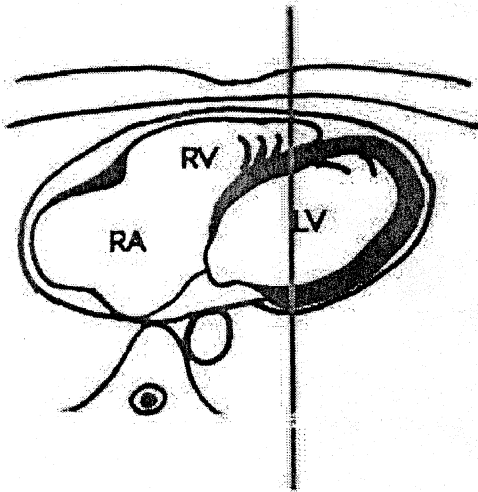


Fig 5A: Transverse view.

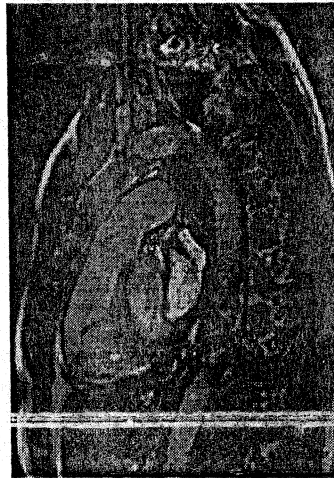

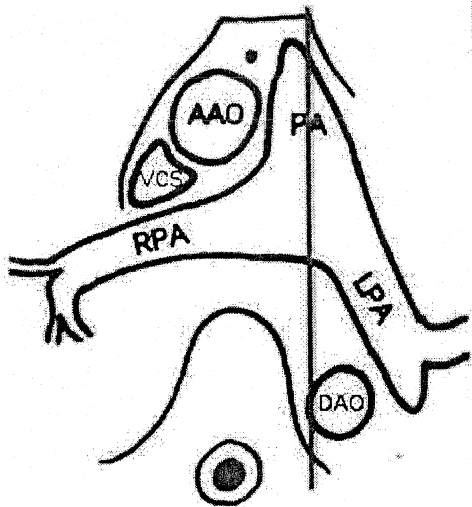

 Fig 5B: Resulting pulmonary artery.


Fig 5C: Transverse view.

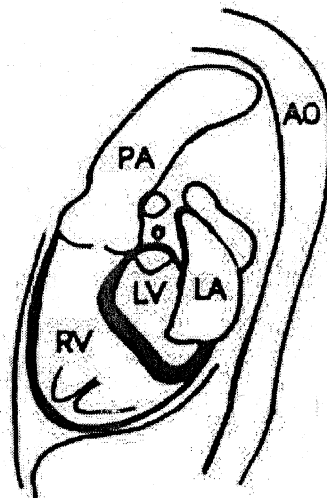


Fig 5D: Resulting pulmonary view.

Pulmonary bifurcation

Define a straight transverse plane right through the pulmonary artery as seen on a PA view (or RVOT ;-). I usually take a ride with InterActive scanning to tilt the plane such that I see both the right and the left pulmonary artery to their full extend

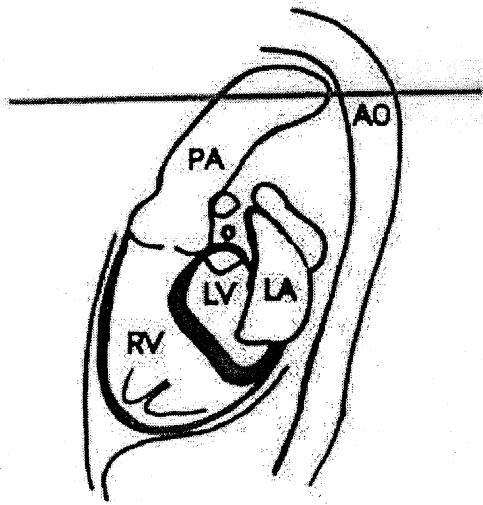


Fig 3A: PA view.

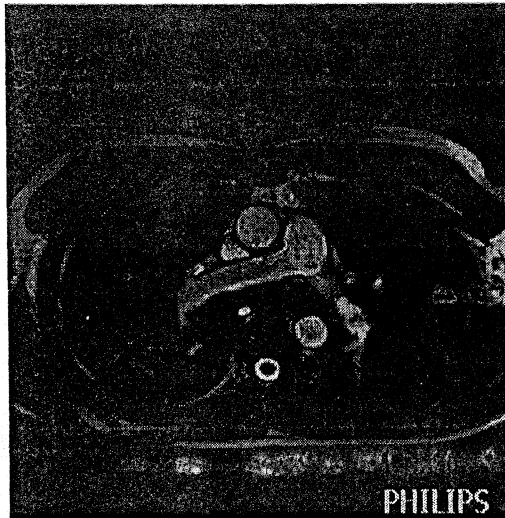

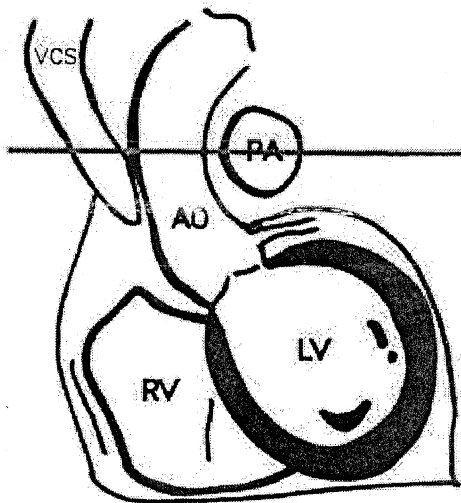

 Fig 3B: Resulting pulmonary bifurcation.


Fig 3C: LA view.

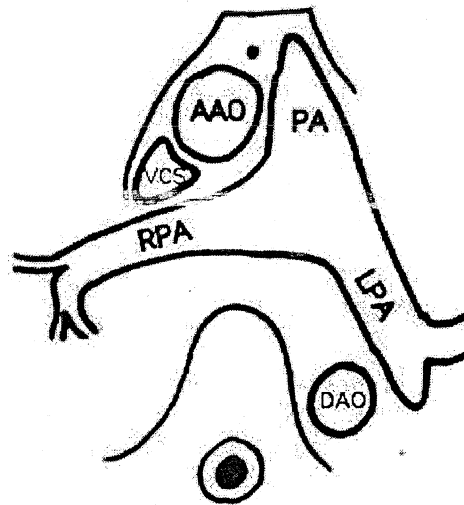


Fig 3D: Resulting pulmonary bifurcation.

Left pulmonary artery

This pulmonary bifurcation view can be used to very easily define the left pulmonary artery. Simply position your slice perpendicular to the bifurcation view, nicely aligned with the left pulmonary artery.

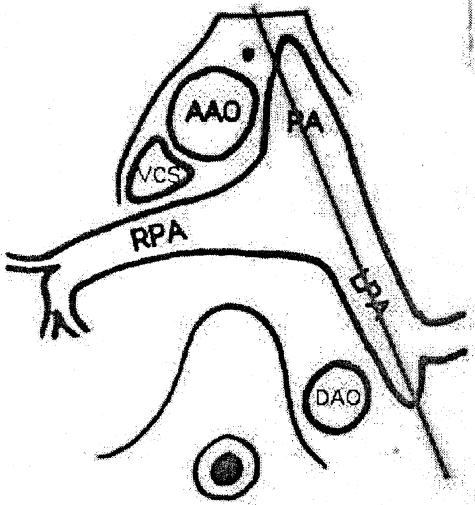


Fig 4A: PT view.

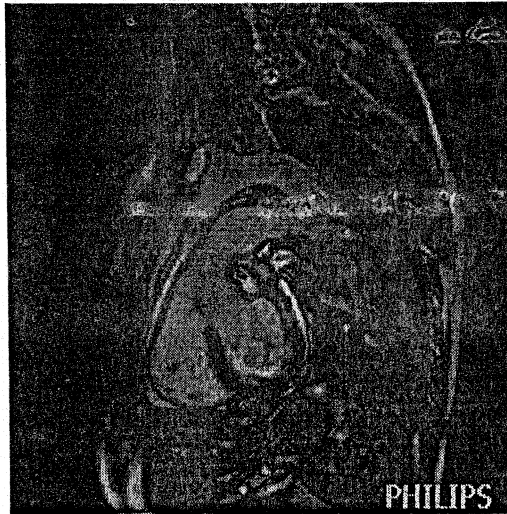

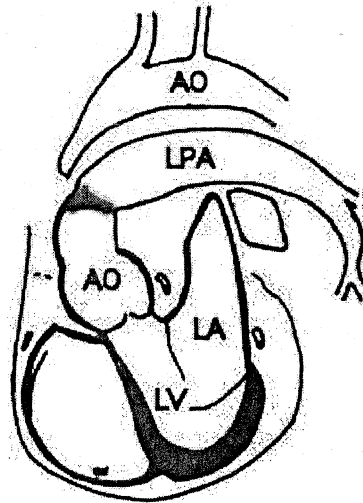

 Fig 4B: Resulting left pulmonary artery.


Fig 4D: Resulting left pulmonary artery.

Right pulmonary artery

The pulmonary bifurcation view can also be used to very easily define the right pulmonary artery. Just position your slice perpendicular to the bifurcation view, nicely aligned with the right pulmonary artery.

Note that the right pulmonary artery tends to slightly descend. That's why I use InterActive scanning to figure out the amount of tilting that's needed to visualize the pulmonary truncus to it's full extent.

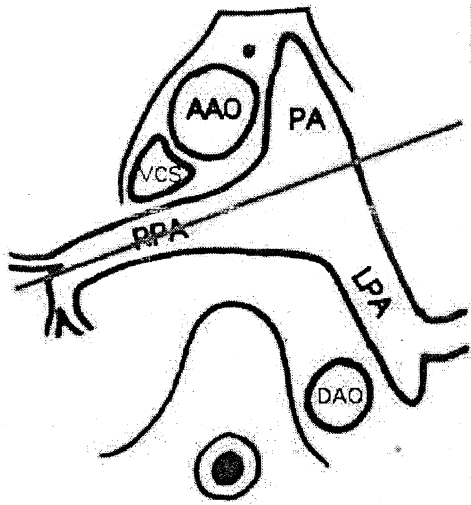


Fig 5A: PT view.


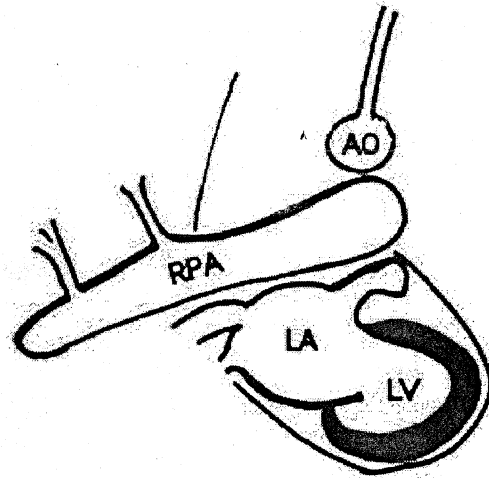

 Fig 5B: Resulting right pulmonary artery.


Fig 5D: Resulting right pulmonary

Flow through arteries

Quantification of blood flow is an interesting application. A nice experiment includes the comparison between the amount of blood pushed into the aorta each heart-beat and the amount of blood pumped into the pulmonary artery. This should be - more or less - the same. Another experiment is to add the stroke volumes for LPA and RPA together and compare this to the PA and AO. And another one is just to take a single slice cutting through ascending aorta, descending aorta and vena cava superior. The difference between ascending and descending aorta represents the blood supply to the head and arms, which should be equal to the amount of blood that returns via the vena cava superior...

For each flow measurement we have to understand the basic rules for flow quantification:

1. Define your scans perpendicular to the vessel as much as possible.
2. Stay away from areas with fast turbulent flow (stenoses, valves).
3. Adjust the VENC to the highest expected flow velocity.

Aortic flow

Use two orthogonal views that show the aorta (Aortic Arch and VLOT) and position the slice perpendicular to the aorta. Stay away from the aortic valve to avoid too much interference from turbulent flow. A more or less straight transverse orientation at the level of the right pulmonary artery is a good starting point as shown in Fig 1A. If flow is to be evaluated both for ascending and descending aorta, make sure to average your angulation such that both parts of the aorta are cut orthogonal.

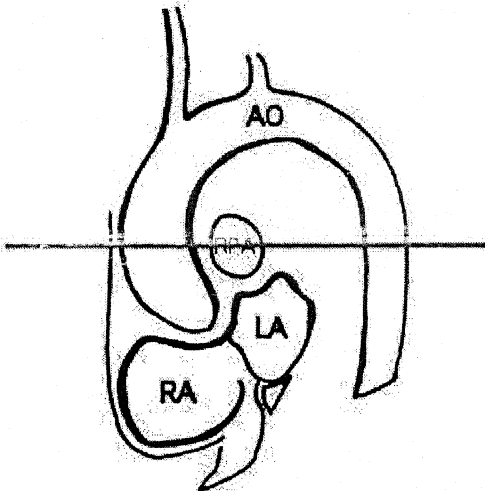


Fig 1A: Aortic arch.

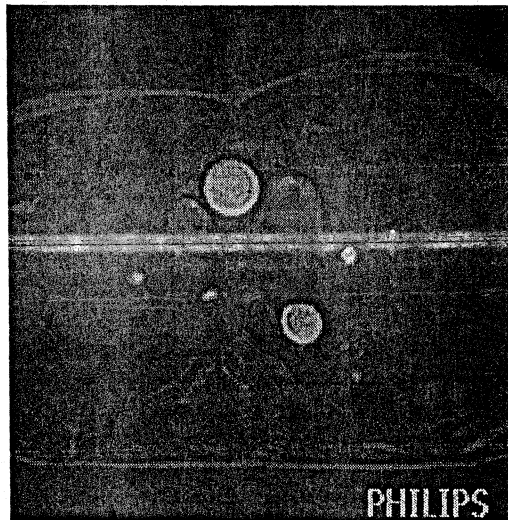


Fig 1B: Resulting aortic cross-section. (FFE-modulus)

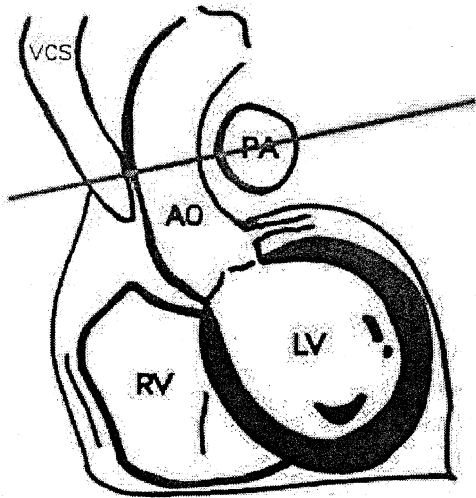


Fig 1C: VLOT view.

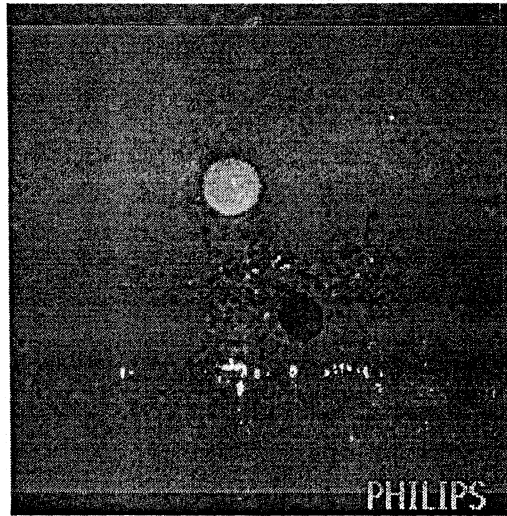

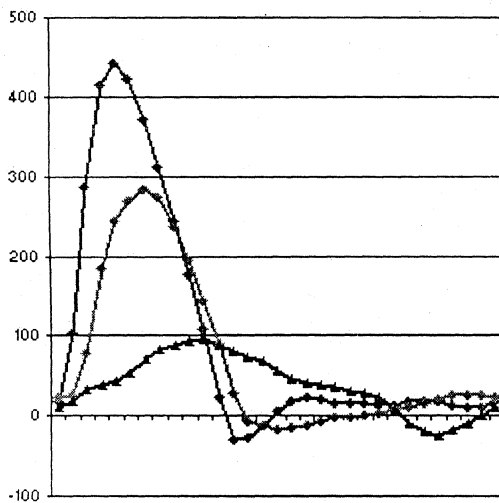

 Fig 1D: Resulting aortic cross-section. (PCA-phase)


Fig 1E: Flow rate (ml/s) Red: AAO, Green: DAO, Blue: SVC

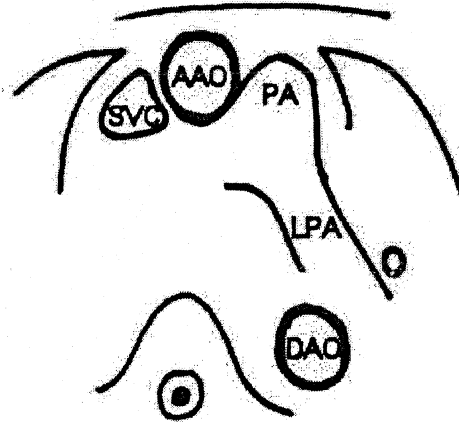


Fig 1F: Resulting aortic cross-section.

Pulmonary flow

Select the pulmonary artery view (or RVOT) to position your slice on. It sometimes is difficult to recognize the pulmonary valve, but try to stay a few cm away from it. Check on the pulmonary bifurcation view if you're not too close to the bifurcation.

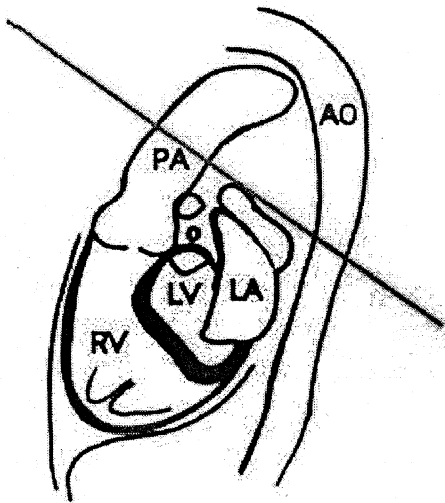


Fig 2A: Pulmonary view (RVOT).

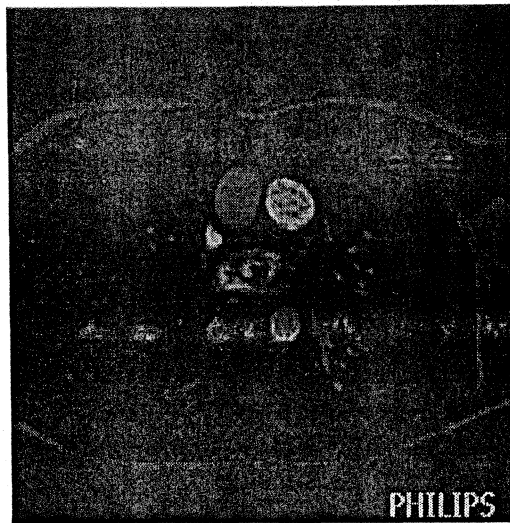


Fig 2B: Resulting pulmonary cross-section. (FFE-modulus)

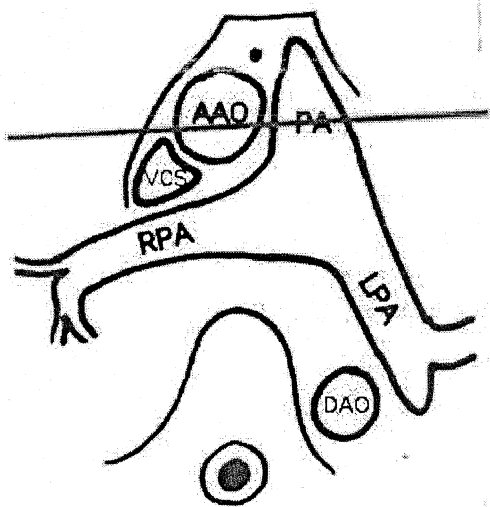


Fig 2S: Pulmonary bifurcation.

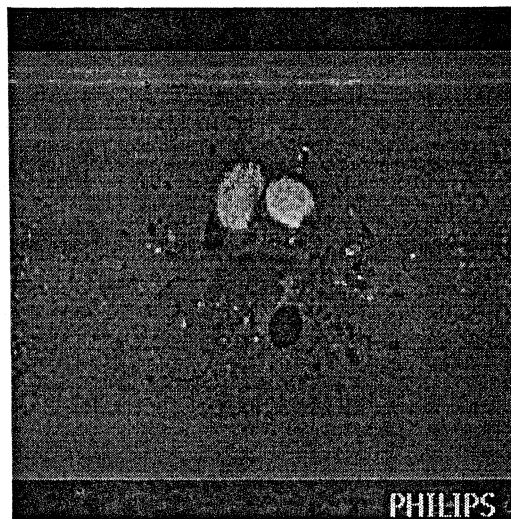


Fig 2D: Resulting pulmonary cross-section. (PCA-phase)

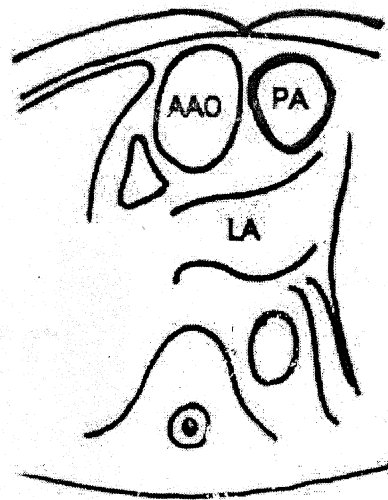
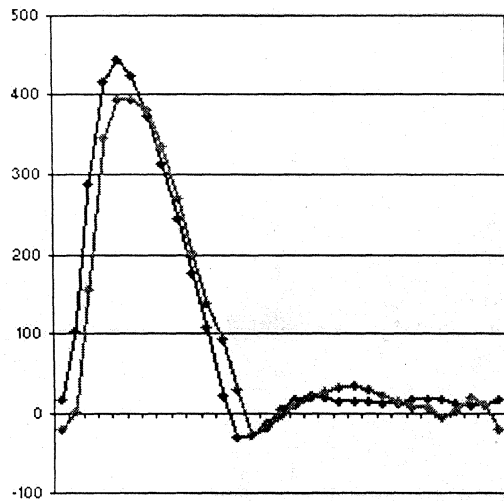


Fig 2E: Flow rate (ml/s) Red: AO, Fig 2F: Resulting pulmonary cross-section. Green: PA

Left pulmonary flow

Select both the bifurcation view and LPA, and position your slice orthogonal to the left pulmonary artery. Position your slice at a place not too close to the bifurcation to avoid any chance of turbulent flow.

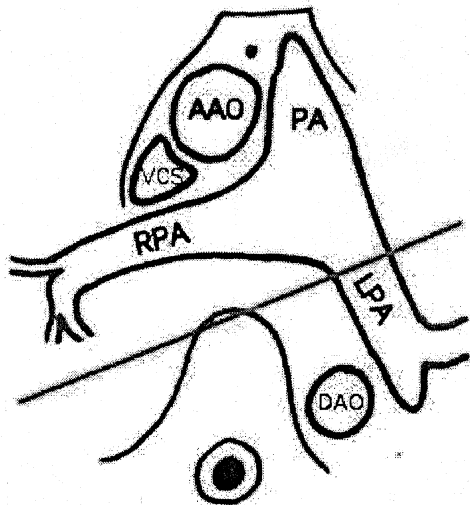


Fig 3A: Pulmonary bifurcation.

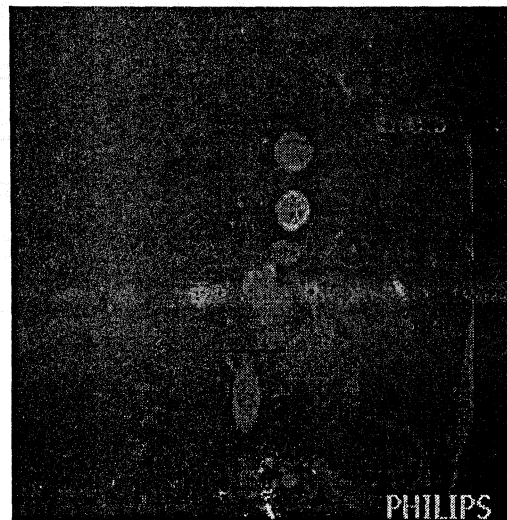


Fig 3B: Resulting LPA cross-section. (FFE-modulus)

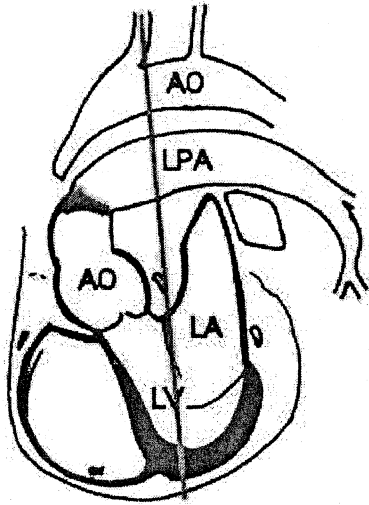


Fig 3C: LPA view.

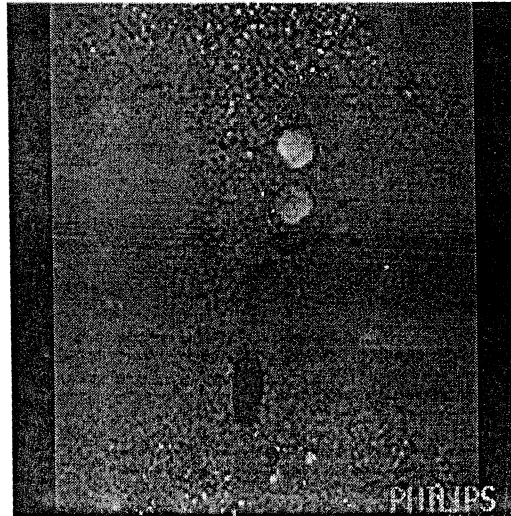


Fig 3D: Resulting LPA cross-section. (PCA-modulus)

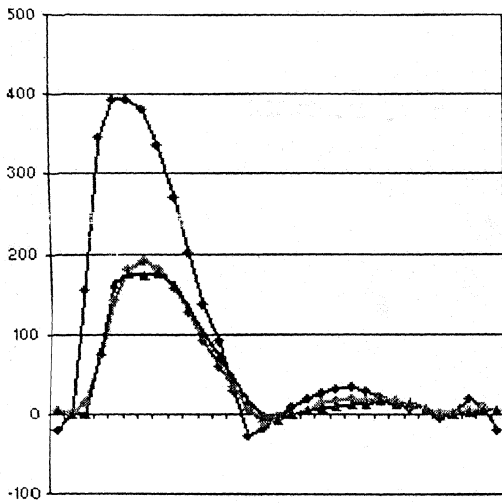


Fig 3E: Flow rate (ml/s) Red: PA, Green: LPA, Blue: RPA

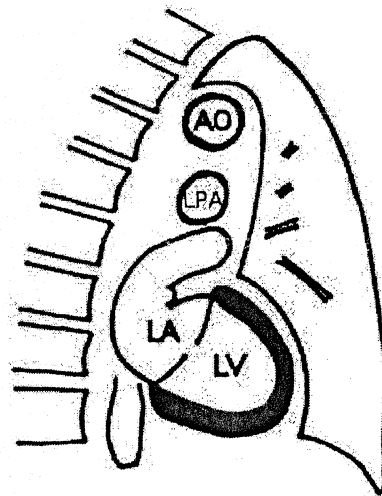


Fig 3F: Resulting LPA cross-section.

Right pulmonary flow

Very similar to the LPA. Now select both the bifurcation view and RPA, and position your slice orthogonal to the right pulmonary artery.

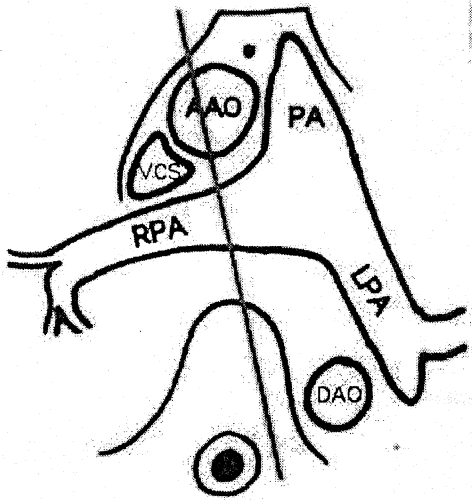


Fig 4A: Pulmonary bifurcation.

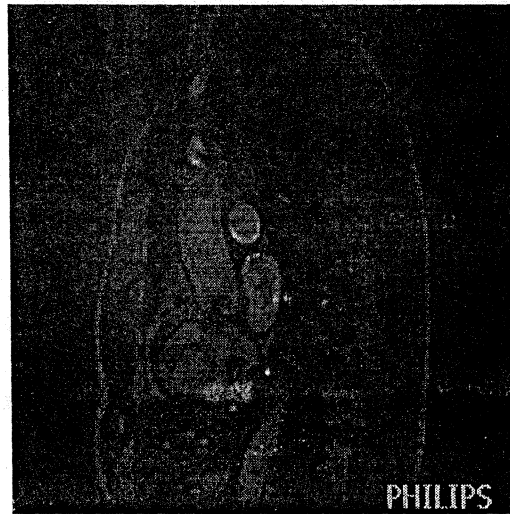


Fig 4B: Resulting RPA cross-section. (FFE-modulus)

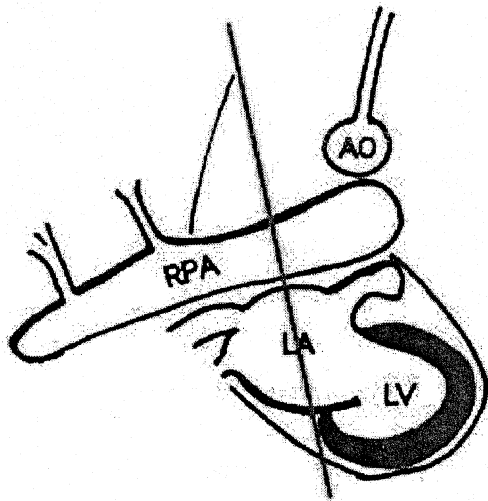


Fig 4C: RPA view.

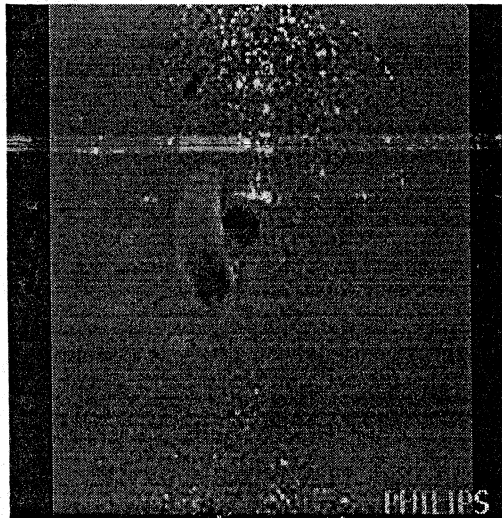


Fig 4D: Resulting RPA cross-section. (PCA-modulus)

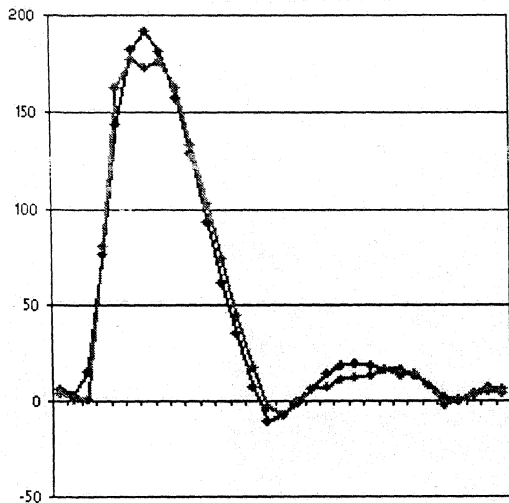


Fig 4E: Flow rate (ml/s) Red: LPA, Green: RPA

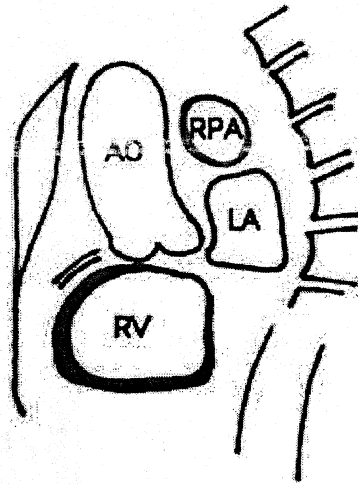


Fig 4F: Resulting RPA cross-section.

Flow through valves

Mitral valve

Perhaps the most difficult view of all. Take the four-chamber view and the long axis view (L2CH) that both show the mitral valve. The difficulty lies in the fact that both the mitral valve and the tricuspid valve experience some through-plane movement during the cardiac cycle. For that reason I try to select images that were acquired in early diastole, the moment that the mitral valve opens. On these images I position the slice parallel with the mitral valve plane.

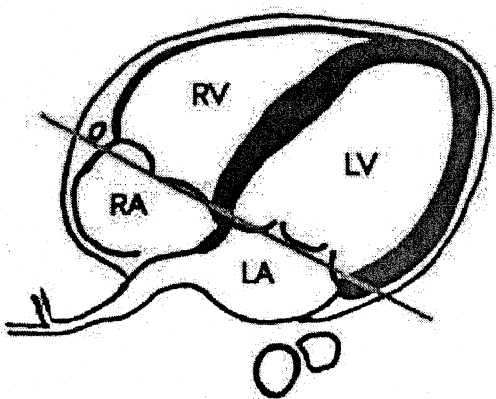


Fig 1A: Four-chamber view.

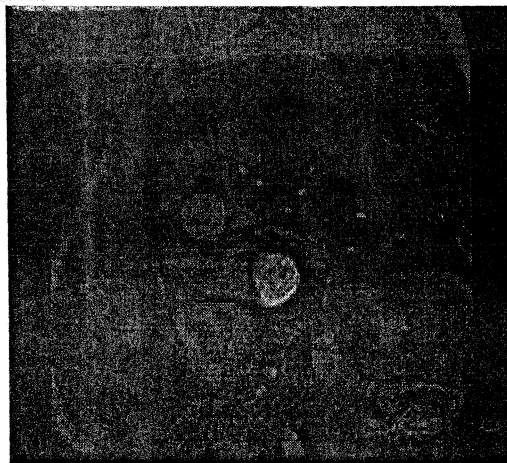


Fig 1B: Resulting mitral valve

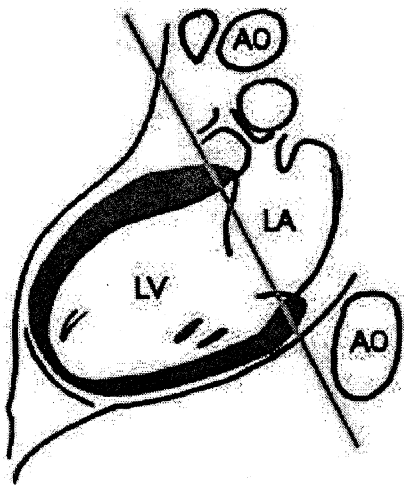


Fig 1C: Left 2 chamber view.

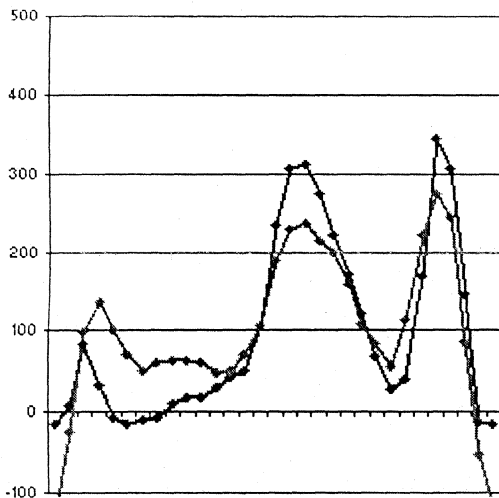


Fig 1E: Flow rate (ml/s) Red: Mitral, Green: Tricuspid

Tricuspid valve

Basically it's the same position as for the mitral valve. The only difference is that I adjust the angulation such that the plane is now parallel with the tricuspid valve. So select the four-chamber view and the right two chamber view with images acquired in early diastole. On these images I position the slice parallel with the tricuspid valve plane. valve plane.

view. (FFE-modulus)



Fig 1D: Resulting mitral valve view. (PCA-modulus)

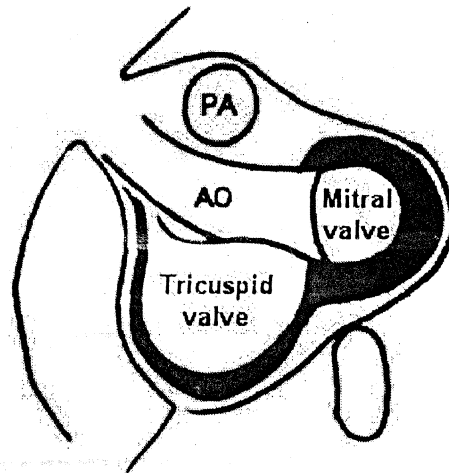


Fig 1F: Resulting mitral valve.

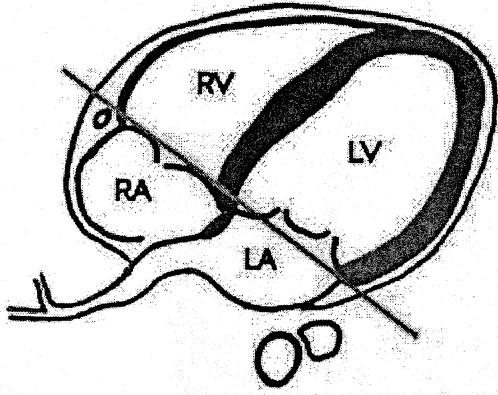


Fig 2A: Four-chamber view.

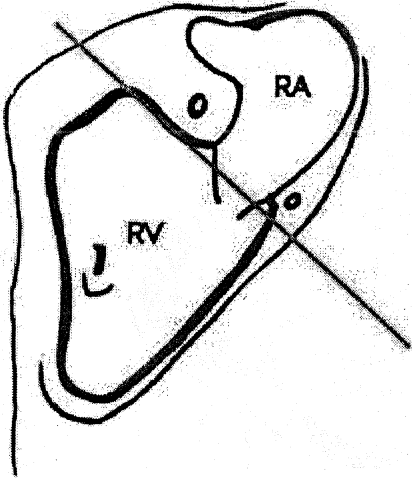


Fig 2C: Right 2 chamber view.

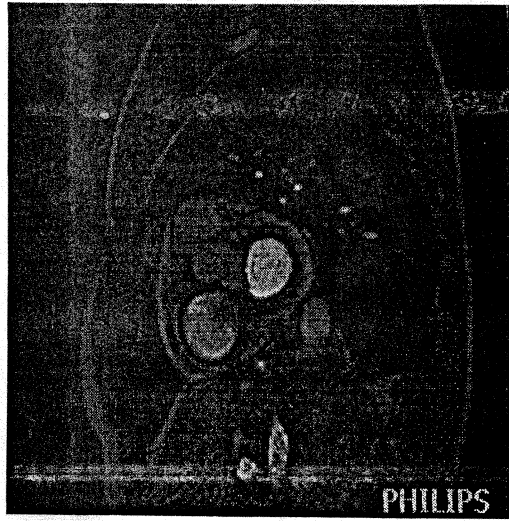


Fig 2B: Resulting tricuspid valve view. (FFE-modulus)

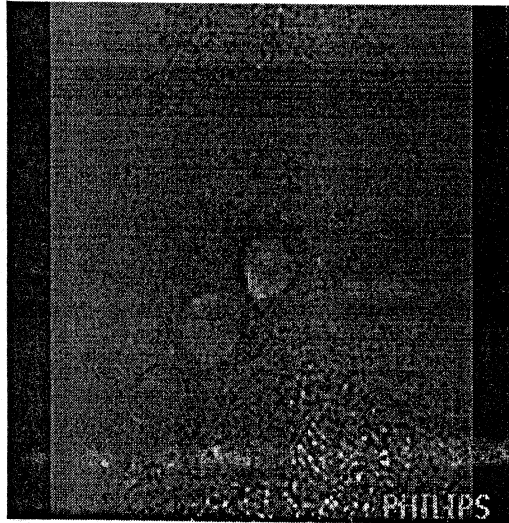


Fig 2D: Resulting tricuspid valve view. (PCA-modulus)

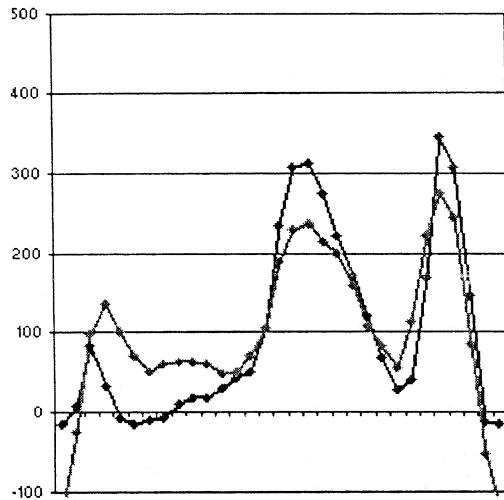


Fig 2E: Flow rate (ml/s) Red: Mitral, Green: Tricuspid

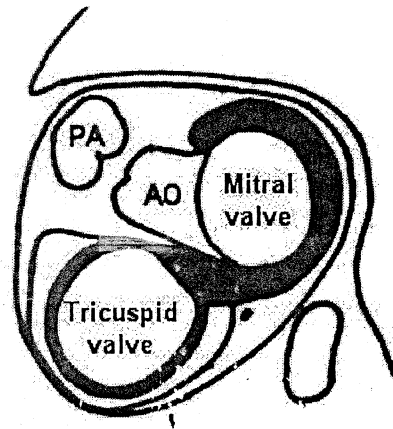


Fig 2F: Resulting tricuspid valve.

Myocardial tagging

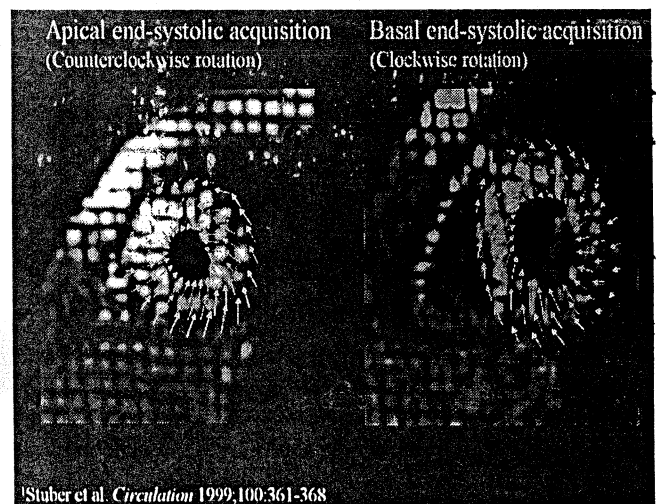
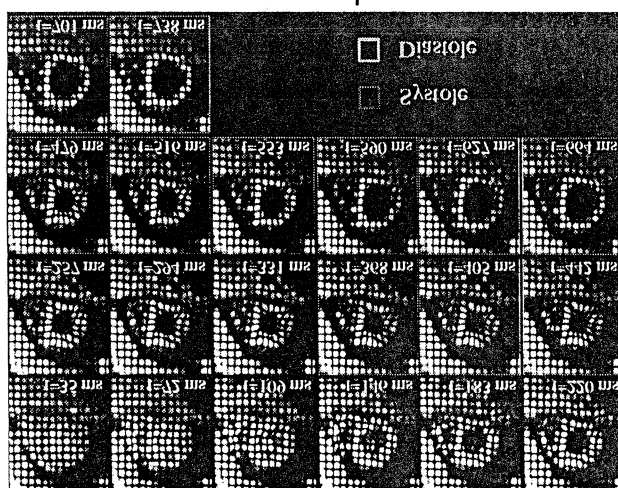
Introduced by Elias Zerhouni

SPAMM (spatial modulation of magnetization), line tagging, radial tagging, ring tagging are the most commonly used techniques

Physics

Ultra small saturation pulses are applied immediately after the ECG trigger. The RF excitation pulse tags specific parts of the myocardium by nulling the magnetization. The tagged tissue will produce markedly reduced signal intensity during certain phase of the saturation pulse. Tagging in short axis is then performed with a series of radial tagging through the center of myocardium.

It is possible to track the contraction of left ventricle & estimate the related wall stress in particular



Stuber et al. *Circulation* 1999;100:361-368

CARDIAC PERFUSION

TRUFISP-SR-FIRSTPASS-3 SLICE

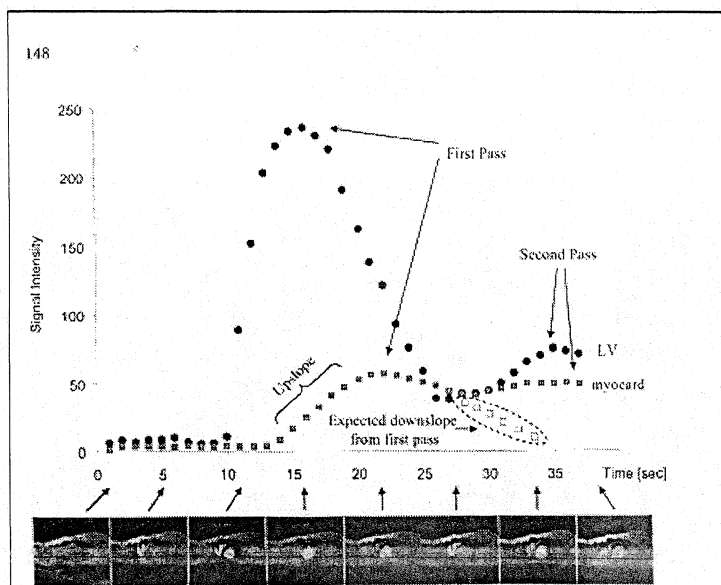
Myocardial perfusion of ischemic heart disease can be studied. Gadolinium based contrast agents are injected intravenously and first pass metabolism is studied. Fast t1 weighted MR techniques used

Fast gradient echo pulse sequences to be used. Magnetization prepared turbo field echo or echo planar or flash. Repetitive registration in minimum of 3 planes at every heartbeat to be acquired. Quantitative image analysis is done with first pass bolus parameters. Level of enhancement, time to peak, mean transit time all measured.

Delayed Enhancement

After contrast injection, plasma conc. reaches a maximum value and then rapidly declines thru diffusion into interstitial space & renal washout. In infarcted tissues, contrast resorption rate is reduced. In normal myocardium, at 15-30 min of contrast injection complete washout is noted

Signal intensity time curve

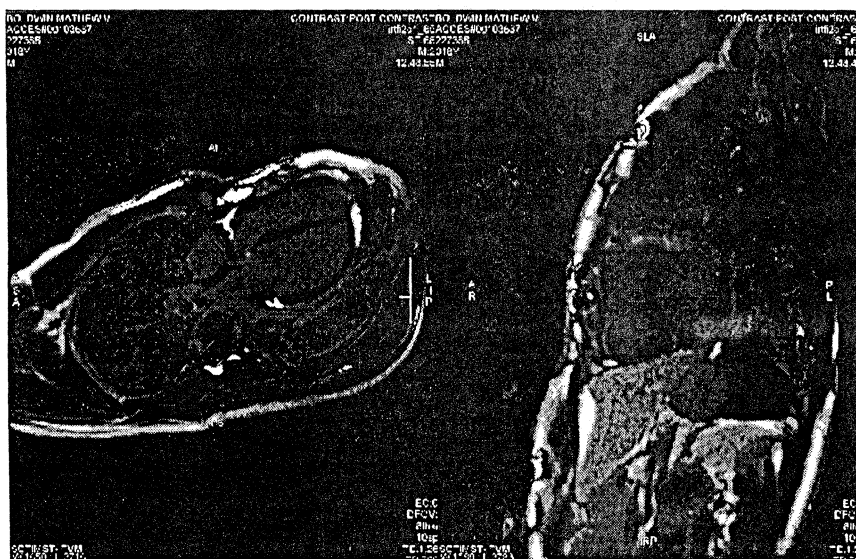


The advent of rapid MRI techniques has made it possible to study in real-time physiological events such as myocardial perfusion. The ideal MR perfusion sequence is an ultra fast sequence that entirely covers the heart. The ideal contrast agent for perfusion is not yet available, Extra cellular contrast agents are hampered by their rapid diffusion into interstitial space. Intra vascular contrast agents provide lower SNR.

VIABILITY

TFI 65-PSIR-T1-1SLICE

Appropriate inversion time (approximately 200 msec) is used to null the signal intensity of normal myocardium - critical for accurate delineation of the infarcted region. The healthy myocardium appears dark, whereas the infarcted myocardium appears bright. Infarcted or edematous tissue - initially no contrast uptake or little signal change. After 15min of contrast injection - no complete washout - delayed enhancement. Bright tissue on delayed images - DEAD tissue.



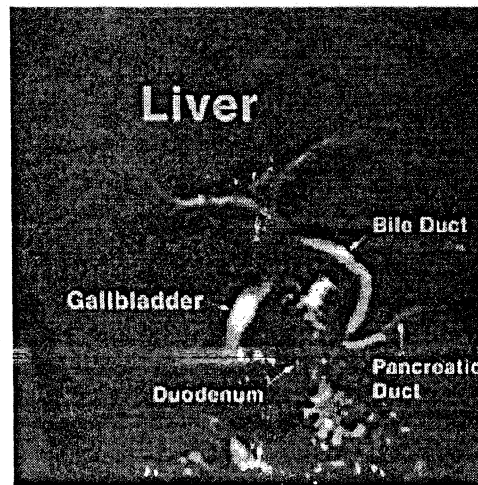
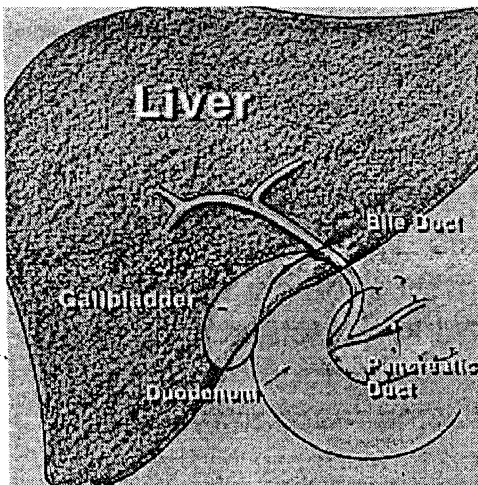
disorders, such as tumor, myocarditis, and hypertrophy cardiomyopathy. Thus, the location of myocardial enhancement (ie, subendocardial vs midmyocardial) and clinical circumstances are key to proper interpretation of pathologic contrast enhancement.

MAGNETIC RESONANCE CHOLANGIOPANCREATOGRAPHY

MRCP is introduced in 1991; it represents a relatively new development in MR technology that allows for rapid evaluation of the biliary tract, pancreatic duct and gallbladder without contrast material administration, instrumentation or radiation.

Special imaging sequences that are heavily-T2-weighted are utilized to depict the biliary tract, pancreatic duct and gallbladder as high signal intensity or bright structures owing to the fluid within them. The accuracy of MRCP is comparable to that known as ERCP (endoscopic retrograde cholangiopancreatography, the traditional but invasive means of imaging the pancreaticobiliary system) in the evaluation of choledocholithiasis, malignant obstruction, anatomic variants and chronic pancreatitis. MRCP can be completed in 10 minutes and is easily performed as an outpatient examination. In MRCP and its clinical applications. Technical refinements such as fast MR sequences that allow for imaging of the entire biliary tract and pancreatic duct in a single breathhold have resulted in marked improvement in the quality and diagnostic yield of MRCPs. As the quality of MRCPs has improved, the clinical applications of this technique have expanded such that MRCP is now replacing diagnostic ERCP in many instances.

Some basic principles apply to the acquisition of MRCP images. First the biliary tract and pancreatic duct must be localized; this is often accomplished with a scout MRCP performed with a thickness of 40-80 mm so that all portions of the pancreaticobiliary tract are included. This technique is referred to as multislice, thin-slab MRCP



The performance of MRCP does not require intravenous or oral contrast material administration because the biliary tract, pancreatic duct and gallbladder are depicted as high signal intensity or bright structures due to the fluid within them. Although patients are not required to fast prior to MRCP, patients are asked to avoid heavy ingestion of food or liquid immediately prior to the MRCP since a fluid-filled stomach may obscure the bile or pancreatic ducts. Prior to scheduling a patient for MRCP, the patient is carefully screened for contraindications to being placed in the MR scanner. Such contraindications include cerebral aneurysm clips and cardiac pacemakers. If patients give a history of possibly having metallic fragments in their eyes related to grinding metal, a radiograph (x-ray image) of the eyes is obtained prior to the MRCP.

MR TECHNIQUES

Because MRCP is a relatively new imaging technique and because of ongoing advances in software and coils, the technique is evolving and thus varies from institution to institution. MRCP is usually performed with heavily T2-weighted sequences by using fast spin-echo or singleshot fast spin-echo software and both a thick-collimation (single-section) and thin collimation (multisection) technique with a torso phased-array coil. The coronal plane is used to provide a cholangiographic display, and the axial plane is used to evaluate the pancreatic duct and distal common bile duct. In addition, some

institutions perform three-dimensional reconstruction by using a maximum-intensity projection (MIP) algorithm on the thin-collimation source images. Although the thick-collimation and three-dimensional MIP images more closely resemble conventional cholangiograms and are familiar to many clinicians, spatial resolution is degraded because of volume-averaging effects. The source images, which provide greater spatial resolution, must be carefully scrutinized so as not to overlook small luminal filling defects and strictures, which may be obscured on the thicker-collimation images. Fast spin-echo MRCP is performed by using respiratory gating; a long echo train (ie, 32); a long repetition time (three to five respiratory cycles, 8,000-10,000 msec); an echo time greater than 250 msec; fat saturation; thin collimation (3 mm with no gap); and three excitations. Imaging time is usually 4-6 minutes. Although the use of a long-echo-train sequence considerably reduces imaging time compared with that of conventional T2-weighted spin-echo sequences, artifacts from respiration and peristalsis are still problematic because the signal is averaged over multiple respiratory cycles. Single-shot fast spin-echo is a newer and more rapid MRCP sequence that can be performed in a single breath hold, thereby significantly reducing motion artifacts and increasing image quality. As a result of less motion artifact (noise) with single-shot fast spin-echo MRCP, the signal-to-noise ratio increases compared with that of fast spin-echo MRCP. However, there is also a decrease in signal, albeit not as great as the decrease in noise. The result is less spatial resolution. The speed of the acquisition is due to a very long echo train (ie, 128); an infinite repetition time; and the ability to reconstruct images after acquiring only half of the phase-encoding steps. We use an echo time of greater than 800 msec to suppress signal from most extraductal structures including fat, eliminating the use of fat saturation. All the information is acquired in less than 2 seconds from approximately 128 echoes generated from a single 90° pulse. Imaging time for both single-section and multisection sequences is less than 30 seconds. Because all the image information from one section is acquired before proceeding to the next section, breath-hold and non-breath-hold techniques can be used. However, we prefer the breath-hold technique to avoid artifacts from respiratory motion and to avoid section misregistration. The combination of rapid

sequences and the torso phased-array coil, which increases signal-to-noise ratio, makes



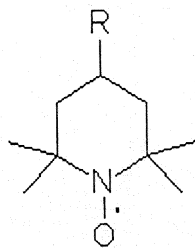
CONTRAST AGENTS FOR MRI

Paramagnetic Contrast Agents :

General considerations:

Spin-lattice relaxation time T_1 and spin-spin relaxation time T_2 may be shortened considerably in presence of paramagnetic species. The resulting effects can be seen best in NMR spectroscopy. While shortening of T_1 leads to increase in signal intensity, shortening of T_2 produces broader lines with decreased intensity. The net result is a nonlinear relationship between signal intensity and the concentration of the contrast agent. At low concentrations, an increase in contrast agent provides an increase in signal intensity due to effect on T_1 until the optimal concentration is reached. Further increase in concentration reduces the signal because of effect on T_2 . Therefore, in clinical practice it is possible to achieve less than optimal contrast effect and even produce a negative contrast effect. This dictates the use of agents that have a relatively greater effect on T_1 than on T_2 , as well as the use of pulse sequences that emphasize the changes in T_1 .

Paramagnetic species are species which have unpaired electrons. It may be simple substance (i.e. molecular oxygen), stable radical (i.e. nitroxide radical) or metal ion (i.e. many transition metal ions).



Stable nitroxide radical

Figure 1

Radicals generally cause damage to the living tissues. Therefore, they are not suitable candidates for medical MRI purposes. The paramagnetic effect of oxygen, although demonstrable, seems too weak for practical applications. Paramagnetic metal ions do show suitable effect which depends on number of unpaired electrons in the ion. The following table shows some of the paramagnetic metal ions.

Table 1

Atomic N	Ion	3d	4f	Magnetic moment (Bohr magneton)
24	Cr 3+	$\uparrow \uparrow \uparrow$		3.8
25	Mn 2+	$\uparrow \uparrow \uparrow \uparrow \uparrow$		5.9 (weak field)
26	Fe 3+	$\uparrow \uparrow \uparrow \uparrow \uparrow$		5.9 (weak field)
29	Cu 2+	$\uparrow \downarrow \uparrow \downarrow \uparrow \downarrow \uparrow \downarrow \uparrow$		1.7-2.2
63	Eu 3+		$\uparrow \downarrow \uparrow \uparrow \uparrow \uparrow \uparrow \uparrow$	(6.9)
64	Gd 3+		$\uparrow \uparrow \uparrow \uparrow \uparrow \uparrow \uparrow \uparrow$	7.9
66	Dy 3+		$\uparrow \downarrow \uparrow \downarrow \uparrow \uparrow \uparrow \uparrow$	(5.9)

There are more transition metals and lanthanide metals with unpaired spins, but for the metal to be effective as a relaxation agent the electron spin-relaxation time must match the Larmor

frequency of the protons. This condition is met better for the Fe³⁺, Mn²⁺ and Gd³⁺ ions (about 10⁻⁸ - 10⁻

10 s, others have 10⁻¹¹ - 10⁻¹² s).

The main problem with paramagnetic heavy metal ions in their native form is their toxicity. Investigation has focused on the development of stable paramagnetic ion complexes. Both the metal ion and the ligand usually exhibit substantial toxicity in the unbound state. Together, however, they may create a thermodynamically and kinetically stable compound which is much less toxic. Complexation of the metal ion with organic ligand, while considerably decreasing toxicity, may alter paramagnetic properties of the metal.

Chromium-EDTA complex was the first such agent tried, but problems with synthesis and long-term stability prevented its clinical application.

Gadolinium-DTPA complex, a renally excreted chelate with a very high formation constant (Table 2), had sufficiently favorable properties to be approved by Food and Drug Administration of USA for use in cranial disease diagnostics in mid-1988.

Table 2

Gd-DTPA formation constant = 10^{23}
 Half-life in urine = 21.0 min after intravenous injection
 in blood = 19.6 min

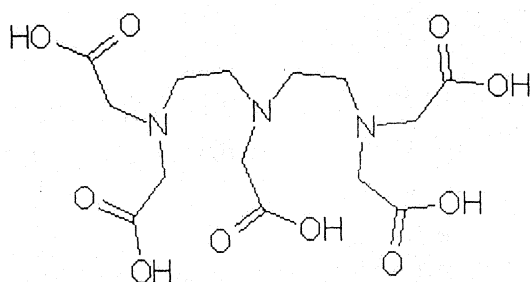
Compound	LD ₅₀ with intravenous dose in rats (mmol/kg body weight)
Gd-DTPA	10
Gd-EDTA	0.3
Gd-Cl ₃	0.4
Meglucamine diatrizoate (common X-ray contrast agent)	18

Another, relatively new type of paramagnetic contrast agents is so-called superparamagnetic iron oxide (SPIO) based colloids.

They consist of nonstoichiometric microcrystalline magnetite cores which are coated with dextrans or siloxanes. Use of these colloids as tissue-specific contrast agents is now a well-established area of pharmaceutical development.

Gadolinium(III) complexes

The prominent feature of Gadolinium(III) is the high number of unpaired electrons - seven. The Gd^{3+} ion retains a number of unpaired spins when bound to the organic ligand. The free Gd^{3+} ion is extremely toxic. Number of its complexes are very stable and thus exhibit much less toxicity. As mentioned before, Gd-DTPA complex has been approved to clinical use and is now marketed in USA under the name "Magnevist".

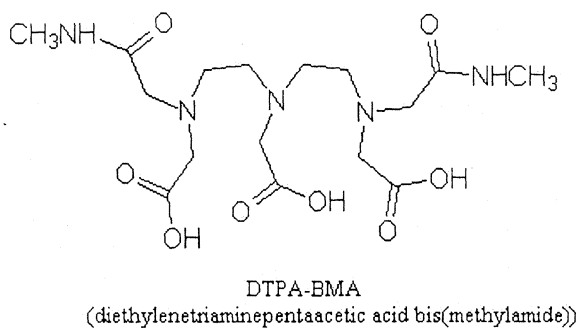


DTPA (diethylenetriaminepentaacetic acid)

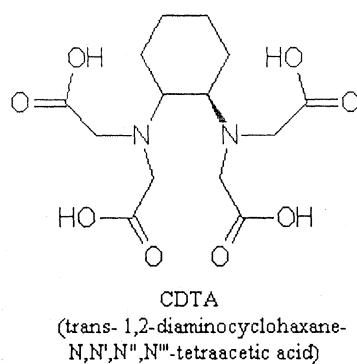
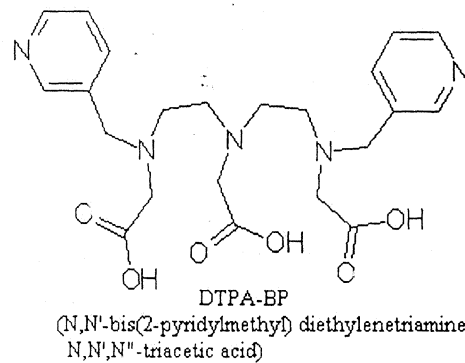
1

The relationship between the thermodynamic stability of the complex and the acute toxicity *in vivo* seems to be more complex, as demonstrated by study on that matter. Authors investigated stability of the series of complexes of several ligands with Gd^{3+} versus acute toxicity on mice. They also attempted to measure the rate of transmetallation of Gd^{3+} by Cu^{2+} and selectivities of different ligands towards Gd^{3+} as compared to Zn^{3+} , Cu^{3+} , Ca^{3+} . Iron(III) was not considered because it is tightly bound *in vivo* by the storage proteins ferritin and hemosiderin and is essentially unavailable for interaction with Gd^{3+} complexes. The main competitor to Gd^{3+} was found to be Zn^{2+} , and the most important thermodynamic criterion of toxicity is the selectivity of the ligand for Gd^{3+} over other endogenous metal ions. Zinc transmetallation was found to be the most likely mode of both acute and subchronic toxicity in experiments on rats. Complexes whose structure make *in vivo* transmetallation

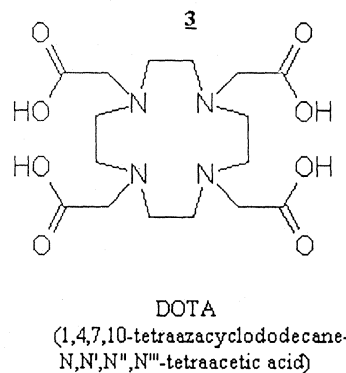
reactions much slower than renal excretion rates have significantly improved toxicities than would be predicted by thermodynamics. Slower clearance from the body is likely to significantly increase the toxicity of any Gd^{3+} complex. The following Figure shows other ligands often used in gadolinium studies.



2



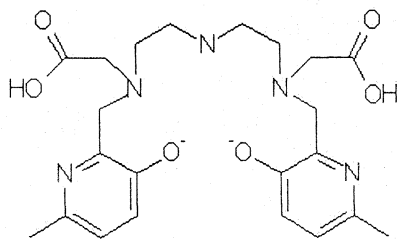
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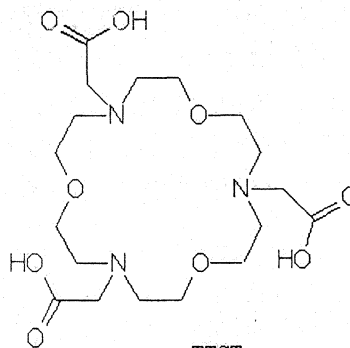
Gd^{3+} has been shown to inhibit Ca^{2+} binding to mammalian cardiac sarcoplasmic reticulum. The mechanism of toxicity could involve hemodynamic disruption.

An example of two another perspective ligands is shown on Figure 3.



DTTA-HP
(N,N'-bis(3-hydroxy-6-methyl-2-pyridylmethyl)di-
ethylenetriamine-N,N',N''-triacetic acid)

6



TTCT
(1,7,13-triaza-4,10,16-trioxacyclooctadecane-
N,N',N''-triacetic acid)

7

Potential improvement also lies in the idea of covalently coupling the ligand to protein to generate tissue-specific contrast agents.

Understandably, search for new potential ligands for Gd³⁺ complexation is still hot area of investigation. The complexes are evaluated against Gd-DTPA, the only approved compound for human use so far. Criteria include thermodynamic stability, rates of excretion, toxicity, lipophilicity, biodistribution, percent change in MR signal intensity. Some of the complexes are slightly better than Gd-DTPA in particular tests, others are a little worse. However, neither of them got as much attention as Gd-DTPA itself. This complex is far not be the best choice as of today, but it is relatively well-known choice, widely used in many MRI facilities on a daily basis. Accumulated experience allows in many cases offset the shortcomings of Gd-DTPA, so the agent continues to play indispensable role in modern MR imaging.

Monocrystalline Iron Oxide Nanocompounds Compounds called MION constitute relatively new but rapidly evolving area in MRI contrast agents. As compared to the single approved gadolinium-containing complex, there are variety of MION (also called SPIO - SuperParamagnetic Iron Oxide) reagents available on the market. Flashy names

Feridex I.V.TM, EndoremTM, GastromarkTM, LumiremTM, SineremTM and more patents pending tell us that the last word in the area is yet to be said.

These compounds consist of nonstoichiometric microcrystalline magnetite cores which are coated with dextrans (in ferumoxides) or siloxanes (in ferumoxils). SPIO agents are much more effective in MR relaxation than their paramagnetic counterparts. Since they are nonstoichiometric, there was and still is much interest in studying these compounds with all the vast array of modern physical-chemical methods: single crystal X-ray diffraction, powder X-ray diffraction, Mossbauer spectroscopy, transmission electron microscopy, dynamic light scattering, atomic adsorption spectroscopy, spectrophotometry, electron microscopy, superconducting quantum interference devices etc.

The compositions and physiochemical properties of nonstoichiometric magnetites are continuously variable between those of Fe_3O_4 and Fe_2O_3 . Conceptually, these cation-deficient, inverse-spinel phases are formed by partial oxidation of Fe(II) in stoichiometric magnetite²⁴. The Fe^{2+} content is typically 8-15 mol%. The lattice parameters of these colloids also fall between those of Fe_3O_4 and Fe_2O_3 .

The particles are usually of varying sizes from several to several hundred nanometers. They are irregular in shape and highly light-absorbing. They have no magnetic hysteresis at ambient temperatures, which is characteristic of superparamagnetic materials. Mossbauer spectra are characteristic of small (<10nm) magnetic domains which undergo superparamagnetic relaxation or collective magnetic excitation on the Mossbauer time scale. There is no evidence of covalent bonding between the iron surface and the surrounding dextran.

SPIO compounds are promising contrast agents since their properties may be fine-tuned for the specific application. They are non-toxic and rapidly cleared from the organism. Experiments have been successful in receptor-specific SPIO delivery.

Metalloporphyrines of Iron(III) and Manganese(III)

Porphyrins have been known for decades as indicators of various metabolic disorders and disease states. They are used in photodynamic therapy of tumors. Low toxicity of metalloporphyrins and their selective retention in tumors have led recently to their study as a MRI contrast media. Article contains theoretical treatment of relaxivity of some of the metalloporphyrins. This is a fairly new area of development, but metalloporphyrins of Mn(III) and

Fe(III) show favorable properties as MRI contrast agents for tumor detection. No doubt, it will lead to new discoveries soon.

Native Proteins Acting as Contrast Agents

Heme-containing proteins may act as "natural" contrast agents, just like previously discussed iron(III) porphyrins. Hematomas are easily identified by MRI. Paramagnetic deoxyhemoglobin within intact blood cells causes a local region of high magnetic field. This results in rapid dephasing of water protons diffusing in the region of acute hematoma with shortening of T₂, which results in low signal. A similar phenomenon is observed when magnetically susceptible ferritin is deposited in macrophages in hemochromatosis.

Although these not a specifically designed contrast agents, their properties may be successfully used for diagnostics in specific cases.

Gastrointestinal Contrast Agents

The gastrointestinal tract cannot be reliably studied by MRI without the use of contrast agents. Oral contrast agents may dramatically improve utility of MRI for gastro- diagnostics. The only clinically approved agents for that purpose are soluble iron compounds. (ferrous gluconate, ferric ammonium citrate) and Gd-DTPA. There is a problem with dosage of iron salts, which may not exceed the levels above those when iron supplementation is used. There are no particulate agents approved for oral use yet.

Other Contrast Agents

Examples of contrast agents other than paramagnetic involve rather mechanical action than changing the characteristics of the surrounding tissue. They are thus similar to agents used in X-Ray methods. One recent example successfully utilized vegetable oil for rectal MRI applications. Other possibilities include insufflation of air to achieve better contrast of intestinal walls .

ARTIFACT

An image artifact is a structure not normally present but visible as a result of a limitation or malfunction in the hardware or software of the MRI device, or in other cases a consequence of natural processes or properties of the human body. The knowledge of MRI artifacts and noise producing factors is important for continuing maintenance of high image quality. Artifacts may be very noticeable or just a few pixels out of balance but can give confusing artifactual appearances with pathology that may be misdiagnosed.

Image distortions related to changes in patient position, pulse sequence, metallic artifacts, or other imaging variables can be affected by the operator; artifacts due specifically to the MR system may require a service engineer.

Many types of artifacts may occur in magnetic resonance imaging. Artifacts in magnetic resonance imaging are typically classified as to their basic principles,

- Physiologic (motion),
- flow.
- Hard ware (electromagnetic spikes, ringing)
- Inherent physics (chemical shift, susceptibility, metal)

1. Chemical Shift Artifacts
2. Aliasing
3. Black Boundary Artifacts
4. Gibbs or Truncation Artifacts
5. Zipper Artifacts
6. Phase-encoded Motion Artifacts
7. Entry Slice Phenomenon
8. Slice-overlap Artifacts
9. Magic Angle Effects
10. Moire Fringes
11. RF Overflow Artifacts
12. Central Point Artifact
- Susceptibility Artifacts
- Zero-fill Artifact (Zebra Artifact)

Several techniques are developed to reduce these artifacts (e.g. respiratory compensation, cardiac gating, eddy current compensation) but sometimes these effects can also be exploited, e.g. for flow measurements

Aliasing Artifact

Aliasing is an artifact that occurs when the field of view (FOV) is smaller than the body part being imaged. The part of the body that lies beyond the edge of the FOV is projected on to the other side of the image.

Aliasing in the frequency direction can be eliminated by sampling the signal twice as fast or by applying frequency specific filters to the received signal.

A similar problem occurs in the phase encoding direction, where the phases of signal-bearing tissues outside of the FOV in the y-direction are a replication of the phases that are encoded within the FOV.

Phase encoding gradients are scaled for the field of view only, therefore tissues outside the FOV do not get properly phase encoded relative to their actual position and 'wraps' into the opposite side of the image

Use a larger FOV, RFOV or 3D Volume, apply presaturation pulses to the undesired tissue, adjust the position of the image center, or select an imaging coil which will not excite or detect spins from tissues outside of the desired FOV. In the phase direction, the number of phase encoding steps must be increased with a longer study as a result.

When this is not possible it can be corrected by oversampling the data. Oversampling is used in the frequency direction to eliminate the aliasing. No Phase Wrap (Foldover Suppression) options typically correct the phase encoding by doubling the field of view, doubling the number of phase encodes (to keep resolution constant) and halving the number of averages (to keep scan time constant) then discarding the additional data and processing the image within the desired field of view (but this is more time consuming).

Tissue outside this doubled area can be folded nevertheless into the image as phase wrap. In this case combine more than 2 number of excitations / number of signal averages with foldover suppression.

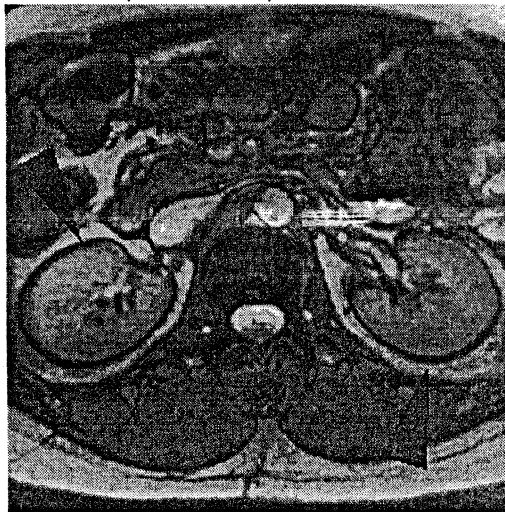
Artifact by Patient Movement

Patient movement during the scans are often an imaging problem. Artifacts from patient movement are widely varied due to a dependence when during k-space filling the motion occurs. When the patient moving causes only in the last few seconds of the scan at that time the outside edges of K-space were being filled, and as a result the artifact does not overly affect the image (there are only fine lines.

Black Boundary Artifact

The black boundary artifact is an artificially created black line located at water-fat interfaces such as muscle-fat interfaces. If the MR imager is tuned or centered to the frequency of water, fat will be shifted or spatially mismatched relative to its true spatial location. When chemical shift is greater than or equal to the size of an individual pixel a dark or bright band of signal intensity will occur at the lipid -water interface in the frequency encoding direction of the image. This pixel-by-pixel misregistration along the frequency encoding direction visibly manifests itself as a bright or dark band running perpendicular to the frequency encoding direction.

This artifact can occur for a couple of reasons. The most common reason (for gradient echo sequences) is a result of selecting an echo time in which the fat and water spins are out of phase, canceling each other. At 1.5 T, the 3.5 ppm difference in frequency between water and saturated fat results in cancellation of spins at 4.5 ms multiples; for example at 6.8ms, 11.3ms, and 15.9 ms.



Cardiac Motion Artifact

Movement of the heart causes blurring and ghosting in the images. The artifacts appear in the phase encoding direction, independent of the direction of the motion.

These artifacts can be reduced by using cardiac synchronization: triggering, gating or retrospective triggering. Maximum reduction can be achieved by using triggering in combination with flow compensation, respiratory triggering or breath hold and regional saturation techniques

Central Point Artifact

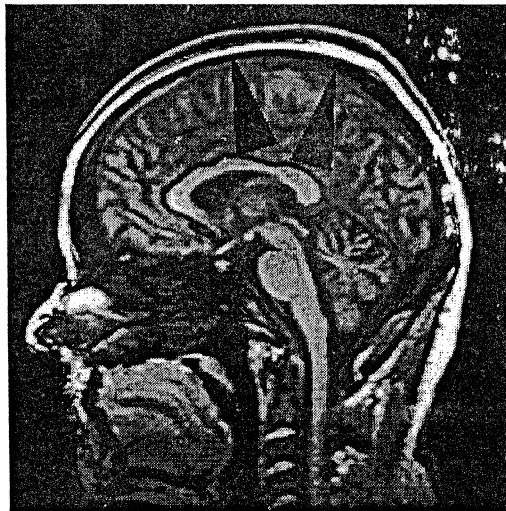
The central point artifact is a focal dot of increased signal in the center of an image. Temperature fluctuations often cause dc-coupled amplifiers to have non-zero outputs with zero input. It is caused by a constant offset of the DC voltage in the receiver and therefore an offset in signal intensity of all raw data points. Normally software compensation prevent it, by applying a 'dc correction' or 'baseline correction' before the FT.

Chemical Shift Artifact

Large B_0 and chemical shift differences between tissues cause artifacts. The MRI scanner uses the frequency of the signal to indicate spatial position. During frequency encoding, fat protons precess slower than water protons in the same slice because of their magnetic shielding. The signal from the fat protons is incorrectly encoded. The net effect is that the fat and water, located in the same voxel are encoded as located in different voxels and are shifted in the frequency direction from their true positions. This chemical shift misregistration cause accentuation of any fat-water interfaces along the frequency axis of MRI's and may be mistaken for pathology. Where fat and water are in the same location, this artifact can be seen as a bright or dark band at the edge of the anatomy. Protons in fat and water molecules are separated by a chemical shift of about 3.5 ppm. The actual shift in hertz (Hz) depends on the magnetic field strength of the magnet being used. The artifact is greater at higher field strengths and lesser at higher gradient strengths. For a 0.3 T system operating at 12.8 MHz the shift will be

44.8 Hz compared with a 223.6 Hz shift for a 1.5 T system operating at 63.9 MHz

For artifact reduction helps a smaller water fat shift (higher bandwidth), a higher matrix, an in phase TE or a spin echo technique. Since the misregistration offset is present in the read out axis the patient may be rescanned with this axis parallel to the fat-water interface. Steeper gradient may be employed to reduce the chemical shift offset in mm. Another strategy is to employ specialized pulse sequences such as fat saturation or inversion recovery imaging. Practically about the best way to eliminate this artifact is to use a fat suppression technique



Crisscross / Herringbone Artifact

This artifact appears as a herringbone pattern scattered over the whole image in any direction only on one slice or on multiple slices. The causes of this are many and various, like e.g. electromagnetic spikes created by the gradients, electronic equipment inside the MR procedure room, or fluctuating AC current.

Crosstalk (Artifact)

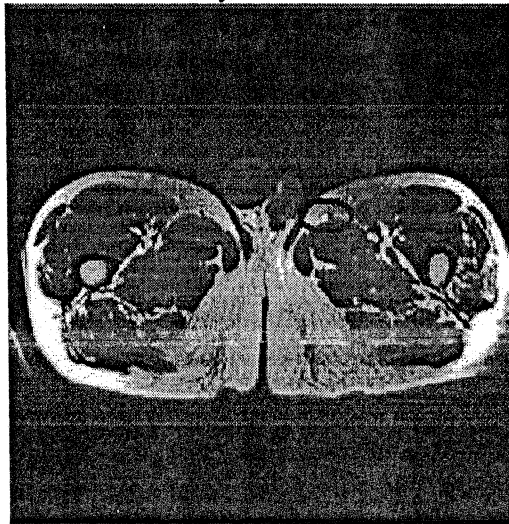
Crosstalk is an artifact introduced into images by interference between adjacent slices of a scan, caused by a slice profile that is not ideal due to the constraints of the measurement technology. If the slice distances are too small, there is cross talk between the slices, which can affect T1 contrast

This artifact can be eliminated by limiting the minimum spacing (for the most sequences a minimum gap 10% and for IR sequences 20%) between the slices. Crosstalk can also be reduced by selection of interleaved slices (so a slice gap will not be necessary), but interleaved data acquisition can produce large mean intensity differences between adjacent slices

Entry Slice Phenomenon (Artifact)

Entry slice phenomenon occurs when unsaturated spins in blood first turns into a slice or slices. This artifact is characterized by bright signal in a blood vessel (artery or vein) at the first slice that the vessel enters. Often the signal is seen on more than one slice, fading with distance.

Confusion with thrombosis can be the result of this artifact. The characteristic location and the use of gradient echo flow techniques can be used to differentiate entry slice artifacts from occlusions



Ghosting Artifact

An image artifact caused through moving (respiratory, bowel motion, heartbeat) primarily associated with the phase direction

Artifacts can be reduced by respiratory and cardiac triggering, the use of breath holding pulse sequences and flow compensation. Another method of reducing ghosting artifacts is to use a pharmaceutical,

such as glucagon or scopolamine, to reduce bowel motion. This will decrease artifacts from both peristalsis and breathing.

Gibbs Artifact

This artifact is a series of bright or dark lines that are seen parallel and adjacent to borders of abrupt intensity change (sharp intensity edge) in an image, as when going from bright CSF to dark spinal cord on a T2 weighted image.

In the spinal cord, the gibbs phenomenon can simulate a small syring to the unaware. It is also seen in other locations as at the brain/calvarium interface.

Fine lines visible in an image may be due to undersampling of the high spatial frequencies, respectively incomplete digitization of the echo. It is related to the finite number of encoding steps used by the Fourier transformation to reconstruct an image. This means the signal has not decayed to zero by the end of the acquisition window, and the echo is not fully digitized.

It is related to the finite number of encoding steps used by the Fourier transformation to reconstruct an image. The more encoding steps, the less intense and narrower the artifacts. Therefore, e.g. the artifact is more pronounced in the 256 point dimension of a 512x256 acquisition matrix

This problem can be easily resolved by taking more samples - a higher acquisition matrix and/or a smaller FOV.

Gradient Related Artifact

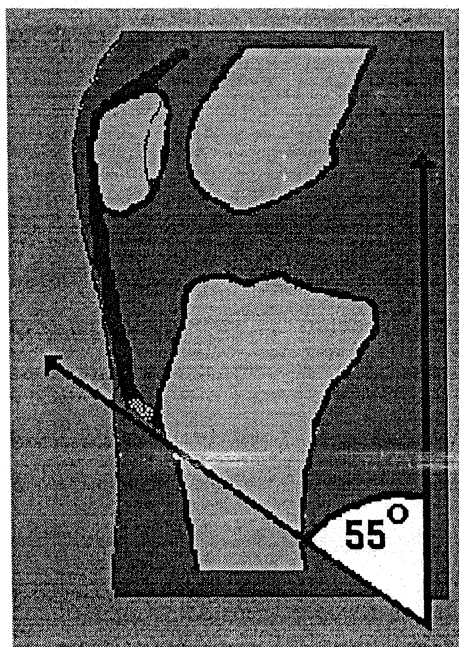
Artifacts arising from problems with the gradient system are sometimes very similar to those described as B0 inhomogeneities. A gradient that is not constant with respect to the gradient direction will distort an image. This is typically only possible if a gradient coil has been damaged. Other gradient related artifacts are due to abnormal currents passing through the gradient coils.

Flow Artifact

Flow effects in MRI can produce a range of artifacts. Flow artifacts are usually include intravascular signal void by 'time of flight' effects, turbulent dephasing and first echo dephasing, caused by flowing blood. A liquid flowing through a slice can experience a RF pulse and then flow out of the slice by the time the signal is recorded (because the repetition time (TR) is asynchronous with the pulsatile flow). The blood flow occasionally produces an intravascular high signal intensity due to flow related enhancement, even echo rephasing and diastolic pseudogating. The pulsatile laminar flow within vessels often produces a complex multilayered band that usually propagates outside the head in the phase encoded direction. Blood flow artifacts should be considered as a special subgroup of motion artifacts. Artifacts can be reduced by reduction of phase shifts with flow compensation (gradient moment nulling), suppression of the blood signal with saturation pulses parallel to the slices, synchronization of the imaging sequence with the heart cycle (cardiac triggering) or can be flipped 90° by swapping the phase/frequency encoding directions.

Magic Angle Effects

Magic angle effects are seen most frequently in tendons and ligaments that are oriented at about a 55 degree angle to the main magnetic field. Signal from water molecules associates with the tendon collagen fibers is not normally seen because of dipolar interactions that result in very short T2 Times. At an angle of about 55 degrees to the main magnetic field, the dipolar interactions become zero, resulting in an increase of the T2 Times about 100 fold. This results in signal being visible in tendons with ordinary pulse sequences. A bright signal from this artifact is commonly seen in the rotator cuff and occasionally in the patellar tendon and elsewhere. The following image shows increase signal in the distal patellar tendon from this magic angle effect.



Moire Fringes

Moire fringes are an interference pattern most commonly seen when doing gradient echo images with the bodycoil. Because of lack of perfect homogeneity of the main magnetic field from one side of the body to the other, aliasing of one side of the body to the other results in superimposition of signals of different phases that alternatively add and cancel. This causes the banding appearance and is similar to the effect of looking through two screen windows.

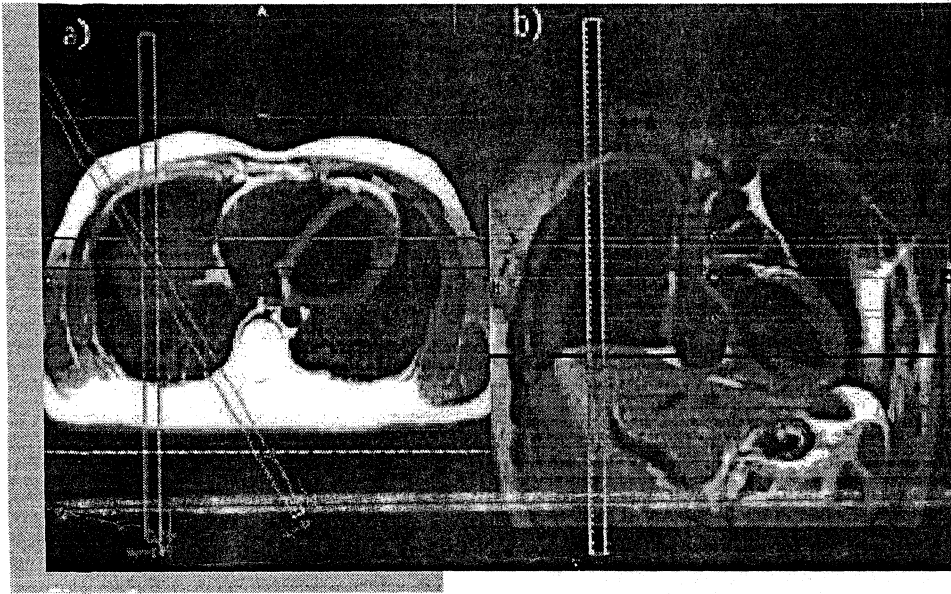
Motion under Control

with Prospective Acquisition Correction (PACE)

1D-PACE and 2D-PACE

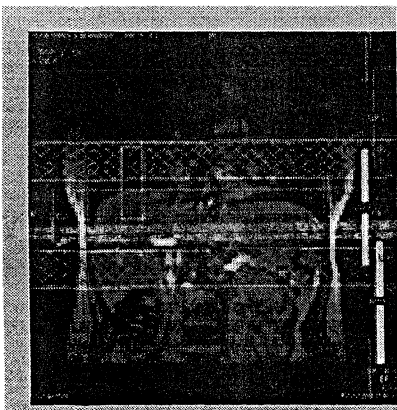
Method

The fastest method of detecting motion is 1DPACE (also known as a "navigator" technique). It typically requires only 30 ms and is used primarily for minimizing the effects of breathing motion in cardiac exams. For this purpose, a single line of data from a pencil-shaped volume that crosses the diaphragm is acquired. The volume is interactively placed (Fig. 1) in such a way that the position of the diaphragm can be calculated and used for motion correction – in real time.



2D-PACE,

An image is acquired by means of a low-resolution gradient echo sequence featuring a low flip angle; this ensures that magnetization is not saturated, so that dark lines in the image are avoided. The user places a small box across the diaphragm on the 2D image. The change in signal intensity along the axis of the box is used to determine the position of the diaphragm. Since a 2D image provides more information than a single line, this method is very robust. The time needed to acquire an image for 2D-PACE is around 100 ms. The highly reliable 2D-PACE technique is unique to Siemens.



Application:

Multiple Breath-hold Examinations

For patients who can hold their breath for only a short time, the acquisition can be split up into multiple breath-holds. The information about the diaphragm position allows the operator to monitor the breathing pattern of the patient online. Furthermore, acquisition of slices during different breath-holds can be aligned in order to compensate for imperfect reproducibility of the breathhold position. In this way, gaps between slices or overlaps are avoided.

3D-PACE

Functional MR imaging (fMRI) is another application where motion detection, and instantaneous adjustment of the acquisition according to this information, are crucial. Here, complete multi-slice EPI datasets of the head are acquired in rapid succession during presentation of various stimuli. In order for the statistical analysis to be successful, the datasets need to be aligned perfectly. For this purpose, each 3D dataset is compared with the previous one and the translation as well as the rotation of the head are calculated (and displayed) in real-time. The software is able to compensate for rotations and translations in all 6 degrees of freedom (a feat that is unmatched in the MR industry). The technique can therefore account, in real time, for any so-called "rigid-body motion". For acquisition of the next dataset, slice position and orientation are adjusted according to the altered position of the head.

MRI- SAFETY

Safety can be divided up into three areas:

- Main Magnetic Field
- Varying Magnetic (Gradient) Fields
- Radio Frequency
-

1. Main Magnetic Field and Safety

The main magnetic field of a 1.5 T magnet is about 30,000 times the strength of the earth's magnetic field. It is strong enough to pull fork-lift tines off of machinery, pull heavy-duty floor buffers and mop buckets into the bore of the magnet, pull stretchers across the room and turn oxygen bottles into flying projectiles. Deaths have occurred from trauma as a result of these effects. Smaller objects such as pagers, bobby pins and pens have been known to be pulled off the person carrying them.

The strong field also effects common devices such as pacemakers and watches. The magnetic reed switch in modern pacemakers is disturbed by strong magnetic fields resulting in possible deleterious effects to the patient with one implanted. Mechanic watches will "freeze up" in a strong field, sometimes permanently. Many intracranial aneurysm clips are ferromagnetic and as a result experience a torque or twisting in a magnetic field. Not everyone with an aneurysm clip experiences a fatal hemorrhage when placed in a magnet, but several cases have been reported.

Some types of heart valves (e.g., Star-Edwards) are torqued in a magnetic field: however, this torque is less than the stresses that occur normally as a result of blood flow. Therefore heart valves are now considered not to be an absolute contraindication for MRI. More of an annoyance than a safety problem is the ability of the magnetic field of a MRI machine to erase the information contained on the magnetic strip on ATM and credit cards. This may occur a short distance inside of the scanner room of a MRI machine.

Some metallic objects that are usually safe near an MRI machine are gold jewelry and eyeglass frames

2. Varying Magnetic (Gradient) Fields

Varying magnetic fields are necessary in order to obtain images from MRI scanners. Changing magnetic field induce electrical currents in conductors (.this is how an electrical generator works). In patients

with metal in their body, the potential exists for electrical currents being induced in the metal with subsequent heating. This may occur with metal foreign bodies or some surgical implants. It does not universally occur and some patients with hip prostheses, for example, may, be scanned without harm. Very rapidly changing magnetic fields as may be achieved with echo planar imaging can cause nerve stimulation. This stimulation can effect motor nerves with resulting muscle contraction as well as the retina with resulting flickering lights called "magnetophosphenes".

3. Radio Frequency and safety.

The radio frequency power that is capable of being produced matches that of many small radio stations (15-20 kW). As a result there is the presence of heating effects from the RF. In most pulse sequences, the heating is insignificant and does not exceed the FDA guidelines. New pulse sequences such as for echo planar imaging and some spectroscopy localization techniques are capable of exceeding the FDA guidelines. Monitoring of the power deposition in patients is a requirement for FDA approval of clinical MRI scanners. Potential for electrical shock exists with RF coils so proper grounding and insulation of coils is necessary. Any damage to coils or their cables needs prompt attention. Also looping of the cable to a coil can result in burns to patients that come into contact with them. It is best to avoid all contact with the RF coil cables.

Human Exposure Considerations.

Potential hazards lies in,

1. Static magnetic fields.
2. Gradient magnetic fields.
3. RF electromagnetic fields.

Magnetic field Effects.

According to FDA (Food and Drug Administration - 1982) guidelines there

is maximum permissible levels for static mag. Fields. They are,

Static field - 2 Tesla.

Rate of varying field - 3 Tesla/sec.

Below these suggested guidelines no documented permanent adverse biologic Effects have been observed. The only biological effect associated with exposure to a static magnetic Field is

augmentation of the T-wave amplitude observed on an ECG and have no significant effect on health. Gradient fields can induce current in the patient's body. The current can produce mild cutaneous sensation, involuntary muscle contraction and cardiac arrhythmias. Subjective visual effects, light flashes known as magneto phospheres, may result stimulation of retina.

RF field effects.

The measure of dose of RF energy is SAR(Specific AbsorptionRate). Unit of SAR is Watt per Kilogram (W/Kg.)SAR is the power absorbed per unit mass. The main effect of RF exposure is an increase in the body temperature. Exposure to an SAR up to 0.4 W/Kg should produce no temperature increase. However, additional evaluation are needed for heat sensitive organs such as testes and eyes.

MRI During Pregnancy.

In general MRI is not believed to be hazardous to the fetus. However variety of mechanism have the potential to produce harmful interaction between electro-magnetic field and fetus. By FDA Recommendations, it is considered inadvisable to undergo MRI routinely during pregnancy until more information regarding potential hazards become available.

Magnetic Field Hazards.

Fringe field exists around the magnet. The 0.5 mT field extends about 28 ft.

from a 0.5 T magnet and 36ft. from a 1T magnet. One of the greatest potential hazard is the missile effect. Hazards also exists for patients who have medical devices implanted in their bodies such as, Cardiac pacemaker, Aneurysm clips, Shrapnel or other metallic foreign bodies, Implanted electrodes, Neuro-stimulators, Joint replacements etc.

Objects that may be damaged by a 0.5 mT field includes watches, credit cards, taperecorders, calculators, cameras etc. Careful

- No computer room required for a total of just two electronic cabinets (water-cooled)

MAGNETOM Avanto – Application

Tim Whole Body Suite

MAGNETOM Avanto features a unique telescopic patient table which enables a full Field-of-View of 205 cm, without increasing total system length. No additional table extension is required.

The table top has standard length, additional space is only required at the rear part of the magnet. Table movement to its full extent can be controlled from the operator console.

The large FoV helps in imaging metastases with sequences such as TIRM (Turbo Inversion Recovery). Whole-Body MR Angiography is possible on the entire volume with iPAT.

- Max. scan range of 205 cm
- Protocols and programs for Whole-Body MR Angiography and metastasis detection

CISS & DESS

Unique Siemens sequences and protocols

3D DESS (Double Echo Steady State):

- T2/T1-weighted
- Excellent fluid-cartilage differentiation in orthopedic imaging

3D CISS (Constructive Interference in Steady State):

- Excellent visualization of fine structures such as cranial nerves
- High resolution imaging of inner ear and spine

Advanced Cardiac

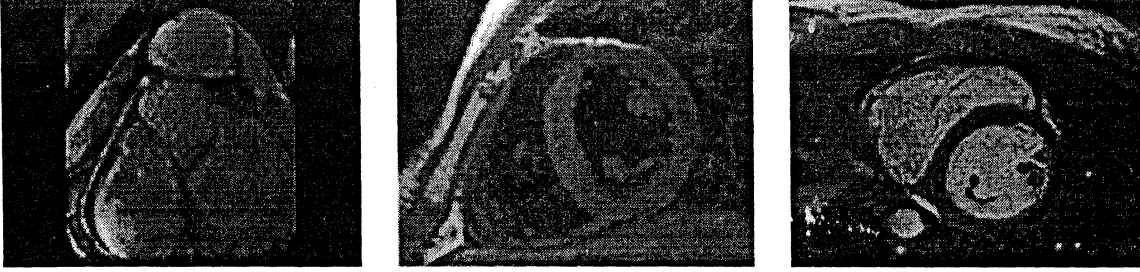
Special sequences and scan protocols for MR studies of the heart.

Morphology - Heart and Vessel Structure and Valve Function

- Dark-blood sequences using breath-hold technique
- Ventricular Function and Wall Motion
- Dynamic CINE TrueFISP imaging of cardiac function with prospective and retrospective ECG triggering, with or without breath-hold technique
- Cine imaging with echo sharing for high temporal resolution
- Triggered retrogated cine imaging with arrhythmia rejection for automatic adjustment of the number of phases to the heart rate
- Real-time cine TrueFISP imaging without need for ECG triggering of breath-hold commands
- Real-time radial imaging for high speed and high resolution cine studies
- Visualization of myocardial contractility using various tagging techniques
- Tissue characterization (differentiation of tissues with different T1 values)
- Ultra-fast, high SNR sequences for first pass imaging using TrueFISP iPAT and Half Fourier techniques. These protocols provide multi-slice information for the assessment of coronary heart disease (TrueFISP)
- Assessment of myocardial viability in breath-hold with single shot IR TrueFISP for delayed enhancement studies
- Robust and reproducible contrast between infarct and normal myocardium with phase-sensitive Inversion recovery. Adjustment of TI is no longer necessary with this technique
- Protocols for pediatrics, plaque imaging and stress imaging

Coronary Imaging

- Dedicated sequences for coronary imaging and angiography providing free breathing navigator (1D PACE) and breath hold techniques (2D and 3D Flash and TrueFISP)



Flow Quantification

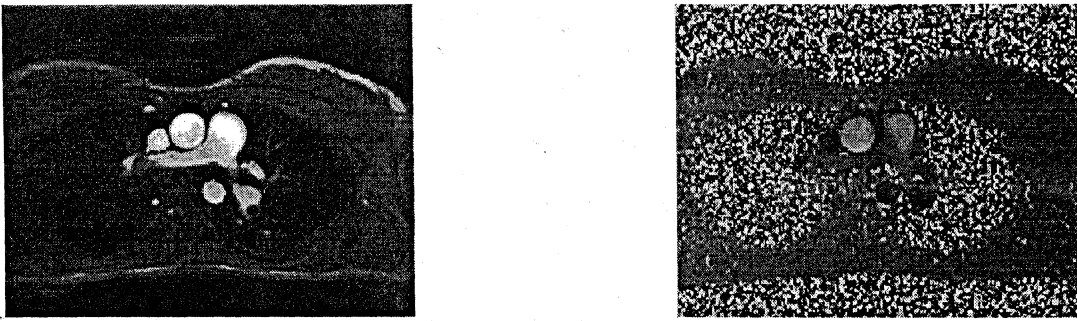
Special sequences for quantitative flow determination studies

Measuring blood/CSF flow non-invasively

- Requires Physiological Measurement Unit (PMU) option

RetroGated Flow

- Dynamic representation of temporally changing flow



Interactive Realtime

Sequences and hardware for interactive real-time scanning.

Uses ultra-fast TrueFISP and other gradient-echo sequences for high image contrast

Real-time reconstruction of the acquired data

The user can navigate in all planes on-the-fly during data acquisition

- Real-time cardiac examinations
- Real-time interactive slice positioning and slice angulation
- 3D Magellan SpaceMouse included

TGSE

Ultra-fast sequence providing high-resolution imaging or extremely short acquisition times

Hybrid turbo spin echo/gradient echo used primarily for T2-weighted imaging

- Shorter measurement time
- Decreased RF power deposition
- Improved visualization of hemorrhage, due to magnetic susceptibility differences

High resolution imaging of brain and spine

Inline Diffusion Automatic real-time calculation of trace-weighted images and ADC maps with Inline technology. Compatible to single-shot diffusion-weighted EPI

Inline Perfusion

Automatic real-time calculation of Global Bolus Plot (GBP) and Time-to-Peak map (TTP) with Inline technology

Inline Bold Imaging

Examination of intrinsic susceptibility changes in different areas of the brain, induced by external stimulation (e.g. motor or visual)

Automatic real-time calculation of z-score (t-test) maps with Inline technology, for variable paradigms

- Compatible to single-shot EPI with high susceptibility contrast for fast multi-slice imaging
- ART (Advanced Retrospective Technique) for fully automatic 3D retrospective motion correction, for 6 degrees of freedom (3 translations and 3 rotations)
- Mosaic images for efficient storage and transfer of large data sets
 - 3D spatial filtering
- Overlay of Inline calculated t-test results on the EPI images

Advanced Functional Neuro

3D PACE (Prospective Acquisition CorrEction)

Prospective motion detection and correction in the volume to eliminate motion artifacts during BOLD measurements

- Fully automatic 3D prospective motion correction during data acquisition, for 6 degrees of freedom (3 translations and 3 rotations)
- Motion correction covering the complete 3D volume
- Provides high accuracy
- Substantially reduced motion-related artifacts in t-test calculations
- Significantly increased signal changes in the activated neuronal volume
- Increased functional MRI (fMRI) sensitivity and specificity

3D Inline fMRI

- Overlay of functional information on 3D dataset during image acquisition

Multi-Directional Diffusion-Weighted Imaging (DWI)

- Measurements of multiple diffusion directions and b-values
- Suitable for investigation of anisotropic diffusion in tissue such as calculation of diffusion tensors.
-

Single Voxel Spectroscopy

Streamlined for easy push-button operation

SVS techniques SE and STEAM

- Short TEs available
- Fully automated adjustments including localized shimmering and adjustment of water suppression pulses
- Also available: Interactive adjustments and control of adjustments
- Optimized protocols for brain applications

Chemical Shift Imaging

Integrated software package with sequences and protocols for Chemical Shift Imaging (CSI)

Extension of the SVS package, offering the same level of user-friendliness and automation

- 2D and 3D Chemical Shift Imaging
- Hybrid CSI with combined Volume selection and Field of View (FoV) encoding
- Short TEs available (30 ms for SE, 20 ms for STEAM)
- Automated shimming of the higher order shimming channels for optimal homogeneity of the larger CSI volumes
- Weighted acquisition, leading to a reduced examination time compared to full k-space coverage while keeping SNR and spatial resolution
- Outer Volume Suppression
- Spectral Suppression
- Protocols for prostate spectroscopy

Breast Biopsy Software

Easy to use *syngo*-based post-processing software helps finding the coordinates for needle insertion for biopsy of localization of breast lesions detected by MR

Allows calculation of the coordinates after clicking the center of the lesion and the 0 marker of the breast biopsy device

- Printout of working sheet
- Multi-lesion calculations

Prerequisites:

- Breast Biopsy Device
- Loop Flex coil, large

Image Filter Software

For noise reduction in MR images and better edge definition

Uses high-pass and low-pass filtering and adjusts automatically to the local image content (adaptive filtering)

Main Benefits:

Noise reduction

Adaptive filtering

Applications:

Reduces noise in acquired MR images

Improves the visual appearance of the image

Image filtering

This image filtering software uses an adaptive filter from Context Vision that searches for conspicuous structures in the image and then checks whether these structures are random or part of other structures. It considers local distribution of signal intensity and the continuity of local tissue structures.

Filter strength values from 1-20 can be selected individually. Three different image filter values (smooth, medium, sharp) are individually storable for frequent use. Single and multiple images or entire series can be selected together for evaluation. An additional, selectable window shows the actual calculation status and provides a preview of already filtered images.

Argus Function

Automated tool for cardiac function evaluation

Main Benefits:

- Fully automated segmentation of high-contrast TrueFISP cine images
- A full set of analyses
 - End-diastolic and end-systolic volume
 - Ejection fraction, stroke volume and cardiac output
 - Cardiac mass
 - Ejection and filling rates
 - Regional wall thickening calculation
- Efficient review of dynamic studies
 - Up to 8 series simultaneously in a synchronized movie displayed
 - Rapid multi-level sorting
- Fast AVI creation
- For conferences, presentations and consultation over the internet

Argus Flow

Automated tool for analysis of cerebral spine fluid flow

Main Benefits:

- Semi-automatic segmentation of through-plane velocity-encoded images
- Color-encoded display of velocity in static and movie studies
- Quantification of velocity, flow of blood and CSF

Argus DynamicSignal

Region- of- Interest based analysis of dynamic image intensity variations

Main Benefits:

- Semi- automatic detection or manual selection of Regions- of- Interest
- Analysis of temporal tissue signal variations (e.g. Peak signal, time- to- peak)

- Color output curves and parametric displays
More information about Argus? Go to [syngo >>](#)

Vessel View

Interactive analysis of vessel disease using MR or CT angiography data

Main Benefits:

- Clinical protocols for reliable results
- Automatic visualization of vessel trees within the volume of interest
- Automatic visualization and measurement of vessel cross-section in MPR views
- Integrated vessel navigator

3D VRT - Volume Rendering Technique

Main Benefits:

- Fully integrated into the 3D workflow and user interface
- More productive surgical planning and discussion with referring physicians
- Complete visualization modes
Volume Rendering Technique, Maximum Intensity Projection, Multiplanar Reconstruction or Surface Shaded Display
- Many predefined settings, easily selectable from gallery
- Optimal display quality
Independent control of color, opacity and shading
- Easy elimination of obstructing anatomy
3D Object Editor

Neuro Perfusion Evaluation

Dedicated task card for quantitative processing of neuro perfusion data

Main Benefits:

- Color display of relative Mean Transit Time (relMTT)
- Flexible selection of Arterial Input Function (AIF) for reliable quantification

Echo Planar Imaging, Diffusion imaging, Perfusion and Diffusion Tensor imaging

- Single shot and segmented EPI sequences for fast acquisition
- Diffusion weighted imaging with b max of 10,000 s/mm²
- Single shot EPI for perfusion imaging
- Multidirectional Diffusion Weighted (MDDW) imaging for diffusion tensor imaging

BOLD Evaluation

Blood Oxygenation Level Dependent

Non-invasive MR-technique to detect functionally activated regions of the human brain

Examination of intrinsic susceptibility changes in different areas of the brain, induced by external stimulation (e. g. motor or visual)

Main Benefits:

Colored overlay of functional and anatomic data

BOLD

BOLD= Blood Oxygen Level Dependent

Spectroscopy Evaluation

Comprehensive and user-friendly evaluation of spectroscopy data

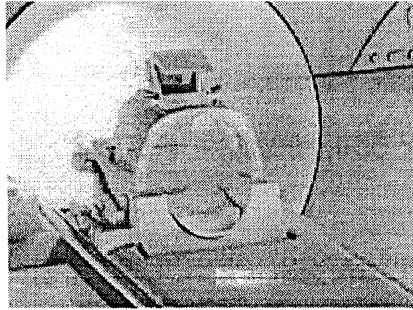
Main Benefits:

- Display of spectroscopic data as colored metabolite images or spectral overview maps and overlaid over anatomical images
- Protocol driven quantitation for routine use
- Flexible algorithms and interactive UI for research use

Easy transferring, exporting and printing of data

COILS

Head Matrix Coil



The multi-element Matrix coil technology is an essential part supplementing the most innovative Total imaging matrix.

Description:

- 12-element design with 12 integrated preamplifiers, two rings of 6 elements each (i.e. 4 clusters of 3 elements each)
- Upper coil part removable
- Lower coil part may stay on the patient table for most of the examinations
- Smoothly integrated into the patient table for most of the examinations
- Smoothly integrated into the patient table with Neck Matrix coil and Spine Matrix coil
- Open patient-friendly design
- Cushioned head stabilizers (removable)
- No coil tuning
- iPAT-compatible
- Detachable double mirror.
-
-

Applications:

- Head examination
- MR Angiography
- Combined head/neck examination
- TMJ (temporo mandibular joints)
-

Can be combined with:

- Neck Matrix coil
- Spine Matrix coil
- Body Matrix coil (up to 4; optional)
- PA Matrix coil (optional)
- All flexible coils (e.g. CP Flex coil, small, CP Flex coil, large)

- Weight: 5 kg
- Dimensions: 300 mm x 300 mm x 280 mm (LxWxH)

Neck Matrix Coil

The new multi-element Matrix coil technology is an essential part supplementing the most innovative Total imaging matrix.

Description:

- 4-element design with 4 integrated preamplifiers, 2 clusters of 2 elements each
- Upper coil part removable
- Lower coil part may stay on the patient table for most of the examinations
- Operates in an integrated fashion with the Head Matrix coil and Spine Matrix coil, for coverage of the posterior and anterior neck region
- No coil tuning
- iPAT-compatible

Applications:

- Cervical Spine
- Neck
- Larynx/Esophagus
- MR Angiography
- Mediastinum
- Combined head/neck examination

Can be combined with:

- Head Matrix Coil
- Spine Matrix Coil
- Body Matrix Coil (up to 4; optional)
- PA Matrix coil (optional)
- All flexible coils (e.g. CP Flex coils, small, CP Flex coil, large)

- Weight: 2.6 kg
- Dimensions: 190 mm x 330 mm x 332 mm (L x W x H)

Spine Matrix Coil

The multi-element Matrix coil technology is an essential part supplementing the most innovative Total imaging matrix.

Description:

- 24-element design with 24 integrated preamplifiers, 8 clusters of 3 elements each
- Smoothly integrated into the patient table and streamlined with Head Matrix coil and Neck Matrix coil
- May remain on the patient table for almost all exams
- No coil tuning
- iPAT-compatible

Applications:

- High resolution imaging of the whole spine
- Various applications in combination with additional coils

Can be combined with:

- Head Matrix Coil
- Neck Matrix Coil

- Body Matrix Coil (up to 4; optional)
- PA Matrix Coil (optional)
- All flexible coils (e.g. CP Flex coils, small, CP Flex coil, large)
- Weight: 11 kg; may remain on the patient table for almost all exams
- Dimensions: 1185 mm x 485 mm x 33 mm (L x W x H)

Body Matrix Coil

The new multi-element Matrix coil technology is an essential part supplementing the most innovative Total imaging matrix.

Description:

- 6-element design with 6 integrated preamplifiers, with 2 clusters of 3 elements each
- Operates in an integrated fashion with the Spine Matrix coils (2 rings of 6 elements each = 12-element design)
- Can be combined with further Body Matrix coils for larger coverage
- No coil tuning
- iPAT-compatible

Applications:

- Thorax (incl. Heart)
- Abdomen
- Pelvis
- Hip

Can be combined with:

- Head Matrix coil
- Neck Matrix coil
- Spine Matrix coil
- Additional Body Matrix coils (typically 2 – 3 in total)

The new multi-element Matrix coil technology is an essential part supplementing the most innovative **Total imaging matrix.**

Description:

- 16-element design with 16 integrated preamplifiers, in 8 CP pairs, i.e. 4 levels with 2 CP elements each
- Operates in an integrated fashion with the Body Matrix coils and Spine Matrix coil and for Whole-Body examinations also with the Head and Neck Matrix coil
- Automatic table feed and active coil switch
- Can be utilized Head and Feet First
- Both legs are independently covered with coil elements, maximizing the coil filling factor and the signal-to-noise ratio
- No coil tuning
- Includes special non-ferro-magnetic coil cart for safe, user-friendly storage
- iPAT-compatible

Applications:

- High-resolution angiography of both legs incl. Pelvis with highest signal-to-noise ratio
- Visualization of the iliac arteries and aorta

Can be combined with:

- Head Matrix coil
- Neck Matrix coil
- Spine Matrix coil
- Body Matrix coils (up to 3)
- All flexible coils (e.g. CP Flex coils, small, CP Flex coil, large)
- Weight: 5,75 kg
- Dimension: 970 mm x 300 – 600 mm x 270 mm (L x W x H)

Shoulder Array Coil

Description:

- 4-coil design with 4 integrated preamplifiers

- Two coils are included, one for small and one for large shoulders
- For narrow or wide shoulders the coil can be attached at different positions on the base plate
- Includes one base plate pad and one head rest for high patient comfort
- No coil tuning
- iPAT-compatible

Applications:

- Best visualization of small anatomical structures (e.g. labrum)
- Higher SNR and better field homogeneity
- Reduced slice thickness and measurement times.
-

Small Shoulder Array Coil:

- Opening: 165 mm
- Weight: 1.3 kg

Large Shoulder Array Coil:

- Opening: 200 mm
- Weight: 1.6 kg
- Dimension: 445 mm x 490 mm (L x W)
- Weight: approx. 5 kg

CP Extremity Coil

Circularly Polarized no-tune transmit/receive coil for joint examinations in the region of the lower extremities

Description:

- Transmit/receive coil
- CP coil with 2 integrated preamplifiers
- Upper coil part removable
- Holder allows off-center positioning to keep knee or foot which is not under examination in a comfortable position
- No coil tuning

Applications:

- Knee
- Ankle
- Peripheral MR Angiography

- Weight: 6.5 kg
- Dimensions: 405 mm x 270 mm x 290 mm (L x W x H)

Double Loop Array Coil

Consisting of two flexible, anatomically adaptable 7 cm ring coils for simultaneous examination of both TMJs, the eyes or the wrist. Including Double Loop Array Holder

Description:

- Circular coil pair with Double- Loop Array Holder
- No coil tuning
- Extra arm rest for wrist imaging
- IPAT-compatible

Applications:

- Simultaneous visualization of the left and right TMJs (temporo mandibular joints) with high resolution
- High-resolution imaging of the eye
- High-resolution imaging of the wrist

Can be combined with:

- Body Matrix coils
- All flexible coils (e.g. CP Flex coil, small, CP Flex coil, large)
- CP Head Array Coil

- Weight: 1.1 kg

- Dimensions: 82 mm x 135 mm x 82 mm (L x W x H)

CP Head Array Coil

Dedicated coil for head examinations especially for utilization with a stereotactic frame for therapy planning and for pediatric imaging, e.g. of the spine.

- CP coil design with 2 integrated preamplifiers
- Upper coil part removable
- Lower coil part usable without upper part
- Open patient-friendly design
- Highly homogenous illumination
- no coil tuning
- Cushioned head stabilizers (removable)
- Detachable double mirror
- Also compatible with MAGNETOM Symphony and Sonata systems
- iPAT-compatible

Applications:

- Head examinations
- Head examinations with utilization of a stereotactic frame
- Pediatric imaging

Can be combined with:

- Body Matrix Coil
- All additional flexible coils (e.g. CP Flex coil, small, CP Flex coil, large)
- Double Loop Array Coil
- Weight: 5.7 kg
- Dimensions: 480 mm x 330 mm (L x W)

Quadrature Tx/Rx Head Coil

Transmit/receive coil for MR proton spectroscopic head examinations. It is especially suited for Chemical Shift Imaging (CSI) and Single Voxel Spectroscopy (SVS)

- CP Send/Receive head coil with integrated preamplifier

- Upper coil part removable
- Open patient-friendly design
- No coil tuning
- Cushioned head stabilizers (removable)
- Detachable double mirror (optional)

Applications:

- Head examinations
- High-resolution brain spectroscopy
- Weight: 7 kg
- Dimensions: 400 mm x 360 mm x 360 mm (L x W x H)

CP Flex Coil, large

CP Flex coils cover all extremities and joints; beneficial for pelvic, spinal and pediatric imaging; provide excellent soft tissue contrast, small field of views and thin slices, cost-effectively.

Applications:

- Imaging of large regions such as medium to large shoulder , hip and knee
- MR Angiography
- Abdominal imaging
- Brachial plexus

Applications:

- Imaging of large regions such as medium to large shoulder , hip and knee
- MR Angiography
- Abdominal imaging
- Brachial plexus

Description:

- Wrap- around coil made from soft and flexible material provides high patient comfort
- Excellent soft-tissue contrast and delineation of small details

- Coils are wrapped around the joint and are easily stabilized with Velcro straps

Integrated Panoramic Array™ (IPA)

- Optimizes coil positioning and virtually eliminates coil changing times
- Up to 4 different coils can be positioned and connected simultaneously
- Up to 16 CP coil elements can be selected simultaneously

Technical Data

- Circularly polarized Wrap around coil
- No coil tuning
- Requires Flex Coil Interface
- Size: 210mm x 520mm
- Circular polarized coil design offers up to 40% higher signal-to-noise than the linear design
- Integrated Panoramic Array

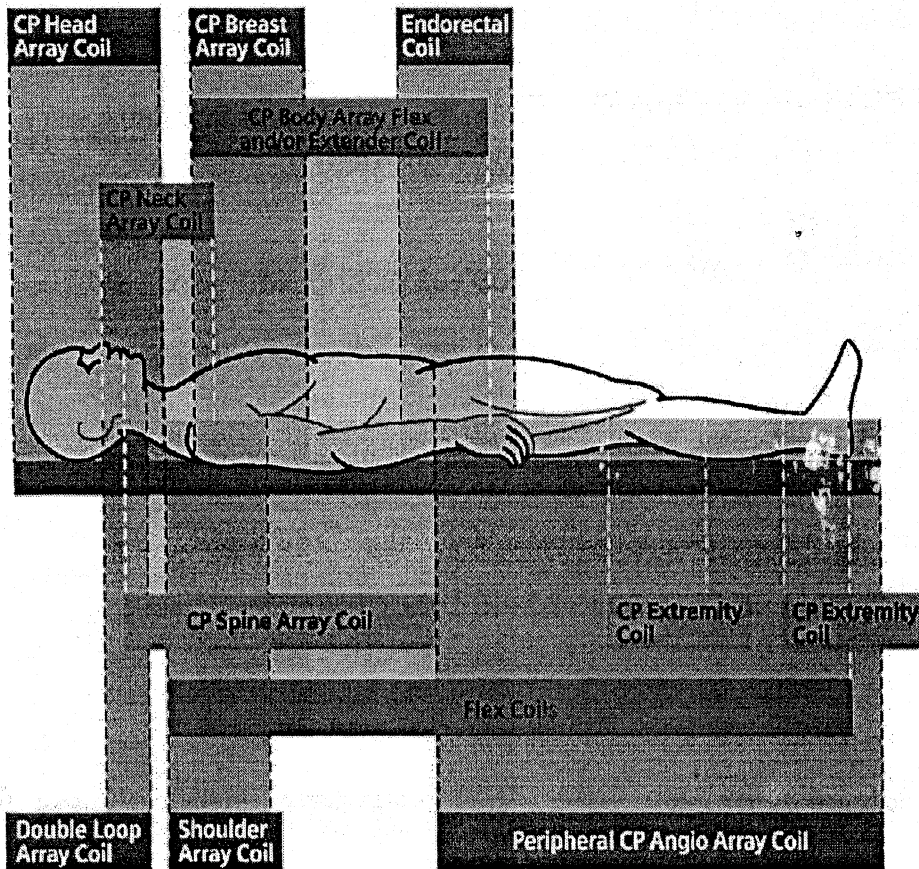
Main Benefit:

- Up to 4 different coils can be positioned and connected simultaneously
- Up to 16 CP coils from up to 4 coils can be selected simultaneously via software
- No coil changing with multi-exam studies
- CP Spine and CP Head Array Coil are integrated into the patient table top
- Remain on the table for approx. 95% of all examinations
- All IPA coils are time saving 'no tune' coils.
- Low-noise preamplifiers
-
- IPA Coil Combination

Up to 4 different coils can be positioned and connected simultaneously

Up to 16 coil elements from up to 4 IPA coils can be selected

simultaneously by soft buttons via ergonomic syngo MR user interface at the MR console



IPA Coil Combination Matrix MAGNETOM Sonata

Possible combinations
(not necessarily simultaneously)

1. 2nd Flex Coil Interface connectable
2. If CP Spine Array is not used

Combination of up to 4 IPA coils possible

Examples:

- Head+ Neck+ Spine
- Spine+ Body+ Flex
- Peripheral CP Angio + Body Array Flex + Body Array Extender + CP Spine incl. large FOV- Adapter
- Spine+ Body+ Endorectal

- etc.
-

Loop Flex Coil, large

Flexible linear polarized loop coil for shoulder and extremity examinations.

The coil provides excellent soft tissue contrast with small field of views

Applications:

- Examination of upper or lower extremities (e.g. shoulder, axilla, knee and ankle)
- High soft tissue contrast

Description:

- Flexible circular coil
- The design allows easy access to joints for patients with limited mobility

Technical Data

- Large, flexible circular coil
- No coil tuning
- Requires Flex Coil Interface
- Diameter: 190mm, 7.5in

Loop Flex Coil, small

Flexible linear polarized loop coil for examinations of very small structures near the surface. The coil provides excellent soft tissue contrast with small field of views.

Applications:

- Examination of small structures near the surface (e.g. joints of fingers and toes, wrist, skin, Temporo Mandibular Joints TMJ)
- High soft tissue contrast

Description:

- Small circular coil

- The Flex Loop Coil is placed directly on the joint and easily stabilized with Velcro straps

Technical Data

- Small, flexible circular coil
- Easy positioning
- No coil tuning
- Requires Flex Coil Interface
- Diameter: 40mm, 1.6in

Endorectal Coil and Interface

Transrectal surface coils for non-invasive evaluation of pelvic disease. High soft tissue contrast to show infiltration and extension of lesion. Improved treatment planning, evaluation of abnormalities, and staging of malignancies possible

Applications:

- Excellent visualization of the prostate, colon, rectum and cervix without the risk of invasive procedures
- Non- invasive preoperative diagnostic evaluation and treatment planning
- Accurate and reproducible image quality as the coil is positioned inside the patient next to the structure or lesion to be imaged

Description:

- Interface device for connecting either the prostate, colon or cervix receive coil
- The endo coil is plugged into the endo interface which is then plugged into the Flex Coil Interface
- Can be used simultaneously with the CP Spine Array and/ or CP Body Array Flex Coil

Integrated Panoramic Array™ (IPA)

- Optimizes coil positioning and virtually eliminates coil changing times
- Up to 4 different coils can be positioned and connected simultaneously
- Up to 16 CP coil elements can be selected simultaneously

Combination of Endorectal Coil, CP Body Array Flex and CP Spine Array Coil

- Non-invasive, high resolution imaging of prostate, rectum, colon and cervix

Technical Data

- Interface device to connect; prostate, colorectal, cervix coil probe
- The endo coil is plugged into the endo interface which is then plugged into the Flex Coil Interface.
- Requires Flex Coil Interface
- No coil tuning
- Field of view 8 to 20 cm
- Easily expandable with CP Body Array Flex and CP Spine Array Coil
- Additional probes have to be ordered from MEDRAD Inc. with the ordering name MR Innervu Disposable Prostate Endorectal Coil, Colon Endorectal Coil, Cervix Endorectal Coil

Console and Plasma Screen.

Operator console :

It is multifunctional with its own dedicated computer archive storage system and data base. It supports scan, display, archiving, filming, utilities and networking functions. Intercom control provides communication with the patient.

Plasma Screen:

When we touch the soft key on the screen, breaking up of an infrared beam which sends the commands to the system. The console is 8.5 x 17 inches in size, touch sensitive screen by which the operator can interact with the system. The screen is divided into two as scan screen and image processor(IP).

PALSMA SCREEN

15 inches diagonal monochrome monitor with 512 x 512 pixel display. Pulse Sequences available.

2D Optiopns:

SE, GRE, IR, TSE, FMPIR, FLASH, PISIF

3D Options :

FLASH, PISIF, GRE.CISS,MP-RAGE

Vascular Options :

2D & 3D PC,

2D & 3D TOF,

3D Vascular TOF SPGR (MOTSA).

Image enhancement Optons.

Flow compensation (FC), respiratory compensation (RC), ECG & Pulse gating, graphic Rx (GRx), Phase over sampling (POS), fat & water suppression (stF/W), phase offset multi planar (POMP), variable band width (VBW), image intensity correction (IIC)< extended dynamic range (EDR), contiguous section (CS), classic (CL), square pixel (SqP), sequential (Sq), magnetisation transfer (MT), multi phase ramped pulse direction (SPF).

Contiguous slices :

Reduces chemical shift artifacts, reduces cross talk and provides flow compensation option with slice thickness below 3mm.

Extended dynamic range:

Allows 32-bit data processing, instead of the conventional 16 bits, to improve SNR, in some applications. It is compatible with virtually all pulse sequences and imaging options.

Fast :

Turbo SE to minimize the motion artifact in body imaging and fast cardiac gating to acquire multiple phases cardiac cycle at single slice location. This option provides high-resolution images with short scan times depending upon the echo train length.

Fat/Water suppression :

Suppress signal from fat or water. Uses a frequency – selective saturation pulse to suppress signal within the imaging volume. This will reduce the chemical shift artifact .

Flow compensation :

Minimize motion artifacts created by the flow of slow moving blood and csf. This will cause strain on the gradient hardware.

Fractional Echo (Short Echo)

Allows more slices in single echo scan and increase T1WI, decreases magnetic susceptibility and decreases flow artifact.

Average (Number of excitations)

NEX gives the indication of how many times signals is averaged before doing the fourier transformation. Fractional nex allow to reduce imaging time but with decrease in SNR.

Classic (CL)

Increases the variance between muscle and fat signals. Classic images are annotated CL.

Driven Equilibrium Preparation (DE prepped)

A fast grad option that uses 90/180/90- degree pulse series where the longer the time between the 180 -degree pulse and the last 90-degree pulse, the more T2 weighted is the image.

Graphic Rx (GRx)

Prescribe the imaging locations by graphically.

Image intensity correction

Activates a filtering process to minimize excessive brightness of anatomy close to surface coil. IIC minimize the fat signal when using extremity coil. This is equivalent to the TGC of ultra sound imaging. IIC increase the image reconstruction time by adding 1.7 seconds to the first echo display and 1 second to subsequent echoes.

Magnetization Transfer (MT).

Improves contrast by saturating the short T2 component of tissue, such as grey white matter and skeletal muscle. Magnetization enhances inferior to superior flow.

Phase Over Sampling

Prevent wrap around(aliasing) artifacts and in small FOVs. Phase wrap takes "close-up", high resolution images of FOVs smaller than available anatomy. NPW doubles the FOV and image acquisition matrix in phase direction and during data reconstruction , discards data from outside the FOV.

Phase Offset Multi-Planar (POMP).

Acquires twice the number of allowable slices in a given TR without increasing scan time. Designed for rapid coverage of anatomy including the head, c-spine, and L-spine, particularly T2 axis. POMP excites two adjacent slices at once, with a pulse combining two frequency bands and reconstructs as two separate images.

Respiratory Compensation (RC).

It reduces the ghosting artifacts from respiratory motion in chest or abdomen and reduces motion caused by respiration in flow analysis. RC uses chest wall motion to identify the best time for motion reduced imaging. The bellow is positioned exactly over the imaging part to get better triggering.

Saturation parameters (SAT).

This is addition RF pulse that chemically or spatially saturate specified structures. Sat pulses helps to minimize flow artifacts, reduce residual signal, suppress phase encoding ghosts caused by flow related enhancement.

Sat Concatenated.

In conventional studies, moves slice select pulses along with the acquisitions and saturating tissues before acquiring each slice.

Swapping of phase and frequency.

This allow to change the phase and frequency direction of an acquisition.

Variable Bandwidth (VB).

Allows the operator to select the receiving frequency range of the signal. Decrease band increases SNR, decrease the number of imaging slice.

Sequential (SQ).

Acquires all the images at the first slice before moving to the next location. With sequential, total number of acquisition equals to the number of slices.

Square Pixel(SqP).

Permits scans with an FOV scaled by the ratio of phase to frequency points in the phase direction, resulting in a square pixel resolution at a lower scan time.

Slice thickness, Spacing, FOV and Excitation.

Slices :

2D : 3 – 20 mm in 0.7 increments.

3D : 0.7 – 5 mm with volume of 28, 60 and 124, contiguous slices.

Spacing : Can be varied with an increment of 0.0mm.

FOV :

Head coil : 8 – 28 cm in 1cm increments.

Body coil : 8 – 48 cm in 1cm increments.

(Variable FOV and off centre FOV are available.)

Extremity coil : 8 – 18 cm.

QuardT/L coil : 30cm. S/I direction.

Post neck coil : 24cm. S/I direction.

NEX :

Fractional NEXs : 0.5, 0.75, and 1.5 (1.5 NEX is only available with NPW and POMP & 0.5 NEX is not available with GRE, SPGR, SSFP and Fast sequences.)

Multiple NEXs : 1 – 150.

Acquisition matrix.

Frequency Matrix : 256 and 512.

Phase Matrix : 128 – 512 in steps of 32.

Need of pre Scanning :

Optimizes the image quality by providing shim accuracy (Ensure precise flip angle, optimize receiver dynamic range).

Tune the frequency you are interested in, by adjusting the centre of the frequency. It ensure the optimal excitation of proton.

Sets the volume / slice at just right level by fine tuning the transmit and receive gain or amplification.

Oxygen Monitoring :

Oxygen monitor gives alarm when oxygen level falls below 18%.

Working : Oxygen gas in the space surrounding the cell (oxygen detection transducer cell) diffuses through a Teflon membrane and reaches on the surface of the cathode, corresponding oxidation occurs at the anode. Electrical current generated is proportional to the oxygen. Cell output is limited by the rate at which the oxygen enters the cell and by the amount of anode material stored within the cell. The resulting cell output drives the meter circuit of the oxygen monitor.

Operating temperature : 32 – 131 °F.

Cell Electrolyte used : Potassium Hydroxide in a barrier bag.

Quench.

Unexpected loss of superconductivity in a superconducting magnet that causes heating and very rapid vaporization of the cryogen such as liquid Helium. This can cause damage to the magnet and can force oxygen out of the scanner room potentially causing anoxic condition.

Cold Head and Compressor (Chiller).

Cold head use simple compression-expansion process. Helium is used for compression and compressed heat is removed by air cooled heat exchanger and He gas is allow to expand. The flow of low and high pressure He is controlled by control valve located at the base of cold head. The control valve is rotated clockwise by 72 rpm by an electronic synchronized stepping motor. As the control valve rotates on the top of the control disc, Helium in the valve align with different holes in the disc. When one set of holes are aligned, high pressure He flows in to the expansion chamber. As the valve rotates this valve is sealed and another set of holes align, this second set aligned holes forms a channel which allows the He in the chamber to expand, cool and escape. The volume of expansion chamber is also controlled by another control valve. A relief valve pressure from cold head if it is exposed to hot ambient temperature when shut off. This relief valve also protect the cold head from internal damage if the flux lines are removed while the helium inside the cold head is expanding. The helium connection have self sealing coupling which keep the cold head under pressure and gas line is removed.

Cold head is a two stage cryo-refrigerator. Its first stage produces a temp. 40 – 60K and its second stage produces a temp. between 10 – 20 K. The available cold head capacity is 40 Watt for first stage and 2 watt for second stage. The cold head require compressor (chiller) to operate. The compressor provides power and high pressure helium need to produce low temp. in the cold head. Helium gas circulated between cold head and compressor through flux lines. Cold head is started and stopped by start and stop button situated in front of compressor. Compressor have a guage to monitor the helium pressure. The pressure should be between 305 – 334 psi (21 – 23bar) during operation and 218 – 232psi (15 – 16bar) when compressor is not operated. Cooling water temperature is between 40 – 80°F and circulating with a rate of 3.5 litter per minute. The pH of cooling water is between 6 – 8.

* * * * *

Brain Routine Coil

Rf Coil-Quad Head

Imaging Parameters	Views				
	SAGITT AL. Loc	Axial T1	Axail T2	Coronal T1	Coronal T2
Pulse Seqs	SE	SE	SE/V	SE	SE/V
PL	SAGITA L	OBLIQU E	OBLIQU E	OBLIQU E	OBLIQU E
MD	2D	2D	2D	2D	2D
PSD	-	-	-	-	-

Options	NONE	GRx	VB, EDR, FC,GRx	GRx	VB,EDR, FC, GRx
TR	380	450	2000	450	2000
TE	16	16	30/90	16	30/90
TI	-	-	-	-	-
FLIP (°)	-	-	-	-	-
FOV	24	22	22	22	22
THICKNES S	5	5	5	5	5
INTER SPACE	3	3	3	3	3
LOCATION S	13	16	16/16	16	16/16
MATRIX	256X192	256X192	256X192	256X192	256X192
NEX	2	2	1	2	1
FREQUENC Y	S/I	NOT SWAPP ED	NOT SWAPP ED	NOT SWAPP ED	NOT SWAPP ED
RBW/ RBW2	16	16	16/3.20	16	16/3.20
TIME	2.29	2.10	5.12	2.10	5.12

Brain Pituitary**Rf Coil-Quad Head Coil**

IMAGING PARAMET ERS	VIEWS				
	Sag. Loc	Axial T1	Axial T2	Coronal T1 Thin	Axial T1 Thin
Pulse Seqs	SE	SE	SE/V	SE	SE
PL	SAGITA L	OBLIQU E	OBLIQU E	OBLIQU E	OBLIQU E
MD	2D	2D	2D	2D	2D
PSD	-	-	-	-	-

Options	NONE	GRx	VB, EDR, FC,GRx	CS,GRx	CS, GRx
TR	360	480	2000	300	300
TE	Min.Full	Min.Full	30/90	Min.Full	Min.Full
TI	-	-	-	-	-
FLIP (°)	-	-	-	-	-
FOV	24X24	22X16	22X16	22X16	22X16
THICKNES S	5	5	5	3	3
INTER SPACE	3	3	3	0	0
LOCATION S	13	16	16/16	10	10
MATRIX	256X19 2	256X19 2	256X19 2	256X19 2	256X19 2
NEX	2	2	1	2	2
FREQUEN CY	S/I	NOT SWAPP ED	NOT SWAPP ED	NOT SWAPP ED	NOT SWAPP ED
RBW / RBW2	16	16	16/3.20	16	16

Brain Epilepsy**Rf Coil-Quad Head Coil**

IMAGIN G PARAM ETERS	Views						
	Axial Loc	Coron al Loc	Sag.T1	Axial T1	Axial T2	Coron al T2	Coron al T1
Pulse Seqs	SE	SE	SE	SE	SE/V	SE	FLAS H 3D

PL	AXIAL	CORONAL	OBLIQUE	OBLIQUE	OBLIQUE	OBLIQUE	CORONAL
MD	2D	2D	2D	2D	2D	2D	3D
PSD	-	-	-	-	-	-	-
Options	NON E	GRx	GRx	GRx	VB, EDR, FC, GRx	VB, ED RFC, GRx	FC, GRx, NPW
TR	300	300	400	480	2000	2000	36
TE	Min.Full	Min.Full	Min.Full	Min.Full	30/90	30/90	Min.Full
FOV		-	-	-	-	-	45
THICKNESS	22X16	22X16	24X24	22X16	22X16	22X16	16X16
INTER SPACE	5	5	5	5	5	4	1.5
LOCATIONS	3	5	3	3	3	2	0
MATRIX	10	10	13	16	16/16	16/16	60
NEX	256X192	256X192	256X192	256X192	256X192	256X192	256X128
FREQUENCY	0.75	0.75	2	2	1	1	2
RBW / RBW2	A/P	S/I	NOT SWAPPED	NOT SWAPPED	NOT SWAPPED	NOT SWAPPED	S/I
TIME	16	16	16	16	16/3.20	16/3.20	16
	0.38	0.38	2.37	2.22	5.12	5.12	9.53

C-Spine

Rf Coil-Post Neck Coil

IMAGING PARAMETERS	Views					
	Coronal Loc	Sag.T1	Sag.T2*	Axial T1	Axial T2*	Axial 3D Gradient
Pulse Seqs	SE	SE	GRE	SE	GRE	GRE
PL	CORONAL	SAGGITAL/OBLIQUE	SAGGITAL/OBLIQUE	AXIAL/OBLIQUE	AXIAL/OBLIQUE	AXIAL
MD	2D	2D	2D	2D	2D	3D
PSD	-	-	-	-	-	-
Options	NPW	CS,GRx	FC,GRx	CS,GRx	FC,GRx	FC,GRx
TR	450	320	300	380	340	33
TE	Min.Full	16	Min.Full	19	Min.Full	Min Full
FLIP (°)		-	12	-	16	9
FOV	32X32	24X18	24X18	18X13	18X13	18X13
HICKNESSES	5	4	4	5	5	1.5
INTER SPACE	3	1	1	0	0	0
LOCATIONS	11	9	9	11	11	28

MATRIX	256X1 28	256X1 92	256X1 92	256X1 92	256X1 92	256X19 2
NEX	1	4	4	4	4	2
FREQUENCY	S/I	S/I	S/I	NOT SWAP PED	NOT SWAP PED	R/L
RBW / RBW2 (KHz)	32	16	16	16	16	16
TIME	1.10	3.08	2.56	3.42	3.19	5.08

THORACIC SPINE

Rf Coil-Quad T/L Coil

Patient lies supine on the examination couch with the coil extending from the top of the shoulders to the lower costal margin to ensure the total coverage of the total thoracic spine and conus (Q/D, T/L coil). The patient is positioned so that the longitudinal alignment light lies in the midline and the horizontal alignment light passes through the level of the thoracic vertebra

Sagittal T2 localizer

Slices are prescribed from on either side of the longitudinal alignment light and left borders of the vertebrae. The area from the base of the skull to the conus is included (Body coil). As the first cervical vertebrae is visualized, the offset allows easy assessment of vertebral levels and easy placement of pre saturation pulses

Coronal T2 Localizer

Slices are prescribed relative to the vertical alignment light, from posterior aspect of spinous process to anterior border of the vertebral bodies. The area from C7- conus is included

Sagittal T1

Slices are prescribed from right to left borders of the vertebrae and paravertebral areas if needed. The area from C7-conua is included

Additional sequences

Post contrast sag. T1& Axial T1 fat sat (If needed –bone and
cord tumors, infectious diseases M S plaques , metastatic diseases)
STIR – For metastatic diseases of vertebral bodies

LUMBO SACRAL SPINE

Rf Coil-Quad T/L Coil

Patient lies supine on the examination couch with their knees flexed over a foam pad, for comfort and to flatten the lumbar curve so that spine lies nearer to the coil. The coil (Q/D /T/L) Should extend from the xiphisternum to the bottom of the sacrum for adequate coverage of the lumbar spine. The patient lies so that the longitudinal alignment light lies in the midline and the horizontal alignment light passes through the lower costal margin

ROUTINE SEQUENCES INCLUDE

1. Coronal T1 Localizer

Slices are prescribed from the posterior aspect of the vertebral column to the anterior aspect of the vertebral bodies. The area from the conus to the sacrum is included

2. &3 Sagittal Spin Echo T1 &T2

The slices are prescribed from right to left borders of the vertebral bodies (If needed paravertebral area is included)

4. &5 Axial Spin Echo T1 & Axial T2 TSE

Slices are prescribed parallel to the conus, vertebral bodies, and disc spaces Continuous multiple slices are obtained at L4-L5 & L5-S1

Additional sequences

Post contrast Sagittal.Axial & Coronal T1 fat sat . (if indicated vertebral body or cord tumours, fragment discs, post operative IVDP, metastatic disease etc.

3-D Gradient Sagittal & Axial across the level of L3-S1
coronal SE/FSE T1 for some cord syndrome

LUMBOSACRAL SPINE

IMAGING PARAMETERS	Views					
DIANE	Coronal Loc	Sag.T1	Sag.T2 TSE	Axial T1	Axial T2TSE	Axial 3D Gradient
Pulse Seqs	SE	SE	SE	SE	SE	FLASH
PL	CORONAL	SAGGITAL/OBLIQUE	SAGGITAL/OBLIQUE	AXIAL/OBLIQUE	AXIAL/OBLIQUE	AXIAL
MD	2D	2D	2D	2D	2D	3D
PSD	-	-	-	-	-	-
Options	None	RC,CS,GRx	RC,FC,VB GRx/ED	RC,NP W GRx	RC/FC /FAST/ VB/ED	FC,NP W, GRx
TR	400	300	3000	400	3000	38
TE	Min.Full	16	108	18	108	Min.Full
TI		-	-	-	-	-
FLIP (°)		-		-		
FOV	32X32	28X28	28X18	18X18	18X18	18X18
THICKNESSES	5	5	5	5	5	2
INTER SPACE	3	0	0	5	5	-
LOCATIONS	11	9	9	11	11	28
MATRIX	256X192	256X192	256X192	256X192	256X192	256X192
NEX	0.75	4	4	4	4	1.5
FREQUENCY	S/I	A/P	S/I	SWAPPED	SWAPPED	A/P
RBW / RBW2 (KHz)	16	16	16	16	16	16

Hip Joint
Rf Coil- Body Coil

IMAGING PARAMET ERS	Views					
	Coron al Loc	Axial T1	Axial T2	Coronal T1	Coronal T2	Sag.T1
Pulse Seqs	SE	SE	GRE	SE	GRE	SE
PLANE	CORO NAL	AXIA L/OB LIQU E	AXIAL/ OBLIQ UE	CORON AL /OBLIQ UE	CORON AL /OBLIQ UE	SAGITAL / OBLIQ UE
MODE	2D	2D	2D	2D	2D	2D
PSD	-	-	-	-	-	-
Options	None	CS, GRx, NPW	FC, GRx, NPW	CS,GRx	FC,GRx	CS, GRx
TR	400	420	350	360	250	350
TE	16	16	16	18	Min.Full	Min.Full
TI	-	-	-	-	-	-
FLIP (⁰)	-	-	15	-	15	-
FOV	42	42	42	40	40	38
THICKNES S	5	5	5	5	5	5
INTER SPACE	3	1	1	1	1	1.5
LOCATION S	14	14	14	11	10	11
MATRIX	256X1 92	256X 192	256X1 92	256X19 2	256X19 2	256X192
NEX	0.75	1	2	1	2	1

FREQUENCY	S/I	R/L	R/L	NOT SWAPPED	NOT SWAPPED	NOT SWAPPED
RBW / RBW2 (KHz)	16	16	16	16	16	16
TIME	1.04	1.34	1.30	1.16	1.36	1.11

KNEE JOINT

Rf Coil - Extremity Coil

The patient lies supine with the knee relaxed, slightly flexed position within the coil (Extr. Knee coil). The knee is immobilized with pads. The patient is positioned so that the longitudinal alignment light lies along the midline of the leg under examination. The horizontal light passes through the center of the coil which corresponds to the lower border of the patella

1. Axial T2*GRE-Localizer

Acts as a localizer. If the knee is not at iso center the FOV is offset so that the knee is in the middle of the image. Coronal and Sagittal localizers may also be used

With the axial localizer the slice in which the patella is clearly demonstrated is chosen to prescribe the following sequences

2.&3 Coronal SE T1&TSE T2/PD

Slices are prescribed from the femoral condyles posteriorly to the anterior patella and oriented parallel to the posterior surface of the femoral condyles

4 &5 Sagittal SE T1&TSE T2/ PD (fat sat)

Slices are prescribed from the lateral side of the knee

6 Oblique sag T1

Clear display of the anterior Cruciate ligament is essential for trauma of suspected joint damage . The ligament is best seen in oblique sagittal scans oriented to the ligament .Slices are prescribed from the lateral to the medial collateral ligament and aligned parallel to the anterior Cruciate ligament which runs at an angle 5-10°

Additional sequences

7. Axial GRE T2*

KNEE JOINT

KNEE COIL

IMAGING PARAMETERS	Views						
	Axial Loc	Coronal T2*	Cor.T1	Cor. T2-FSE/S TIR	Sag.GRE	Sag.T 1	Sag.T 2
Pulse Seqs	SE	GRE	SE	SE/FS E	GRE	SE	SE
PLANE	AXIAL	CORONAL /OBLIQUE	CORONAL /OBLIQUE	OBLIQUE	SAG/OBLIQUE	OBLIQUE	SAGITTAL /OBLIQUE
MODE	2D	2D	2D	2D	2D	2D	2D
SAT	-	-	-	FAT-FSE	-	-	-
Options	NPW, CS	NPW,F CGRx,	NPW, CS, GRx,	VB,ED R, NPQ,GRx	NPW, GRx	NPW, CS, GRx	FC, EDR, VB,NPW,GRx,CS
TR	540	440	500	2000/4500	440	640	2000

TE	18	19	18	Min.Ful I/ 18	10	18	30/90
TI	-	-	-	170- STIR	-	-	-
FLIP (°)	-	22	-	-	20	-	-
FOV	16	16	16	16	16	16	16
THICKN ESS	5	5	5	5	5	3	5
INTER SPACE	0	0	0	0	0	0	0
LOCATI ONS	15	14	14	14	20	10	16
MATRI X	256X 224	256X1 92	256X2 56	256X2 24	256X19 2	256X1 92	256X2 56
NEX	1	1.5	1	1	1	1	1
FREQU ENCY	R/L	NOT SWAP PED	NOT SWAP PED	NOT SWAP PED	NOT SWAPP ED	NOT SWAP PED	R/L
RBW / RBW2 (KHz)	16	16	16	16	16	16	16
TIME	2.17	2.53	2.24	5.12/2. 44	2.10	2.22	5.2

**For Cruciate ligament : Obl/Gre – 20o – 300/19 – 16x16 – 3/0-:
1.5Nex =>**

HEART

Rf Coil - Body Coil

Patient positioning

The patient lies supine on the couch with respiratory compensation bellows and ECG gating leads attached . The patient is positioned so that the longitudinal alignment light lies in the midline and the horizontal alignment light passes through the 4th Thoracic vertebra

CARDIAC GATING

Cardiac gating uses the electrical signal detected by the leads placed on to the patient's chest to trigger each RF excitation pulse . In this way each image is always acquired at the same phase of the cardiac cycle so phase mismapping from cardiac motion is reduced . The correct placement of leads is very important to optimize the image quality

Lead placement

There are usually four leads that are color coded for easy use. Leads can be placed either anteriorly or posteriorly. Usually anterior placement is done it is simpler and easy to assess the landmark

The white and red leads are placed across the heart as the voltage difference between the two produces the ECG trace

The green lead is positioned as close, but not touching the red it acting as the ground

The black lead is placed right upper chest below the clavicle

ROUTINE SEQUENCES

Coronal Spin Echo T1 localizer

Slices are prescribed relative to the vertical alignment light from the posterior chest muscle to the sternum

2,3,4 Axial spin echo T1, T2/PD&Oblique

Slices prescribed from inferior border of the heart to the arch of the aorta

5. Sagittal oblique SE T1

Select an image that demonstrate ascending and descending aorta. Slices are angled through the ascending aorta and prescribed from one lateral edge of the vessel to the other

Additional sequences include

Oblique cine GRE T2*

Used to assess cardiac function. Two chambers are best demonstrated on sagittal/ oblique view, four chambers in the coronal/ oblique plane.

HEART

IMAGING PARAMETERS	Views			
	Coronal T1	Oblique Axial	Axial T1	Sagittal T1
Pulse Seqs	SE	SE	SE	SE
PLANE	CORONAL	OBLIQUE	AXIAL / OBLIQUE	SAGITTAL
MODE	2D	2D	2D	2D
SAT	-	S/I & A/P	S/I & A/P	R/L & A/P
Options	RC, Gating	CS, NPW, EDR, VB GRx, RC, Gating	RC, Gating, NPW CS, GRx,	RC, Gating. CS, GRx NPW
TR	1 RR	1 RR	1 RR	1RR
TE	Min.Full	20	Minimum	Minimum
TI	-	-	-	-
FLIP (⁰)	-	-	-	-
FOV	36	36	36	36
THICKNESSES	10	7	10	10
INTER SPACE	3	3	2.5	3
LOCATIONS	23	11	12	12

MATRIX	256X128	256X128	256X128	256X128
NEX	4	4	4	4
FREQUENCY	S/I	NOT SWAPPED	R/L	S/I
RBW / RBW2 (KHz)	16	16	16	16
TIME	8.54	8.54	8.54	8.54

Abdomen
Rf Coil - Body Coil

IMAGING PARAMETERS	Views					
	Coronal Loc	Axial T1	Axial T2	Sagittal T1	Coronal T1	Axial Breath Hold
Pulse Seqs	SE	SE	SE	SE	SE	FLASH 3D
PLANE	CORONAL	AXIAL / OBLIQUE	AXIAL / OBLIQUE	SAGITTAL / OBLIQUE	CORONAL / OBLIQUE	AXIAL
MODE	2D	2D	2D	2D	2D	2D
PSD	MEMP	MEM P	MEMP	MEMP	MEMP	MEMP

Options	NoneS Q	NPW, RC, GRx,	FC, GRx, EDR,R C,NPW VB	RC,GR x	RC,G Rx	SEQ, RC, FC, GRx
TR	400	400	2000	400	400	30
TE	18	16	30/90	18	18	Min.Fu ll
TI	-	-	-	-	-	-
FLIP (°)	-	-	-	-	-	45
FOV	40	42	42	38	40	42
THICKNE SS	5	10	10	10	10	5
INTER SPACE	3	5	5	5	5	5
LOCATIO NS	14	13	16	12	12	1
MATRIX	256X19 2	256X1 92	256X19 2	256X19 2	256X1 92	256X2 56
NEX	0.75	4	1	4	4	1
FREQUE NCY	S/I	R/L	R/L	S/I	S/I	R/L
IMAGING PARAMETE RS	Views					
	2D PC			3D PC		
Pulse Seqs	Vascular PC			Vascular PC		
PLANE	SAGITTAL , AXIAL , CORONAL			SAGITTAL , AXIAL , CORONAL		

TIME	1.51	13.32
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Brain : MRA

RF Coil - Head Coil

IMAGING PARAMETERS	Views		
	2D TOF	3D TOF	3D MOTSA
Pulse Seqs	Vascular TOF SPGR	Vascular TOF	Vascular TOF SPGR
PLANE	SAGITTAL , AXIAL , CORONAL	SAGITTAL , AXIAL , CORONAL	AXIAL
MODE	2D	3D	3D Multiple Slab
Vascular Options			
1. Projection Images	Collapse -19	Collapse -19	Collapse -19
2.Ramp Pulse	-	S→I or I→S	S→I or I→S
Options	FC,GRx, SEQ	FC,GRx,	FC, SQP, EDR, VB, GRx,MT
TR	46	33	56
TE	Min.Full	6.9	6.4
TI	-	-	-

MODE	2D	3D
	Vascular Options	
1. Projection Images	Collapse -2	Collapse -19
2. Acquisition Flow Direction	All	All
3. Additional Images	Magnitude	Magnitude
4. Flow Reconstruction Time	Complex Difference	Phase Difference
Velocity	10-70cm/sec	10-70cm/sec
Options	FC, GRx, SEQ	FC, GRx,
TR	24	33
TE	8.6	6.9
TI	-	-
FLIP ($^{\circ}$)	30	30
FOV	22X22	22X16
THICKNESS	80	1.5
INTER SPACE	0	0
LOCATIONS	1	60
MATRIX	256X192	256X128
NEX	8	1
FREQUENCY	S/I	A/P
RBW / RBW2 (KHz)	16	16

FLIP (°)	40	30	30
FOV	22X22	22X16	22X16
THICKNESS	1.5	1.5	1.5
LOCATIONS/SLAB Overlap	-	-	32 33%
LOCATIONS	120	60	SLAB : 3
MATRIX	256X128	256X128	256X192
NEX	2	2	1
FREQUENCY	R/L	A/P	A/P
RBW / RBW2 (KHz)	16	16	16
TIME	9.02	9.04	15.43

Arterial & Venous flow can be obtained by giving appropriate Sat. Pulse (S or I)

Aorta : Gd enhanced MRA

RF Coil - Body Coil

IMAGING PARAMETERS	Views		
	Coronal T1 Localiser	Sagittal 40ml Gd	Coronal 40ml Gd
Pulse Seqs	SE	Vascular TOF SPGR	Vascular TOF SPGR
PLANE	CORONAL	SAGITTAL	CORONAL
MODE	2D	3D	3D

Vascular Options	-	Ramp Pulse: R/L	Ramp Pulse: R/L
Options	RC, Gating, GRx	GRx,	GRx
TR	-	27	27
TE	Min.Full	Min.Full	Min.Full
Gating	ECG/PULSE	-	-
FLIP (°)	-	30	30
FOV	40	36	36
THICKNESS	10	2.5	2.5
Inter Space	3	0	0
LOCATIONS	23	60 +19 Collapsed	60 +19 Collapsed
MATRIX	256X128	256X192	256X192
NEX	4	1	1
FREQUENCY	S/I	S/I	S/I
RBW / RBW2 (KHz)	16	16	16
TIME	8.54	5.35	5.35

Spinal Angiogram

RF Coil - Body Coil

PC MRA is the technique of choice in spine is done with different flow velocities

IMAGING PARAMETERS	Views	
	2D PC	2D PC
Pulse Seqs	Vascular PC	Vascular PC
PLANE	CORONAL	SAGITTAL
MODE	2D	2D
Vascular Options		
1. Projection Images	Collapse	Collapse
2. Acquisition Flow Direction	All	All
3. Additional Images	Magnitude	Magnitude
4. Flow Reconstruction Time	Complex Difference	Complex Difference
5. Velocity	6-20 cm/sec	6-20/sec
Options	FC, GRx, NPW	FC, GRx, NPW
TR	70	70
TE	11.20	11.20
TI	-	-
FLIP (°)	30	30
FOV	30	30
THICKNESS	15	30
INTER SPACE	0	0
LOCATIONS	1	1
MATRIX	256X192	256X192

NEX	10	10
FREQUENC Y	S/I	S/I
RBW / RBW2 (KHz)	16	16
TIME	8.58	8.58

Few Vol. Of IV. Gadolinium contrast 2-4ml is given prior to Angio.

MR Myelogram

RF Coil – Quad T/L Coil

Heavily weighted FSE T2

IMAGING PARAMETER S	Views	
	SAGITTAL	CORONAL
Pulse Seqs	FSE	FSE
PLANE	SAGITTAL / OBLIQUE	CORONAL / OBLIQUE
MODE	2D	2D
1. Projection Images	Collapse	Collapse
2.Acquisition Flow Direction	All	All
3. Additional Images	Magnitude	Magnitude
4.Flow Recon- struction Time	Complex Difference	Complex Difference

Options	FAST,FC,GRx,NPW,EDR,VD	FAST,FC,GRx,NPW,EDR,VD
User's CV : FSE Optimisation	0	0
TR	10000	10000
Eff.TE	300	300
ETL	16	16
FOV	26 : Sat - A,P,F	26 : Sat - A,P,F
THICKNESS	3	3
INTER SPACE	1	1
LOCATIONS	18	18
MATRIX	256X192	256X192
NEX	2	2
FREQUENCY	NOT SWAPPED	NOT SWAPPED
RBW / RBW2 (KHz)	16	16
TIME	4	4

MR Cholangiogram

RF Coil - Body Coil

IMAGING PARAMET ERS	Views
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	Axial Localiser	Coronal Localiser	Coronal	Coronal/ Oblique
Pulse Seqs	SPGR/BH	SPGR/BH	FSE/BH	FSE/BH
PLANE	Axial	Coronal	CORONAL	Coronal / Oblique
MODE	2D	2D	2D	2D
Options	RC, FC,EDR, SEQ,Gat, GRx	RC, FC,EDR, SEQ,Gat,G Rx	FAST,FC, NPWEDR, VB,GRx	FAST,FC,N PWEDR,V B,GRx
TR	35	35	4000	9000
Effect.TE	Minimum	Minimum	750	340
FOV	40	40	24	24
THICKNE SS	10	10	20	3
Inter Space	2.5	2.5	1	0
LOCATIO NS	1	1	1	13
MATRIX	256X192	256X192	256X128	256X192
NEX	1	1	1	1
FREQUEN CY	R/L	S/I	NOT SWAPPED	NOT SWAPPED
RBW / RBW2 (KHz)	16	16	16	32
TIME	0.38	0.38	0.16	1.30

Flow Compensation direction selected according to the plane of acquisition,

So : Axial – Slice & Coronal / Sagittal – Frequency.

Steriotactic – For Globus Pallidus Internanal capsule

Head RF Coil Coil

IMAGING PARAMET ERS	Views				
	Sag Localis er	Axial Localise r	Coronal Localise r	Axial FMPIR	Coronal FMPIR
Pulse Seqs	SE	SE	SE	IR	IR
PLANE	SAGITT AL	AXIAL	CORON AL	AXIAL	CORON AL
MODE	2D	2D	2D	2D	2D
Options	None	None	None	FAST,F C, SEQ,S QP EDR,VB GRx	FAST,FC , SEQ,SQ P EDR,VB, GRx
TR	420	420	420	4000	4000
TE/Effect. TE	Minimu m	Minimum	Minimum	38	38
FOV	30	30	30	30	30

THICKNESS	5	5	5	3	3
Inter Space	3	3	3	0	0
LOCATIONS	11	11	11	12	12
MATRIX	256X256	256X256	256X192	256x256	256x256
NEX	2	2	2	4	4
FREQUENCY	S/I	A/P	S/I	A/P	S/I
RBW / RBW2 (KHz)	16	16	16	16	16
TIME	2.45	2.45	2.45	14.08	14.08

MISCELLANEOUS

IMAGING PARAMETERS	Views					
	Internal Ear	Urography	FLAIR	STIR	HEAD	SPINE
Pulse Seqs	FSE	FSE	FMPIR	FMPIR	FSE	FSE

PLANE	CORL. OBL/ AXIAL. OBL	CORL. OBL	OBLIQ UE	OBLIQU E	OBLIQU E	OBLIQU E
MODE	2D	2D	2D	2D	2D	2D
TI	-	-	2500	155	-	-
Options	FC, GRx, EDR,,N PW, SQ, VB	VB,EDR ,NPW,G Rx,StF	VB,ED R,GRx, FC	VB,EDR ,GRx,F C	VB,EDR , FC,GRx	VB,EDR , FC,GRx ,StF,a:S at
TR	6000	10,000	10,000	5,000	3,000	3,000
Eff.TE	360	340	34	Min.Full	17/34	68
ETL	12	32	12	8	8	8
FLIP (°)	-	-	-	-	-	-
FOV	16X16	36	22X16	22X16	22x16	-
THICKN ESS	5/20	3	5	5	5	5
INTER SPACE	0	0	3	3	3	1
LOCATI ONS	6/1	11	9	16	16	13
MATRIX	256X25 6	256X19 2	256X1 92	256X19 2	256X19 2	256X19 2
NEX	2	2	1	1	1	2
FREQUE NCY	NOT SWAPP ED	NOT SWAPP ED	NOT SWAP PED	NOT SWAPP ED	NOT SWAPP ED	NOT SWAPP ED
RBW / RBW2 (KHz)	16	16	16	16	16/3.2	16

TIME	4.48	2	6	3	2	2
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PELVIS ROUTINE

Patient positioning

The patient lies supine and positioned so that the longitudinal alignment light lies in the midline and the horizontal light passes through the point midway between the pubic symphysis and iliac crests. When local rectal coil is used it is carefully inserted and positioned prior to the investigation

1. Coronal T1-Localizer

Thick slices/spacing are prescribed from coccyx to the anterior aspect of the pubic symphysis

2&3 Sagittal SE T1&T2 FSE

Thick slices are prescribed from left to right pelvic side walls. Unless lymph node involvement is suspected, small structures such as the prostate require high resolution imaging using thin slices/ gaps needed

4&5 Axial SE T1 & T2 FSE

Slices are prescribed from the pelvic floor to the iliac crests

PSD	Options	TR	TE	ETL	F O V	Thick	S P	Loc	Matrix	Nex	RBW	Time
1SE	RC/NP	440	18		32	5	1	14	256X 192	1	16	1.5
2.SE	RC/NPW	580	18		28	4	1	20	"	2	16	3.4
3.SE	Fast/FC VB/ED NPW	3000	120	6	28	4	1	20	"	2	16	2.5
4.SE	CS/NP W	580	18		28	5	1	20	"	2	16	3.4
5.SE	Fast/FC VB/ED NPW	3000	120		28	5	1	20	"	2	16	2.5
6.SE	CS/NP W	440	18		28	4	1	15	"	2	16	2.2

7.SE	Fast/FC VB/ED/ NPW	2400	110	6	28	4	1	15	“	2	16	1.5
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Additional sequences

Sagittal TSE fat sat T2

Coronal GR for bony lesions

Post contrast sag. Axial, Coronal T1(If indicated)

SHOULDER- ROUTINE

The patient lies supine with the arms resting comfortably by the side. Slide the patient across the table to bring the shoulder under examination as close as possible to the center of the bore. The arm to be examined is strapped to the patient, thumb up and padded so that the humerus is horizontal. Place the coil (12"GP) to cover the humeral head and superior and medial to the anatomy. Centre the FOV to the Glenohumeral joint

1. Axial Localizer SE T1

Slices are prescribed relative to the horizontal alignment light so that

supraspinatus muscle is included in the image

2 Coronal oblique SE T1

Thin slices are prescribed from the infraspinatus posteriorly to the supraspinatus anteriorly and angled parallel to the supraspinatus muscle. The superior edge of the acromion to the inferior aspect of the subscapularis & deltoid muscle laterally and distal third of the supraspinatus muscle medially are included in the image

3. Coronal oblique TSE T2 Fat sat

Fat suppressed T2 images clearly display muscle tear, trabecular injury, joint fluid and tendon tears

4. Sagittal oblique TSE-T2

Slices are prescribed from medial to the glenoid cavity to the bicipital groove laterally

Additional sequences

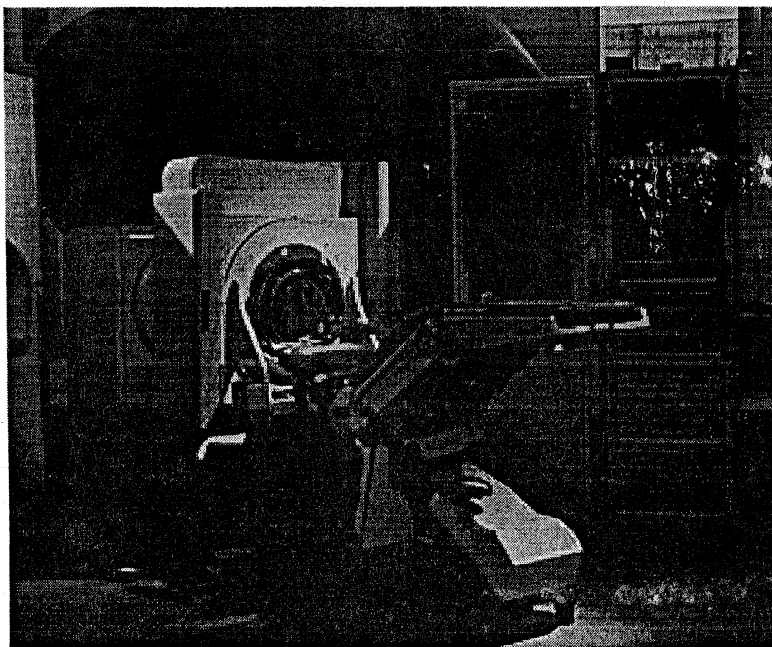
5. Axial GRE T2*

The slices are prescribed from top of the accromio clavicular joint to below the inferior edge of the glenoid

PSD	Options	TR	TE	FLI P°	ET L	FO V	Th k	S P	Loc	Matx	N ex	R b	Ti me
1.SE	CS/NP W	380	17			24	5	1	11	256X 192	1	16	1.4
2.SE	CS/NP W	440	18			24	4	1	14	"	2	16	2.5
3.SE	Fast/FC VB/ED NPW	240 0	120		7	24	4	1	14	"	2	16	1.5
4.SE	Fast/FC/ VB/ED NPW	240 0	120		7	24	4	1	14	"	2	16	1.5
5.GR	FC/NP W	540	18	16°		24	4	1	14	"	2	16	2.1

COMPUTED TOMOGRAPHY

CT is in the fourth decade of clinical use and has been proved to be an invaluable diagnostic tool in many clinical purposes, from cancer diagnosis, trauma to osteoporosis screening. CT was the first imaging modality that made it possible to probe the inner depths of the body, slice by slice. Since 1972, when the first head ct scanner was introduced, ct has matured greatly and gained technological sophistication. Concomitant changes have occurred in the quality of ct images. The first ct scanner, an EMI MARK I, produced images with 80 x 80 pixel resolution (3 mm pixel resolution), and each pair of slices required approx. 4.5 min of scan time and 1.45 min of reconstruction time. Because of the long acquisition times required for the early stage scanners and the constraints of cardiac and respiratory motion, it was originally thought that CT would be practical only for the head scanning.



EMI Scanner

CT is one of the technologies that was made possible by the invention of the computer. The clinical potential of the CT became obvious during its early clinical use, and the excitement

forever solidified the role of computers in medical imaging. Recent advancement in the acquisition geometry., detector technology, multiple detector arrays, and the x-ray tube design have led to scan times that have now measured in fraction of seconds. Modern equipments deliver computational power that allows reconstruction of the image data essentially in real time.



Godfrey Hounsfield

Allan Cormack

The invention of the ct scanner earned Godfrey Hounsfield of Britain and Allan Cormack of united states the Nobel prize for medicine in 1979. CT scanner technology today is used not only in medicine but in other industrial application, such as nondestructive testing and soil core analysis.

Basic principle.

The mathematical principles of CT were first developed by J.Radon in 1917. Radon's treatise proved that an image of unknown object could be produced if one had an infinite number of projections through the object.

Tomographic acquisitions

A single transmission measurement through the patient made by a single detector at a given moment is called a ray. A series of rays that pass through the patient at the same orientation is

called a view or a projection. There are two types of projection geometries, these are the parallel beam geometry and the fan beam geometry, the rays at a given projection angles diverge and have an appearance of a fan beam. All modern ct scanners incorporate fan beam geometry in the acquisition system and reconstruction process. The purpose of the CT scanner hardware is to acquire a large number of transmission measurements through the patient at different positions. The acquisition of a single ct image may be involving approx.800 rays taken at 1,000 different projection angles, for a total of approx 800,000 transmission measurements. Before the axial acquisition of the next slice , the table that the patient is lying down is moved slightly in the cranio-caudal direction which positions a different slice of tissue in the path of the x-ray beam for the acquisition of the next image.

Tomographic reconstruction

Each ray is acquired in ct is transmission measurement through the patient along a line, where these detectors measures an x-ray intensity. I_t . the unattenuated intensity of the x- ray beam is also measured during the scan by the reference detector. And this detects an x-ray intensity I_0 . The relationship between I_t and I_0 is given by the following equation.

$$I_t = I_0 e^{-\mu t}$$

Where t is the thickness of the patient along the ray and μ is the average linear attenuation coefficient along the ray. Notice that I_t and I_0 are machine dependent values, but the product μt is an important parameter related to the patient anatomy along a ray. When the equation is rearranged, the measured values I_t and I_0 can be used to calculate the parameter interest:

$$\ln (I_0/I_t) = \mu t$$

where \ln is the natural logarithm (to base e , $e = 2.79\dots$), t ultimately cancels out, and the value μ for each ray is used in the CT reconstruction algorithm. This computation, which is preprocessing step performed before the image reconstruction,

reduces the dependency of the CT image on the machine dependent parameters, resulting in an image that depends primarily on the patients anatomic characteristics. This is very much a desirable aspect of imaging in the general, and the high clinical utility of CT result, in part, from this feature. By comparison, if a screen film radiograph is underexposed it appears white, and if it is overexposed it appears dark. The density of Ct images is independent of I_0 , although the noise in the image is affected.

After preprocessing of the raw data, a CT reconstruction algorithm is used to produce the ct images. There are numerous reconstruction strategies; however, filtered back projection reconstruction is most widely used in the clinical scanners.

The back projection method builds up in the CT image in the computer by essentially reversing the acquisition steps. During acquisition, attenuation information along the known path of the narrow X-ray beam is integrated by a detector. During the back projection reconstruction, the μ value from each ray is smeared along this same path in the image of the patient. As the data from a large number of rays are back projected onto the image matrix, areas of high attenuation tend to reinforce on each other, as the areas of low attenuation also reinforce, build up the image on to the computer.

GEOMETRY AND HIS TORICAL DEVELOPMENT **GENERATIONS OF CT SCANNERS.**

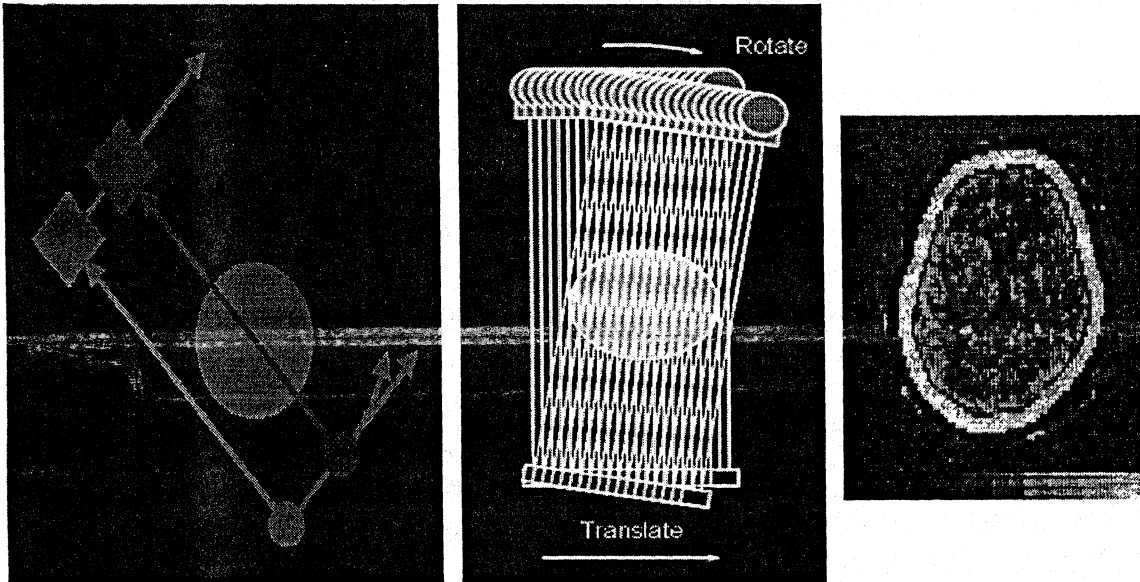
First Generation: Rotate/Transilate, Pencil Beam

CT scanners represent a marriage of diverse technologies, including computer hardware, motor control systems, X-ray

detectors, sophisticated reconstruction algorithms, and X-ray tube/ generator systems. The first generation of CT scanners employed a rotate/ translate, pencil beam systems. Only two X-ray detectors were used, and they measured the transmission of X-rays through the patient for two different slices. The acquisition of numerous projections and the multiple rays per projection required that the single detector for each CT slice be physically moved through out all the necessary positions. The system used the parallel ray geometry. Starting at a particular angle, the X-ray tube and the detectors translated linearly across the FOV, acquiring 160 parallel rays across the 24 cm FOV. When the X-ray tube/ detector system completed its translation, the whole system was rotated slightly, and then another translation was used to acquire 160 rays in the next projection. This process was repeated until 180 projections were acquired at 1° interval. A total of $180 \times 160 = 28,800$ rays were measured.



As the system translated and measured rays from the thickest part of the head to the areas adjacent to the head, a huge change in the X-ray flux occurred. The early detector system could not accommodate such a large change in the signal, and consequently the patient's head was pressed into a flexible membrane filled with water bath. The water bath acted as to bolus the X-rays, so that the intensity of the X-rays outside the patient's head was similar in intensity to that inside the head. The NaI detectors also had significant amount of "after glow" meaning that the signal from the measurement taken at one period of time decayed slowly and carried over into the next measurement if the measurements were made temporarily too close together.



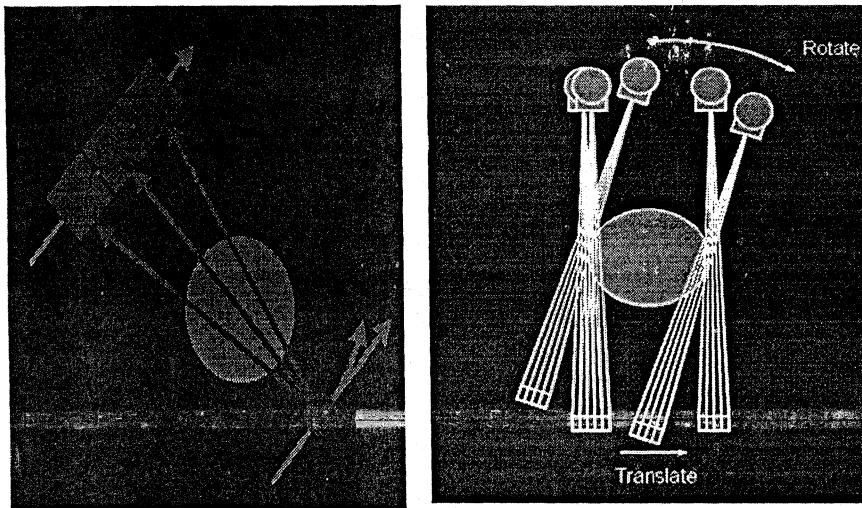
One advantage of the first generation CT scanners was that it employed a pencil beam geometry- only two detectors measured the transmission of X-rays through the patient. The pencil beam allowed very efficient scatter reduction, because scatter that was deflected away from the pencil ray was not measured by the detector. With regards to the scatter rejection, the pencil beam geometry used in the first -generation CT scanners were the best.

Second generation ct scanners: rotate/transilate, narrow fan beam.

The next increment improvement to the CT scanners was the incorporation of a linear array of 30 detectors. This increased the utilization of the X-ray beam by 30 times, compared with the single detector used per slice in the first- generation system. A relatively narrow afn angle of 10^0 as used. In principle a reduction of scan time by 30 fold could be expected. However, this reduction time was not realized, because more data (600rays X 540 views = 324,000 data points) were acquired to

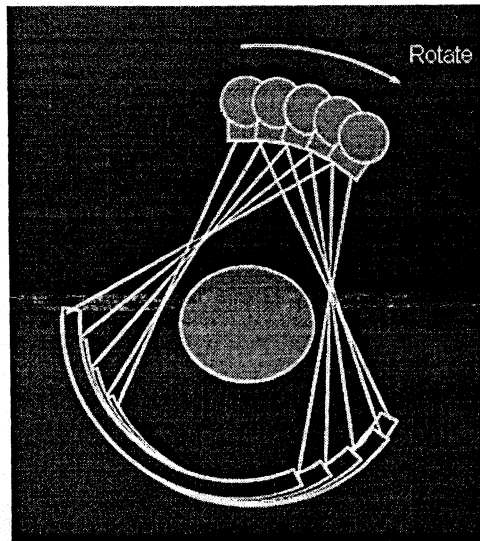
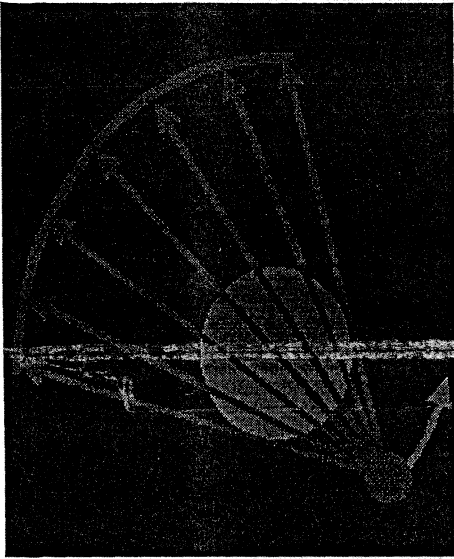
improve image quality. The shortest scan time with a second generation scanner was 18 s/ slice, 15 times faster than the first generation.

Incorporation of an array of detectors instead of just two, required the use of a narrow fan beam of radiation. Although a narrow fan beam provides excellent scatter rejection compared with the plain film imaging, it does allow more scattered radiation to be detected than that with the pencil beam used in the first generation CT.



Third Generation :Rotate/ Rotate, Wide Fan Beam

The translational motion of first and second generation CT scanners was a fundamental impediment to fast scanning. At the end of each translation, the motion of the X-ray tube/detector system had to be stopped, the whole system rotated, and the translation motion restarted. The success of CT as a clinical modality in its infancy gave manufacturers reason to explore more efficient, but more costly, approaches to the scanning geometry.

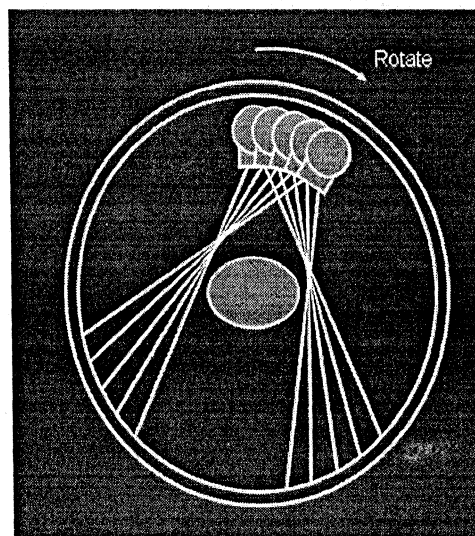


The no: of detectors used in the third generation scanners was increased substantially (more than 800 detectors), and the angle of the fan beam was increased so that the detector array formed an arc wide enough to allow the X-ray beam to interrogate the entire patient. Because detectors and the associated electronics are expensive, this led to more expensive CT scanners. However, spanning the dimensions of the patient with an entire row of detectors eliminated the need for the translational motion. The multiple detector in the detector array capture the same number of ray measurements in one instance as was required by a complete translation earlier. The mechanically joined X-ray tube and the detector array rotate together around the patient without translation. The motion of the third generation CT scanners is called as "rotate/ rotate" referring to the rotation of the X-ray tube and the rotation of the detectors. By eliminating the translational motion, the scan time has reduced substantially. The early third generation scanners could deliver scan times shorter than 5 seconds. Newer systems have scan times of one-half the second.

The evolution from first to third generation involved radical improvement with each step. Developments of the fourth and the fifth generation scanners led not only to some improvements but also some compromises in clinical CT images.

Fourth Generation: Rotate/Stationary

Third generation scanners suffered from the significant problem of the ring artifacts and in the late 1970's fourth generation scanners were designed specifically to address these artifacts. It is never possible to have a large number of detectors in perfect balance with each other, and this was especially true 25 years ago. Each detector and its associated electronics in the system has a certain amount of drift, causing the signal levels of each detector to shift over time. The rotate/rotate geometry of the third generation scanners leads to a situation in which each detector is responsible for the data corresponding to a ring in the image. Detectors towards the center in the detector array provide the data in the reconstructed image in a ring that is small in diameter, and more peripheral detectors contribute to large diameter rings.



Third generation CT uses a fan beam geometry in which the vortex of the fan is in the X-ray focal spot and the rays fan out from the X-ray source to each detector on the detector array. The detectors towards the center of the array make the transmission measurement I_t , while the reference detector that measures I_0 is positioned near the edge of the detector array. If g_1 is the gain of the reference detector, and g_2 is the gain of the other detector, then the transmission measurement is given by the following equation:

$$\ln(g_1 I_0 / g_2 I_t) = \mu t$$

Fourth generation CT scanners were designed to overcome the problems of the ring artifacts. With fourth-generation scanners, the detector removed from the rotating gantry and are placed in a stationary 360-degree ring around the patient requiring many more detectors. Modern fourth-generation CT systems are stationary, fourth generation CT is said to use a rotate/stationary geometry. During acquisition with a fourth-generation scanner, the divergent X-ray beam emerging from the X-ray tube forms a fan-shaped beam. However, the data are processed for a fan beam reconstruction with each detector as the vertex of a fan beam. The rays acquired by each detector being fanned out to a different positions of the X-ray source. In the vernacular of CT, third-generation design uses a source fan. Whereas fourth generation uses a detector fan. The third generation fan data are acquired by the detector array simultaneously, in one instant of time. The fourth generation fan beam data are acquired by a single detector over the period of time that is required for the x-ray tube to rotate through the arc angle of the fan. With fourth generation geometry, each detector acts as its own reference detector. For each detector with its own gain, g , the transmission measurement is calculated as follows.

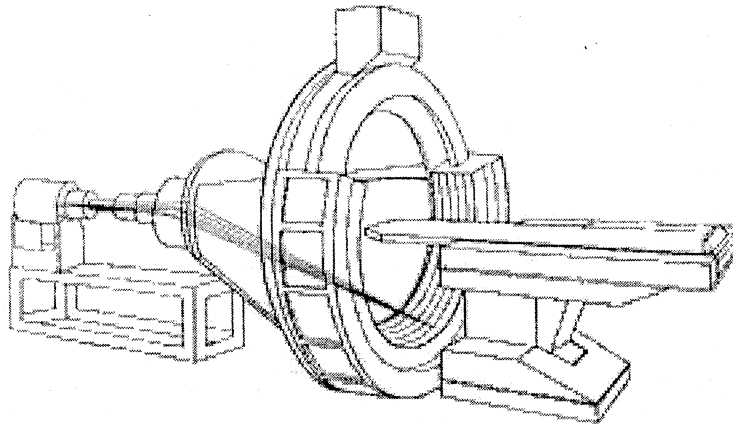
$$\ln (g I_0 / g I_t) = \mu t$$

where the single term g cancels out, therefore the ring artifact are eliminated in the fourth generation scanners. It should be mentioned, however, that with the modern detectors and more sophisticated calibration software. Third-generation CT scanners are essentially free of the ring artifacts as well.

Fifth Generation: Stationary/Stationary

A novel CT scanner has been developed specifically for cardiac tomographic imaging. This "cine-CT" scanner does not use a conventional X-ray tube; instead, a large arc of tungsten encircles the patient and lies directly opposite to the detector ring. X-rays are produced from the focal track as a high-energy electron beam strikes the tungsten. There are no moving parts to this scanner gantry. The electron beam is produced in a cone-like structure (a vacuum enclosure) behind the gantry and is electronically steered around the patient so that it strikes the

annular tungsten target. Cine CT system is called electron beam scanner, are marketed primarily to cardiologists. They are



EBCT

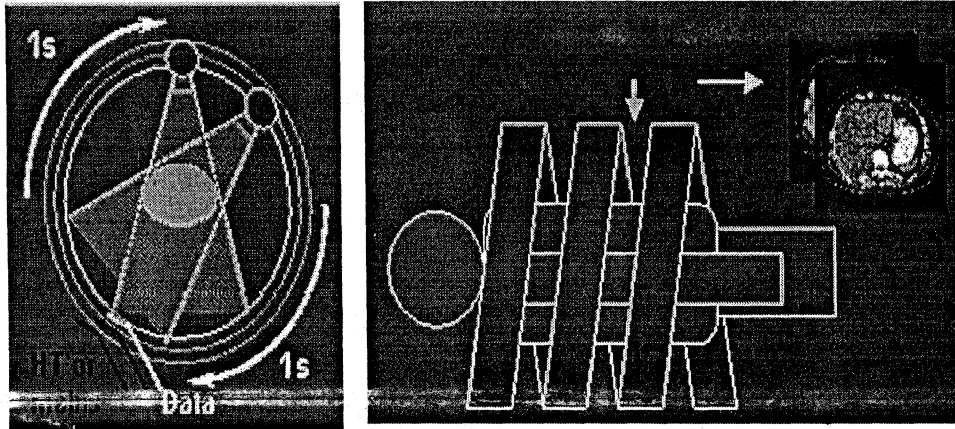
capable of 50 msec scan times and can produce fast-frame-rate CT movies of the beating heart.

Sixth Generation: Helical

Third-generation and the fourth generation scanning geometries solved the mechanical inertia limitation in acquisition of the individual projection data by eliminating the translation motion used in the first- and second-generation scanners. However, the gantry had to be stopped after each slice where acquired, because the detectors and the X-ray tube had to be connected by wires to the stationary scanner electronics. The ribbon cable used to connect the third generation detectors with the electronics had to be carefully rolled out from a cable spool as the gantry rotated, and then as the gantry stopped and began to rotate in the opposite direction and the ribbon cable had to be retracted. In the early 1990's, the design of the third- and the fourth generation scanners evolved to incorporate slip ring technology. A slip ring is a circular contact with sliding brushes that allow the gantry to rotate continuously, unwinded by wires. The use of slip-ring technology eliminated the inertial limitation

at the end of each slice acquisition, and the rotating gantry was free to rotate continuously throughout the entire patient examination. The design made it possible to achieve greater rotational velocities than with systems not using a slip ring, allowing shorter scan times.

Helical CT scanners acquire data while the table is moving; as a result, the X-ray source moves in a helical pattern around the patient being scanned. Helical CT scanners use either third- or fourth- generation slip ring design. By avoiding the time required to translate the patient table. The total scan time required to image the patient can be shorter. Consequently, helical scanning allows the use of less contrast agent and increases patient throughput. In some instances within a single breath hold of patient, avoiding inconsistent level of inspiration.



The introduction of helical CT has introduced many different considerations for data acquisition. In order to produce reconstructions of planar sections of the patient, the raw data from the helical data set is interpolated to approximate the acquisition of planar reconstruction data. The speed of the table motion relative to the rotation of the CT gantry is very important consideration, and the pitch is the parameter that describes this relationship.

The table movement during one gantry rotation to the aperture collimation is expressed as the pitch ratio.

$$\text{Pitch} = \frac{\text{Table feed(s)} \times \text{Gantry rotation / sec.}}{\text{Collimation width.(D)}}$$

In spiral CT the data acquired is a volume data instead of a single 2D data that is acquired in conventional CT. This 3D volume of data can be reconstructed retrospectively for any inter scan spacing (overlapped images), which is useful in detecting very small lesions. Also the area of missed scan is eliminated due to volume data acquisition.

The hardware part of the system uses a built in High-Frequency high-tension generator, and a high heat capacity x-ray tube.

MSHCT refers to the helical CT system equipped with multiple row of detector array for simultaneous collection of data at different slice locations along z direction. It offers substantial improvement of the volume coverage speed and high z axis resolution. It enables sub second scanning (.25sec for 1 slice) and facilitates reduced motion artifact and better temporal resolution.

The multiple detector rows enable to further divide the total x-ray beam into multiple sub beams (by the post collimators, which are equal in number as the detector rows). Let D' be the X-ray beam collimation and d' be the detector row collimation respectively and N' be the number of detector rows; $d = D/N$

The thickness of the individual X-ray beam is determined by the detector row collimation(d) as opposed to X-ray beam collimation in conventional CT. So the helical pitch (P) of multislice CT is given by

$$P = S/d$$

Thus the extended pitch allows high volume coverage without loss of z-axis resolution.

Advantages of spiral CT.

1. Volumetric Data Acquisition
2. Reduced motion and misregistration artifacts.
3. Reduced overall study time - increased patient comfort and quality of the images in seriously ill or un co-operative patients.
4. Reduced radiation and IV contrast dose.
5. Better multiplanar and 3D image reconstruction.

Seventh Generation: Multiple Detector Array CT (MDCT)

The introduction in 1998 of multi-detector row computed tomography (CT) by the major CT vendors was a milestone with regard to increased scan speed, improved z-axis spatial resolution, and better utilization of the available x-ray power. In this review, the general technical principles of multi-detector row CT are reviewed as they apply to the established four- and eight-section systems, the most recent 64-section scanners, and future generations of multi-detector row CT systems.

Evolution Of Spiral Ct: From One Section To 64

The introduction of spiral CT in the early 1990s constituted a fundamental evolutionary step in the development and ongoing refinement of CT imaging techniques. For the first time, volume data could be acquired without misregistration of anatomic detail. Volume data became the basis for applications such as CT angiography, which has revolutionized the noninvasive assessment of vascular disease. The ability to acquire volume data also paved the way for the development of three-dimensional (3D) image-processing techniques such as multiplanar reformation (MPR), maximum intensity projection,

surface-shaded display, and volume-rendering techniques, which have become a vital component of medical imaging today. Ideally, volume data are of high spatial resolution and are isotropic in nature: Each image data element (voxel) is of equal dimensions in all three spatial axes, and this forms the basis for image display in arbitrarily oriented imaging planes. For most clinical scenarios, however, single-section spiral CT with a 1-second gantry rotation is unable to fulfill these requirements. To prevent motion artifacts and optimally utilize the contrast agent bolus, body spiral CT examinations need to be completed within a certain time frame of, ordinarily, one breath hold (25–30 seconds). If a large scan range such as the entire thorax or abdomen (30 cm) has to be covered within a single breath hold, a thick collimation of 5–8 mm must be used. While the in-plane resolution of a CT image depends on the system geometry and on the reconstruction kernel selected by the user, the longitudinal (z-axis) resolution along the patient axis is determined by the collimated section width and the spiral interpolation algorithm. Use of a thick collimation of 5–8 mm results in a considerable mismatch between the longitudinal resolution and the in-plane resolution, which is 0.5–0.7 mm, depending on the reconstruction kernel. Thus, with single-section spiral CT, the ideal of isotropic resolution can only be achieved for very limited scan ranges. Strategies to achieve more substantial volume coverage with improved longitudinal resolution include the simultaneous acquisition of more than one section at a time and a reduction in the gantry rotation time. A first multi slice CT scanner was introduced in 1993 (Elscint TWIN; Elscint, Haifa, Israel). In 1998, several CT manufacturers introduced multi-detector row CT systems, which provided considerable improvement in scanning speed and longitudinal resolution and better utilization of the available x-ray power. These systems typically offered simultaneous acquisition of four sections at a gantry rotation time of 0.5 second.

Simultaneous acquisition of m sections results in an m -fold increase in speed if all other parameters (eg, section thickness) are unchanged. This increased performance of multi-detector row CT relative to single-section CT allowed the optimization of a variety of clinical protocols. The examination time for standard protocols could be substantially reduced, which proved to be of

immediate clinical benefit for the quick and comprehensive assessment of trauma patients and uncooperative patients. Alternatively, the scan range that could be covered within a certain time was extended by a factor of m , which is relevant for oncologic staging or for CT angiography with extended coverage (eg, the lower extremities).

The most important clinical benefit, however, proved to be the ability to scan a given anatomic volume within a given scan time with substantially reduced section width at m times increased longitudinal resolution. Because of this, the goal of isotropic resolution was within reach for many clinical applications.

Examinations of the entire thorax or abdomen could now be routinely performed with a 1.0- or 1.25-mm collimated section width. Despite these promising advances, clinical challenges and limitations remained for four-section CT systems. True isotropic resolution for routine applications had not yet been achieved, because the longitudinal resolution of about 1 mm does not fully match the in-plane resolution of about 0.5–0.7 mm in a routine examination of the chest or abdomen. For large volumes, such as for CT angiography of lower extremity vessels, thicker (eg, 2.5-mm) collimated sections had to be chosen to complete the scan within a reasonable time frame. Scan times were often too long to allow image acquisition during a purely arterial phase. For CT angiography of the circle of Willis, for instance, a scan range of about 100 mm must be covered. With four-section CT at a collimated section width of 1 mm, pitch of 1.5, and gantry rotation time of 0.5 second, this volume can be covered in about 9 seconds, not fast enough to avoid venous overlay, assuming a cerebral circulation time of less than 5 seconds.

As a next step, the introduction of an eight-detector row CT system in 2000 enabled shorter scan times but did not yet provide improved longitudinal resolution (thinnest collimation, eight sections at 1.25 mm). The latter was achieved with the introduction of 16-detector row CT, which made possible the routine acquisition of substantial anatomic volumes with isotropic submillimeter spatial resolution and scan times of less than 10 seconds for 300 mm of coverage. While in-plane spatial resolution is not substantially improved, the two major advantages of fast multi-detector row CT are a true isotropic through-plane resolution and a short acquisition time that enable

high-quality examinations in severely debilitated and severely dyspneic patients .

Traditional CT applications have been enhanced and strengthened by the remarkable, although incremental, improvement in scanner performance by the addition of more detector rows. Multi-detector row CT also dramatically expanded into areas previously considered beyond the scope of third-generation CT scanners that were based on the mechanical rotation of an x-ray tube and detectors, such as cardiac imaging with the addition of electrocardiographic (ECG)-gating capability. With a gantry rotation time of 0.5 second and dedicated image-reconstruction approaches, the temporal resolution for acquisition of an image was improved to 250 msec and less, which proved to be sufficient for motion-free imaging of the heart in the mid- to end-diastolic phase when the patient had a slow to moderate heart rate (ie, up to 65 beats per minute . With four simultaneously acquired sections, coverage of the entire heart volume with thin sections (ie, four sections at 1.0- or 1.25-mm collimation) within a single breath hold became feasible. This 1.0–1.25-mm longitudinal resolution combined with the improved contrast resolution of modern CT systems enabled noninvasive depiction of the coronary arteries . Initial clinical studies demonstrated the potential of multi-detector row CT to not only demonstrate but to some degree also characterize noncalcified and calcified plaques in the coronary arteries on the basis of plaque CT attenuation .

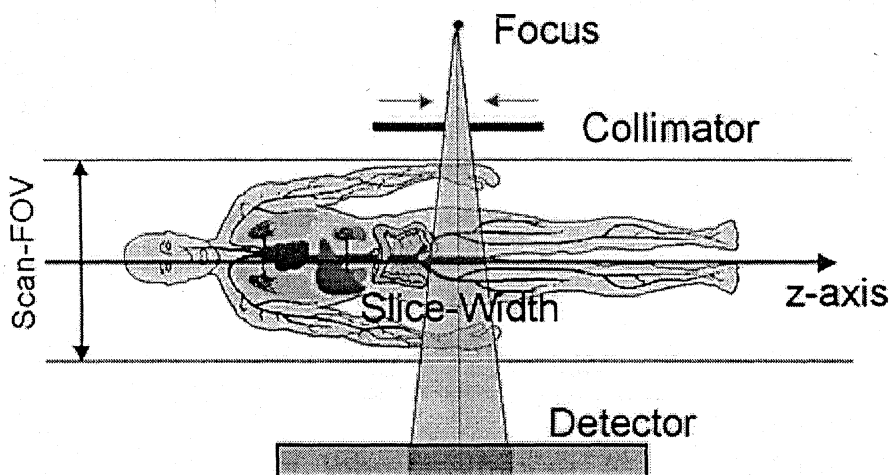
The limitations of four- and eight-detector row CT systems, however, have so far prevented the successful integration of CT coronary angiography into routine clinical algorithms: Stents or severely calcified arteries constitute a diagnostic dilemma, mainly because of partial volume artifacts as a consequence of insufficient longitudinal resolution . For patients with a higher heart rate, careful selection of separate reconstruction intervals for different coronary arteries has been mandatory . It is almost impossible for patients with manifest heart disease to comply with the breath-hold time of about 40 seconds required to cover the entire heart volume (approximately 12 cm) with four-section CT. The ongoing technical refinement of multi-detector row CT, however, holds the promise of gradually overcoming some of

these limitations. The most important steps toward this goal are gantry rotation times faster than 0.5 second for improved temporal resolution and robustness of use, 16-section submillimeter acquisition for increased longitudinal resolution and shorter breath-hold times, and novel sophisticated approaches for image acquisition and reconstruction.

CURRENT TECHNIQUES

System Design

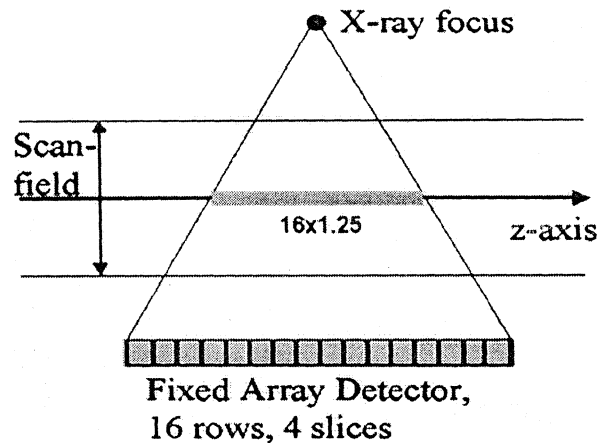
Detector Design.—For clinical purposes, different section widths must be available to adjust the optimum scan speed, longitudinal resolution, and image noise for each application. With a single-detector row CT scanner, different collimated section widths are obtained by means of prepatient collimation of the x-ray beam. For a very elementary model of a two-section CT scanner ($m = 2$, or two detector rows), figure demonstrates how different section widths can be obtained by means of prepatient collimation if the detector is separated midway along the z-axis extent of the x-ray beam. For $m > 2$, this simple design principle must be replaced by more flexible concepts requiring more than m detector rows to simultaneously acquire m sections.



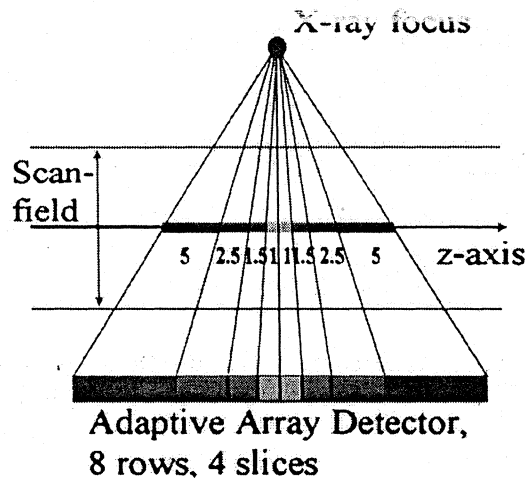
Different manufacturers of multi-detector row CT scanners have introduced different detector designs. In order to be able to select different section widths, all scanners combine several

detector rows electronically to a smaller number of sections according to the selected beam collimation and the desired section width.

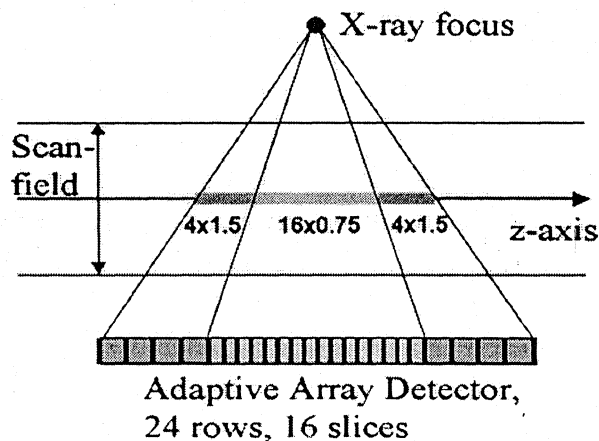
For established four-section CT systems, two detector types are commonly used.



The fixed-array detector consists of detector elements with equal sizes in the longitudinal direction. A representative example of this scanner type, the Lightspeed scanner (GE Medical Systems, Milwaukee, Wis), has 16 detector rows, each of them defining a 1.25-mm collimated section width in the center of rotation. The total coverage in the longitudinal direction is 20 mm at the isocenter; owing to geometric magnification, the actual detector is about twice as wide. By means of prepatient collimation and combination of the signals of the individual detector rows, the following section widths (measured at the isocenter) can be realized: four sections at 1.25 mm, 2.5 mm, 3.75 mm, and 5.0 mm. The same detector design is used for the eight-section version of this system and provides eight sections at 1.25- and 2.5-mm collimated section widths.



A different approach uses an adaptive-array detector design, which comprises detector rows with different sizes in the longitudinal direction. Scanners of this type, the Mx8000 four-section scanner (Philips Medical Systems, Best, the Netherlands) and the Somatom Sensation 4 scanner (Siemens), have eight detector rows. Their widths in the longitudinal direction range from 1 to 5 mm at the isocenter and allow the following collimated section widths: two sections at 0.5 mm, four at 1.0 mm, four at 2.5 mm, four at 5.0 mm, two at 8.0 mm, and two at 10.0 mm.



The selection of the collimated section width determines the intrinsic longitudinal resolution of a scan. In a "step-and-shoot" sequential mode, any multiple of the collimated width of one detector section can be obtained by adding the detector signals

during image reconstruction. In a spiral mode, the effective section width, which is usually defined as the full width at half maximum (FWHM) of the spiral section-sensitivity profile (SSP), is adjusted independently in the spiral interpolation process during image reconstruction. Hence, from the same data set, both narrow sections for high-spatial-resolution detail or for 3D postprocessing and wide sections for better contrast resolution or quick review and filming may be derived.

Sixteen-section CT systems usually have adaptive-array detectors. A representative example for this scanner type, the Somatom Sensation 16 scanner (Siemens), uses 24 detector rows. The 16 central rows define 0.75-mm collimated section widths at the isocenter, and the four outer rows on both sides define 1.5-mm collimated section widths. The total coverage in the longitudinal direction is 24 mm at the isocenter. By means of appropriate combination of the signals of the individual detector rows, either 12 or 16 sections with 0.75- or 1.5-mm collimated section width can be acquired simultaneously. The Lightspeed 16 scanner (GE Medical Systems) uses a similar design: It provides 16 sections with either 0.625- or 1.25-mm collimated section width. The total coverage in the longitudinal direction is 20 mm at the isocenter. Yet another design, which is implemented in the Aquilion scanner (Toshiba, Tokyo, Japan), can provide 16 sections with either 0.5-, 1.0-, or 2.0-mm collimated section width, with a total coverage of 32 mm at the isocenter.

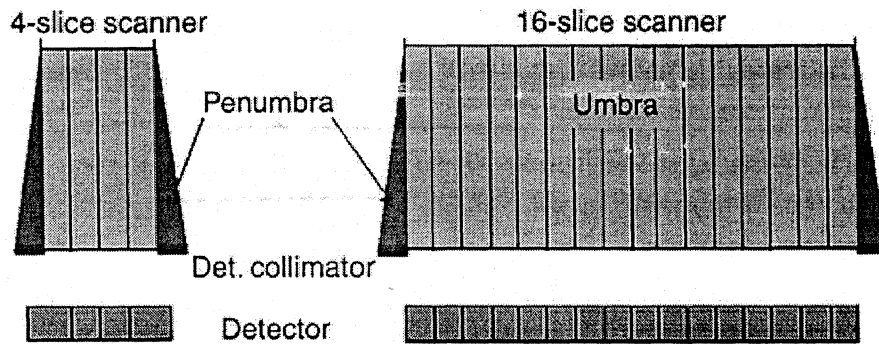
Radiation Dose

Radiation dose and dose efficiency.—Radiation exposure to the patient at CT and the resulting potential radiation hazard have recently gained considerable attention in both the public and the scientific literature. Typical values for the effective patient dose for selected CT protocols are 1–2 mSv for a head CT, 5–7 mSv for a chest CT, and 8–11 mSv for abdominal and pelvic CT. This radiation exposure must be appreciated in the context of the average annual background radiation, which is 2–5 mSv (3.6 mSv in the United States). Despite the undisputed clinical benefits, multisection CT scanning is often considered to require increased patient dose compared with the dose from

single-section CT. Indeed, a certain increase in radiation dose is unavoidable owing to the underlying physical principles.

In the x-ray tube of a CT scanner, a small area on the anode plate, the focal-spot, emits x-rays that penetrate the patient and are registered by the detector. A collimator between the x-ray tube and the patient, the prepatient collimator, is used to shape the beam and to establish the dose profile. In general, the collimated dose profile is a trapezoid in the longitudinal direction. In the umbral region (ie, plateau region of the trapezoid), x-rays emitted from the entire area of the focal spot illuminate the detector. In the penumbral regions, only a part of the focal spot illuminates the detector, while the prepatient collimator blocks off other parts.

With single-section CT, the entire trapezoidal dose profile can contribute to the detector signal, and the collimated section width is determined as the FWHM of this trapezoid. The relative dose utilization of a single-section CT system can therefore be close to 100%. In most cases with multi-detector row CT, only the plateau region of the dose profile is used to ensure an equal signal level for all detector elements. The penumbral region is then discarded, either by a postpatient collimator or by the intrinsic self-collimation of the multisection detector, and represents "wasted" dose. The relative contribution of the penumbral region increases with decreasing section width, and it decreases with increasing number of simultaneously acquired images. This is demonstrated in [Figure 4](#), which compares the "minimum width" dose profiles for a four-section CT system and a corresponding 16-section CT system with equal collimated width of one detector section. Correspondingly, the relative dose utilization with four-section 1-mm-collimation CT is 70% or less, depending on the scanner type. With 16-section CT systems and submillimeter collimation, dose utilization can be improved to 84%, again depending on scanner type. Some multi-detector row CT systems offer special implementations of even more dose-efficient modes that use a portion of the penumbral region.



Dose profiles for four- and 16-section CT systems with identical collimated width of one detector (*Det.*) section. The relative contribution of the penumbral region, which represents wasted dose, decreases with increasing number of simultaneously acquired sections.

A clinically appropriate measure for dose is the weighted CT dose index, or $CTDI_w$, which uses the absorbed dose in a polymerized methyl methacrylate (acrylic plastic) phantom as an approximation of the dose delivered to a cross section of the patient's anatomy. Figure shows $CTDI_w$ at 120 kV for the 32 cm body phantom as a function of the total collimated width of the detector for a four-section CT system and a 16-section CT system with a similar system geometry. The $CTDI_w$ for 16-section CT at 0.75-mm collimation is 7.8 mGy/100 mAs, whereas the $CTDI_w$ for four-section CT at 1.0-mm collimation is 9 mGy/100 mAs. Thus, different from the transition from single-section CT to 4-section CT systems, a further increase in radiation exposure with the more widespread availability of 16-section CT systems is not to be expected.

Concepts for radiation dose reduction.—The most important factor for reducing radiation exposure is an adaptation of the dose to the patient's size and weight. As a general rule, the dose necessary to maintain constant image noise has to be doubled if the patient diameter is increased by 4 cm. Correspondingly, for a patient diameter that is 4 cm smaller than average, half the standard dose is sufficient to maintain adequate image quality. This is of particular importance in pediatric imaging. Dose reduction can be achieved by reductions in the milliampere-seconds and voltage settings. Most CT manufacturers provide

dedicated pediatric protocols with, for example, milliampereseconds and voltage settings adjusted according to the weight of the child.

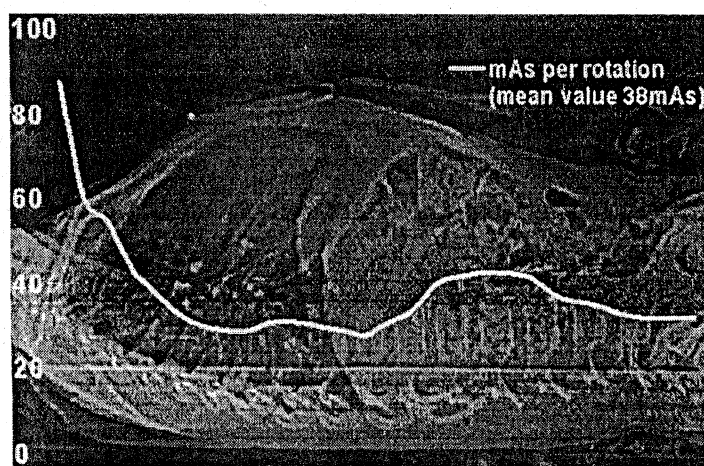
Another means to reduce radiation dose is to adapt the x-ray tube voltage to the intended application. In contrast agent-enhanced studies such as CT angiography, the contrast-to-noise ratio for fixed patient dose increases with decreasing x-ray tube voltage. As a consequence, to obtain the desired contrast-to-noise ratio, the patient dose can be reduced by choosing a lower voltage setting. The potential for dose saving is more substantial for patients with a smaller diameter. This can be demonstrated, for example, by means of phantom measurements of small tubes filled with diluted contrast agent embedded in acrylic plastic phantoms with different diameter. Compared with a standard scan at 120 kV in a 32-cm-diameter phantom (corresponding to that for an average adult), the same contrast-to-noise ratio is obtained with 0.49 times the dose (1.3 times the milliampereseconds setting) for 80 kV and 0.69 times the dose (1.1 times the milliampereseconds) for 100 kV. Thus, ideally, 80 kV should be used for CT angiography in order to reduce patient dose.

Clinical studies have confirmed these findings and demonstrated a potential for dose reduction of about 50% when 80 kV is used for CT angiography instead of 120 kV. In reality, however, the maximum x-ray tube current available at 80 kV is generally not sufficient to scan bigger patients, which limits the routine application of this approach. Therefore, use of 100 kV appears to be a suitable compromise and the method of choice for CT angiography. Figure 5 shows pulmonary CT angiographic images of a patient with pulmonary embolism; the scan was performed on a 16-section scanner at 100 kV and 120 mAs, and the effective patient dose for this scan was 2.3 mSv, 25% less than that for the standard 120-kV protocol.

An approach that is finding increased implementation in clinical practice is anatomic tube current modulation. With this technique, tube output is adapted to the patient geometry during each rotation of the scanner to compensate for strongly varying x-ray attenuation in asymmetric body regions such as the shoulders and pelvis. The variation of the tube output is either predefined by means of an analysis of a localizer scan

(topogram, scout view) or is determined online by evaluating the signal from a detector row. With this technique, dose can be reduced by 15%–35% without degrading image quality, depending on the body region.

In more sophisticated approaches, tube output is modified according to the patient geometry not only during each rotation but also in the longitudinal direction (automatic exposure control), to maintain adequate dose when moving to different body regions (eg, from thorax to abdomen). In one implementation, the attenuation for each body region of a "standard-sized" patient is stored in the control computer. This attenuation corresponds to the milliampere-seconds setting of the standard protocol. If the actual attenuation of the patient deviates from the "standard" attenuation, the tube output is adapted correspondingly. Figure shows the variation of the milliampere-seconds output for a CT scan of the chest and abdomen in a 6-year-old child. Although the standard protocol with 165 mAs was used—which would have resulted in substantially higher radiation dose than necessary in a standard mode of operation—the average milliampere-seconds value throughout the scan was adjusted to 38 mAs by means of automatic exposure control. Automatic adaptation of tube current to patient size prevents both over- and underirradiation, considerably simplifies the clinical workflow for the technician, and eliminates the need for look-up tables of patient weight and size for adjusting the milliampere-seconds settings.



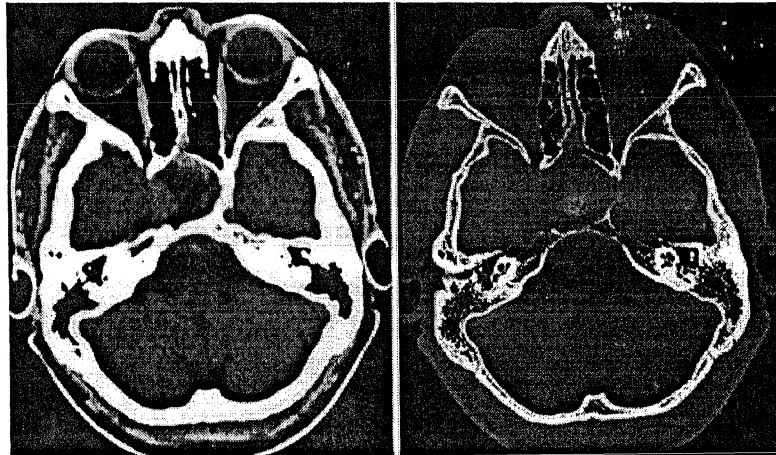
Radiation dose for ECG-synchronized CT for cardiac applications has been a topic of considerable controversy. Recent studies based on four-section CT systems find an effective patient dose of roughly 1 mSv for ECG-triggered calcium scoring with 3-mm section width and roughly 10 mSv for ECG-gated CT angiography of the coronary arteries with 1.0- or 1.25-mm section width. Radiation dose in ECG-gated spiral CT can be reduced by 30%–50% with use of ECG-controlled dose modulation (42,43). During the spiral scan, the output of the x-ray tube is modulated according to the patient's ECG trace. It is kept at its nominal value during a user-defined phase of the cardiac cycle—in general, the mid- to end-diastolic phase. During the rest of the cardiac cycle, the tube output is typically reduced to 20% of the nominal values, although it is not switched off entirely, to allow image reconstruction throughout the entire cardiac cycle. Thus, although the signal-to-noise ratio is decreased at certain phases of the cardiac cycle, the low-dose images are still sufficient for evaluation of functional parameters such as ejection fraction, should this kind of information be desired.

SEQUENTIAL SCANS AND IMAGE-RECONSTRUCTION TECHNIQUES

With the advent of multi-detector row CT, sequential "step-and-shoot" scanning has remained in use for only a few clinical applications, such as head CT, high-spatial-resolution lung CT, perfusion CT, and interventional applications. The number of images acquired during a sequential scan corresponds to the number of active detector sections. By adding the detector signals of the individual sections during image reconstruction, the number of images per scan can be reduced, and the image section width can be increased. As an example, a scan with four sections at 1.0-mm collimation provides either four images with 1.0-mm section width, two images with 2.0-mm section width, or one image with 4.0-mm section width.

The option to realize a wider section by summing several thin sections is beneficial for examinations that require narrow collimation to prevent partial volume artifacts and low image

noise to allow detection of low-contrast details (eg, neurologic examinations of posterior fossa or cervical spine). In the head, partial volume artifacts typically manifest as dark streaks or areas of hypoattenuation and are due to a nonlinear effect. Figure shows an example of a patient who underwent follow-up CT after surgical removal of a pituitary tumor. From the same scan data—four sections at 1.0-mm collimation—both 4.0-mm-thick images with a standard head kernel for soft-tissue evaluation and 1.0-mm-thick images with a bone kernel were reconstructed. For best image quality, the posterior fossa should be scanned with a collimated section width not larger than 1.25 mm, whereas wider collimation can be used in the supratentorial region.



SPIRAL SCANS AND IMAGE-RECONSTRUCTION TECHNIQUES

Spiral scanning is the method of choice for the majority of all multi-detector row CT examinations and requires more attention than does sequential multi-detector row CT because it is conceptually more demanding.

Definition of Spiral Pitch

An important parameter for characterizing a spiral CT scan is the pitch. According to International Electrotechnical Commission specifications, the pitch (p) is given by

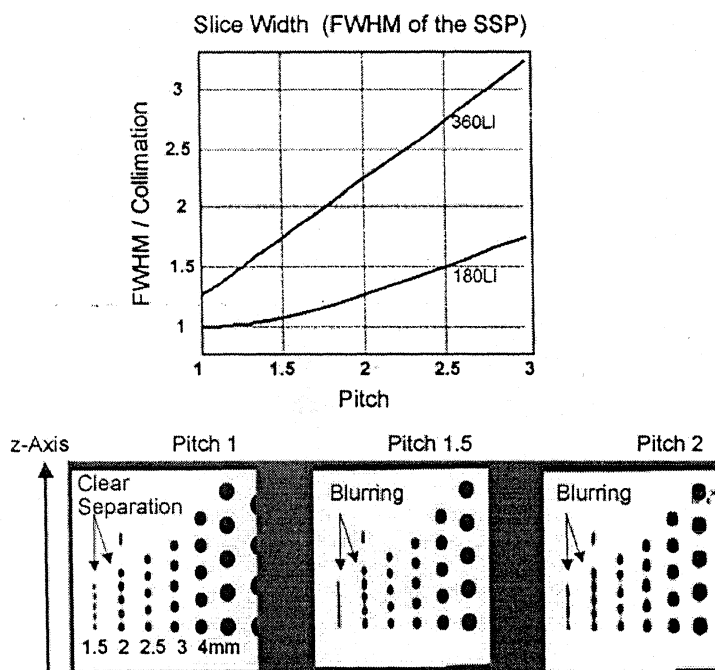
$$p = TF/W$$

where TF is the table feed per rotation, and W is the total width of the collimated beam. This definition holds for both single-section and multi-detector row CT. It shows whether data acquisition occurs with gaps ($p > 1$) or with overlap ($p < 1$) in the longitudinal direction. With 16 sections at 0.75-mm collimation and a table-feed of 18 mm per rotation, the pitch is $p = 18/(16 \times 0.75) = 18/12 = 1.5$. With four sections at 1.0-mm collimation and a table-feed of 6 mm per rotation, the pitch again is $p = 6/(4 \times 1) = 6/4 = 1.5$. In the early days of four-section CT, the term *detector pitch* had been additionally introduced, which accounts for the width of a single section in the denominator. For the sake of clarity and uniformity, the detector pitch should no longer be used.

Short Review of Single-Section Spiral CT Reconstruction

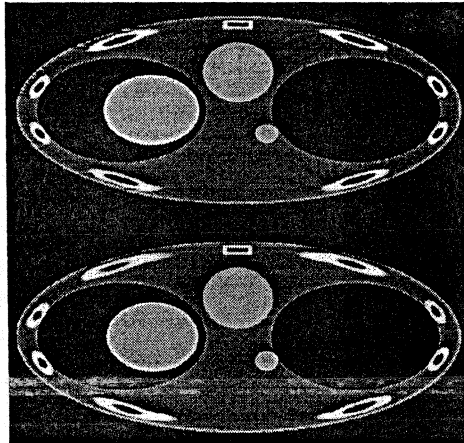
Spiral CT requires an interpolation of the acquired measurement data in the longitudinal (through-plane) direction to estimate a complete CT data set at the desired plane of reconstruction. The most commonly used single-section spiral interpolation schemes are the 360° and 180° linear interpolation methods.

The 360° linear interpolation method exploits the 360° periodicity of the projection data. For each projection angle, a linear interpolation is performed between those two projections on either side of the image plane that are positioned closest to the image plane and are 360° apart (ie, are measured in subsequent rotations). The 180° linear interpolation technique makes use of the fact that for each measurement ray, an interpolation partner is already available after approximately half a rotation, when the x-ray tube and detector have exchanged positions. This is the so-called complementary ray. In spiral CT, z-axis resolution is determined not only by the collimated beam width (as in sequential scanning) but also by the effective section width, which is established in the spiral interpolation process. Usually, the effective section width is defined as the FWHM of the SSP. Effective section width increases with increasing pitch for both 360° and 180° linear interpolation, and longitudinal resolution degrades



This is a consequence of the increasing longitudinal distance of the projections used for spiral interpolation. With 180° linear interpolation, the effective sections width equals the collimated section width at a pitch of 1, but effective section width equals 1.27 times the collimated width at a pitch of 2, so that a collimated 5-mm-thick section is an actual 6.4-mm-thick section at a pitch of 2. The image noise in single-section spiral CT is independent of the pitch if the tube current (in milliamperes) is left unchanged, and patient dose decreases with increasing . Single-section spiral CT is based almost exclusively on 180° linear interpolation, owing to the narrower SSP of this algorithm, despite its increased susceptibility to artifacts and increased image noise. For the same milliamperere-seconds setting, image noise is about 15% higher than that in sequential CT mode. Spiral artifacts gradually increase as pitch is increased. Spiral artifacts typically manifest as hyper- or hypoattenuating "windmill" structures surrounding z-axis inhomogeneous high-contrast objects (eg, bones), which rotate when scrolling through a stack of images. Spiral artifacts are caused by the spiral interpolation process and can also be observed on multi-detector row CT images (see [Fig 8](#)). With single-section CT, scanning at a higher pitch is often used to reduce patient dose at

the expense of section broadening—if the collimation is kept constant—and increased spiral artifacts. For CT angiographic applications in particular, it is more favorable to scan a given volume in a given time by using narrow collimation at a high pitch rather than wider collimation at a low pitch. The motivation for increasing pitch and reducing collimation is to improve longitudinal resolution by narrowing the SSP .



Transverse sections of anthropomorphic thorax phantom from 16-section CT at 0.75-mm collimation and pitch of 1. Images were reconstructed with adaptive multiplanar reconstruction (AMPR; Siemens), with 1.0-mm (top) and 3.0-mm (bottom) section width. Spiral interpolation artifacts are reduced with sections widths that are thick relative to collimation. In clinical practice, best image quality for a desired section width is obtained by acquiring narrow-collimation data.

The Cone-Angle Problem in Multi-Detector Row CT

Two-dimensional image-reconstruction approaches used in commercially available single-section CT scanners require all measurement rays that contribute to an image to run in a plane perpendicular to the patient's longitudinal axis. In multi-detector row CT systems, this requirement is violated. Figure shows the geometry of a four-section scanner: The measurement rays are tilted by the so-called cone angle with respect to the center plane. The cone angle is largest for the sections at the outer edges of the detector, and it increases as the number of detector

rows increases, if their width is kept constant. As a first approximation, the cone angle is neglected in multi-detector row CT reconstruction approaches: The measurement rays are treated as if they traveled perpendicular to the z-axis, and modified two-dimensional image-reconstruction algorithms are used. The data are then inconsistent, however, and produce cone-beam artifacts at high-contrast objects such as bones. It has been demonstrated that cone-beam artifacts can be tolerated if the maximum number of simultaneously acquired sections does not markedly exceed four. As a consequence, the image-reconstruction approaches of all commercially available four-section CT systems and of some systems with even more sections neglect the cone angle of the measurement rays.

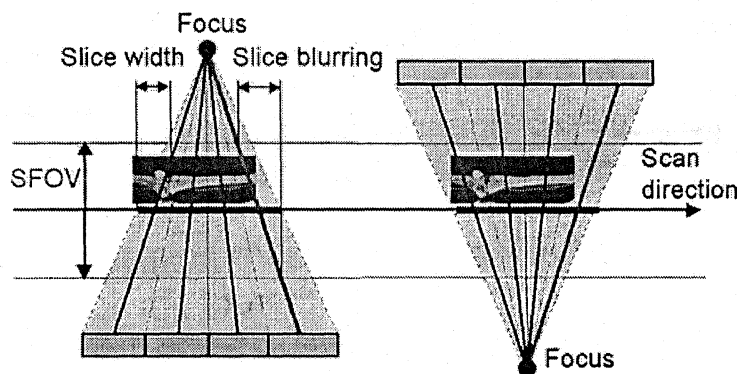


Diagram shows geometry of four-section CT scanner demonstrating the cone-angle problem: Measurement rays are tilted by the so-called cone angle with respect to the center plane. Left and right: Two view angles from sequential scan that are shifted by 180° so that positions of x-ray tube and detector are interchanged. With single-section CT, identical measurement values would be acquired. With multi-detector row CT, different measurement values are acquired. *SFOV* = scan field of view.

MULTI-DETECTOR ROW SPIRAL CT RECONSTRUCTION APPROACHES THAT NEGLECT CONE-BEAM GEOMETRY

Multi-Detector Row 180° and 360° Linear Interpolation

The 360° and 180° linear interpolation single-section spiral reconstruction approaches can be extended to multi-detector

row spiral scanning in a straightforward way. Both 360° and 180° multidetector linear interpolation methods are characterized by a projection-wise linear interpolation between two rays on either side of the image plane. The cone angle of the measurement rays is not taken into account. In the 360° linear interpolation spiral reconstruction approach, rays measured either at the same projection angle by different detector rows or in consecutive rotations of the scanner (ie, 360° apart) are used for spiral interpolation. In the 180° spiral reconstruction approach, both direct and complementary rays are considered. At the isocenter, direct and complementary rays interleave in the z-axis direction for selected pitch values. This way, the distance between measured samples is substantially reduced and equals half the collimated section width, which results in the desired narrow SSPs. Appropriate pitch values are 0.75 for four-section scanning and 0.5625 or 0.9375 for 16-section scanning. The 180° and 360° multidetector linear interpolation approaches are schematically illustrated in Figure E4 for the example of a four-section CT scanner.

In general, scanners that rely on 180° or 360° multidetector linear interpolation techniques and extensions thereof provide selected discrete pitch values to the user, such as 0.75 and 1.5 for four-section scanning or 0.5625, 0.9375, 1.375, and 1.75 for 16-section scanning. These pitch values are intended to provide optimized sampling schemes in the longitudinal direction and, hence, optimized image quality.

The user has to be aware of pitch-dependent effective section widths. For low-pitch scanning (pitch of 0.75 for four sections and 0.5625 or 0.9375 for 16 sections), the effective section width approximates the collimated section width; for a 1.25-mm collimated section width, the resulting effective section width remains 1.25 mm. The narrow SSP, however, is achieved by using 180° multidetector linear interpolation reconstruction with conjugate interpolation at the price of increased image noise. For high-pitch scanning (pitch of 1.5 for four sections and 1.375 or 1.75 for 16 sections), the effective section width is approximately 1.27 times the collimated section width, and a 1.25-mm collimated section width results in a 1.5–1.6-mm effective section width.

When comparing dose and image noise for different pitch values, the widening of the SSP has to be taken into account. To obtain the same image noise as in a sequential scan with the same collimated section width, 0.73–1.68 times the dose (depending on spiral pitch) is required, with the lowest dose at the highest pitch (see reference 50). Some manufacturers provide a semiautomatic adaptation of the milliamperage value to keep the image noise constant if the pitch is changed. In clinical practice, therefore, it is permissible to assume that scanners offering discrete optimized pitch values based on 180° and 360° multidetector linear interpolation techniques are comparable to single-section CT systems in some core aspects: At high pitch, the section widens and the longitudinal resolution degrades; at low pitch, the narrowest possible SSP (comparable to that of 180° single-section linear interpolation at pitch of 1) can be obtained, but a higher dose is necessary to maintain the signal-to-noise ratio. Thus, as a take-home point, when one selects the scan protocol for a particular application, scanning at low pitch optimizes image quality and longitudinal resolution at a given collimation but at the expense of increased patient dose. To reduce patient dose, either milliamperage settings should be reduced at low pitch values or high pitch values should be chosen.

Z-Filter Approaches

In a z-filter multi-detector row spiral reconstruction, the spiral interpolation for each projection angle is no longer restricted to the two rays closest to the image plane. Instead, all direct and complementary rays within a selectable distance from the image plane contribute to the image. The weighting function for the rays is selectable, which allows one to adjust both the functional form and the FWHM of the spiral SSP. Still, the cone angle is neglected. A representative example of a z-filter approach is the adaptive axial interpolation algorithm implemented in Siemens CT scanners (Another example is the "multislice cone-beam tomography," or MUSCOT, algorithm used by Toshiba. Z filtering allows the system to trade off z-axis resolution (the SSP) with image noise (which directly correlates with required dose).

With adaptive axial interpolation, the spiral pitch is freely selectable in the range 0.5–2.0, and the same effective section width, which is defined as the FWHM of the spiral SSP, is generated at all pitch values. Therefore, longitudinal resolution is independent of pitch, unlike single-section spiral CT and multi-detector row CT that relies on 180° and 360° linear interpolation. As a consequence of the pitch-independent spiral section width, the image noise for a fixed tube current (in milliamperes) would decrease as pitch is decreased, owing to the increasingly overlapping spiral acquisition. Instead, the user selects an "effective" milliamperes-seconds value, and the tube current is then automatically adapted to the pitch of the spiral scan to compensate for dose accumulation. The dose for fixed effective milliamperes-seconds is independent of the spiral pitch and equals the dose of a transverse scan with the same milliamperes-seconds setting

Thus, as a take-home point, unlike in single-section spiral CT a change in pitch does not result in a change in dose to the patient. Accordingly, the use of a higher pitch does not result in a dose saving, which is an important practical consideration with CT systems that rely on adaptive axial interpolation.

The intrinsic resolution of a multi-detector row spiral scan is determined by the choice of collimation (eg, four sections at 1.0 or 2.5 mm). Z filtering makes it possible to reconstruct images retrospectively with different section widths from the same raw CT data set. Only section widths equal to or larger than the section width of one active detector row can be obtained. In many cases, both thick sections for initial viewing and recording and thin sections for detailed diagnosis or as an input for advanced 3D postprocessing are routinely reconstructed.

The thinnest available section width is the collimated section width (1.0 mm for four sections at 1.0-mm collimation), which is created by using nonlinear spiral weighting functions at the expense of increased image noise and increased susceptibility to artifacts. Thus, as a take-home point, the thinnest available section should only be used for high-contrast applications such as high-spatial-resolution lung imaging. For general purpose scanning, a 1.25-mm section width for four-section CT at 1.0-mm collimation (and 3.0-mm section width for four sections at 2.5-mm collimation) is recommended as the most suitable trade-off

between longitudinal resolution, image noise, and artifacts, in particular when thin sections are reconstructed as an input for 3D postprocessing such as for MPR, maximum intensity projection, or volume-rendering techniques. For a 1.25-mm spiral section width reconstructed from four-section CT at 1.0-mm collimation, 0.61–0.69 times the dose (depending only slightly on spiral pitch) is required to maintain the image noise of a sequential scan at the same collimation. Unlike 180° and 360° multidetector linear interpolation, image noise is therefore practically independent of pitch at constant dose.

For a given collimation, such as four sections at 2.5 mm, image quality can be optimized with regard to spiral artifacts by lowering the pitch. Another means to reduce spiral artifacts is to use narrow collimation: A given section width (eg, 3.0 mm) can be obtained with different collimations, in this case four sections at 1.0 mm and at 2.5 mm. For optimum image quality, collimation that is narrow relative to the desired section width is preferable. Furthermore, a more rectangular SSP can be established. Figure shows the SSPs of a 3.0-mm section for four-section CT at both 1.0- and 2.5-mm collimation.

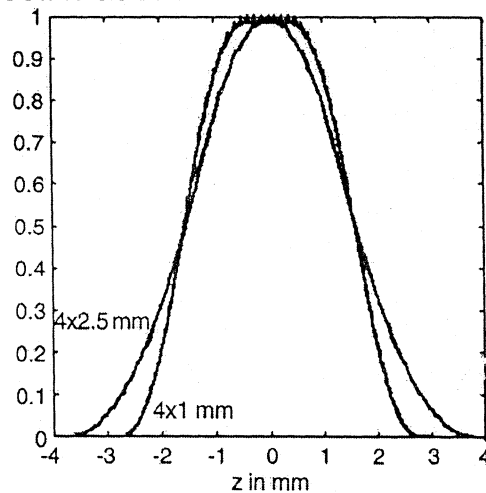
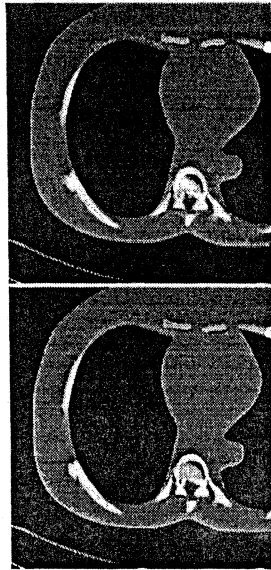
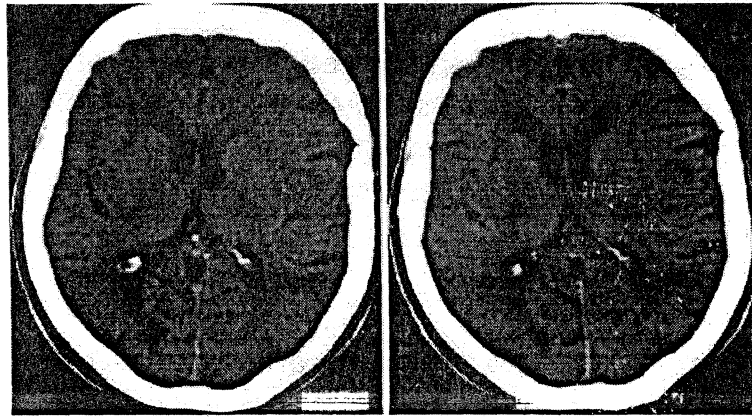


Figure shows 3.0-mm transverse sections of a thorax phantom scanned with four-section CT at 2.5- and 1.0-mm collimation. Despite the higher pitch, the 3.0-mm image obtained at 1.0-mm collimation shows fewer artifacts. Similar to single-section spiral CT, narrow collimation at high pitch is preferable to wide collimation at low pitch for artifact reduction.



Except for a minor dose increase due to the different relative contributions of the penumbral zones of the dose profile, scanning at narrow collimation does not result in higher radiation dose to the patient as long as the effective milliamperere-seconds level is kept constant. Narrow-collimation scanning should, therefore, be the protocol of choice for all applications that require 3D postprocessing as part of the clinical evaluation. In the clinical treatment of uncooperative or trauma patients or for protocols such as routine oncologic staging, the use of wider collimation can be considered. The best suppression of spiral artifacts is achieved by using both narrow collimation (relative to the desired section width) and reduced spiral pitch.

In general, more challenging clinical protocols, such as CT of the spine and of the skull base, are reliant on a combination of narrow collimation and low pitch. When multi-detector row spiral CT of the head is performed with narrow collimation, low pitch, and z-filter reconstruction of wider sections, the results are equivalent to those of traditional sequential CT. Figure shows an example of a head scan performed with a four-section CT system in which a sequential image (two-section CT at 8 mm) and a spiral image (8-mm section width from four-section CT at 1-mm collimation) are compared in the same patient.



Some manufacturers who use a z-filter approach do not provide completely free selection of the spiral pitch but recommend a selection of fixed pitch values (eg, pitch of 0.625, 0.75, 0.875, 1.125, 1.25, 1.375 and 1.5 for four-section CT with the MUSCOT algorithm that are aimed at optimizing the z-axis sampling scheme and reducing spiral artifacts.

MULTI-DETECTOR ROW SPIRAL RECONSTRUCTION APPROACHES THAT ACCOUNT FOR CONE-BEAM GEOMETRY

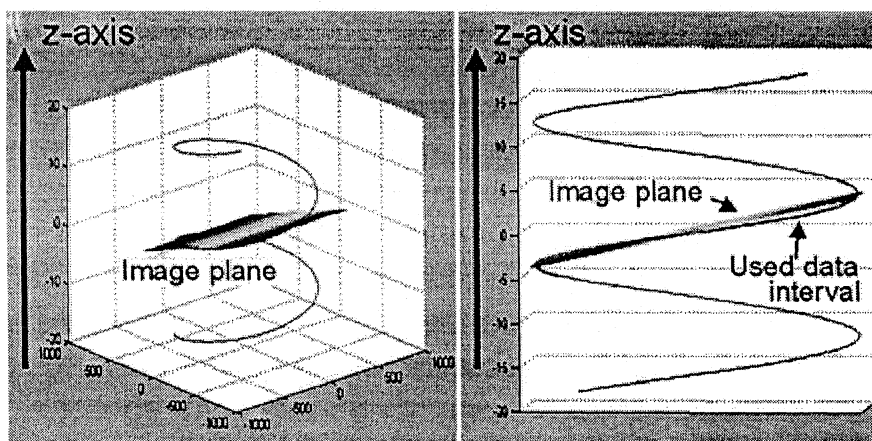
Overview of Cone-Beam Reconstruction Algorithms

For CT scanners with 16 or more detector rows, modified reconstruction approaches that account for the cone-beam geometry of the measurement rays have to be considered. Some manufacturers (Toshiba, Philips) have extended the Feldkamp algorithm, an approximate 3D convolution back-projection reconstruction that was originally introduced for sequential scanning, to multisection spiral scanning. With this approach, the measurement rays are back projected into a 3D volume along the lines of measurement, accounting in this way for their cone-beam geometry. Three-dimensional back projection is computationally demanding and requires dedicated hardware to achieve acceptable image-reconstruction times. Other manufacturers use variations and extensions of nutating-section algorithms for image reconstruction. These algorithms split the 3D reconstruction task into a series of conventional two-

dimensional reconstructions on tilted intermediate image planes, in this way benefiting from established and very fast two-dimensional reconstruction techniques. Representative examples are AMPR (Siemens) and the weighted hyperplane reconstruction (proposed by GE Medical Systems) techniques.

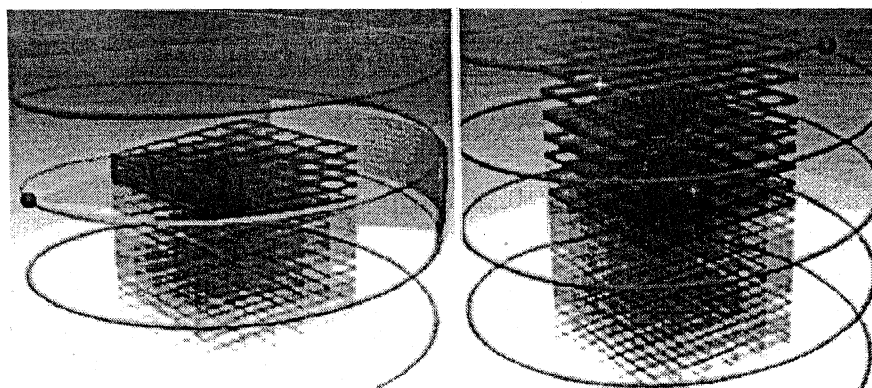
AMPR Method

The AMPR approach is an extension and generalization of the "advanced single-slice rebinning" method. AMPR allows free selection of the spiral pitch with optimized dose utilization, which is beneficial for medical applications. With advanced single-slice rebinning, a partial scan interval (about 240° of scan data) is used for image reconstruction. The image planes are no longer perpendicular to the patient axis; instead, they are tilted to match the spiral path of the focal spot; see Figure for a 16-section scanner at a pitch of 1.5. For every view angle in this partial scan interval, the focal spot is positioned in or near the image plane—that is, measurement rays running in or very close to the image plane are available. These conditions need to be fulfilled for a standard two-dimensional reconstruction. In a final z-axis reformation step, the traditional transverse images are calculated by interpolating between the tilted original image planes.



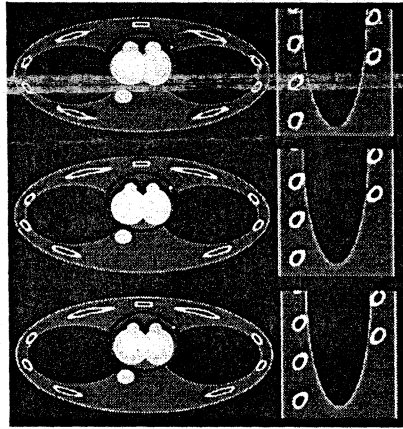
Advanced single-slice rebinning encounters its limitations when the spiral pitch is reduced to make use of the overlapping spiral acquisition and the resulting dose accumulation. The AMPR

algorithm addresses this problem: Instead of all available data being used for a single image, the data are distributed to several partial images on double-oblique image planes, which are individually adapted to the spiral path and fan out like the pages of a book (Fig left). To ensure full dose utilization the number of partial images ("pages" in the book), as well as the length of the data interval per image, depend on the spiral pitch. The final transverse (or arbitrarily oriented) images are calculated by means of z-axis interpolation between the tilted partial image planes (Fig right). The shape and the width of the z-axis interpolation functions are selectable. Different SSPs and different section widths can therefore be adjusted, so that z-axis resolution (SSP) can be traded off with image noise. The spiral pitch is freely selectable and the section width—and consequently the z-axis resolution—are independent of the pitch. The concept of effective milliamperere-seconds and automatic adaptation of the tube current to the pitch also apply to AMPR



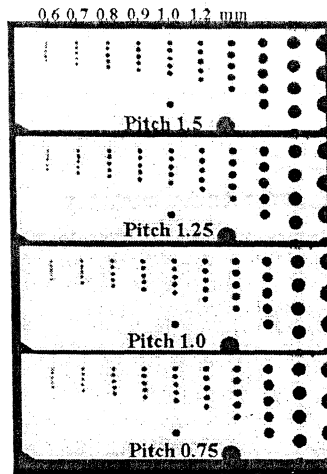
With the AMPR approach, sufficient image quality is obtained for all pitch values between 0.5 and 1.5. Figure shows transverse sections and MPRs of an anthropomorphic thorax phantom. Scan data for 16 sections at 0.75-mm collimation and pitch of 1 were reconstructed with 1-mm section width with z filtering, the AMPR algorithm, and 3D back projection. Neglecting the cone angle leads to artifacts at high-contrast objects and geometric distortions, particularly in MPRs (Fig top). Both AMPR and 3D back projection restore the spatial integrity of the high-contrast objects, reduce cone-beam artifacts, and are fully equivalent for 16-section scanning. Recent studies have demonstrated the

adequacy of extended versions of AMPR for medical CT systems with up to 64 detector rows.



The remaining artifacts in Figure are spiral interpolation artifacts (windmill artifacts), not cone-beam artifacts. Windmill artifacts are not related to the cone-beam geometry and result from the finite width of the detector rows, which require interpolation between the rows for image reconstruction. Hence, windmill artifacts occur independent of the reconstruction approach. They are exaggerated in the mathematic phantom shown (Fig) and can be reduced by decreasing the pitch and/or

increasing the reconstruction section width relative to the collimation. Figure shows MPRs of a z-axis resolution phantom scanned with 16-section CT at 0.75-mm collimation and pitches of 0.75, 1.0, 1.25, and 1.5. Independent of the pitch, all cylinders down to 0.6 mm in diameter can be resolved, the MPRs are relatively free of geometric distortions, and the spatial integrity of the 3D image is maintained.



Multi-detector row spiral CT with AMPR is characterized by the same key properties as adaptive axial interpolation, which can be directly derived from information in the section on z-filter reconstruction presented earlier in this review. Thus, all recommendations regarding selection of collimation and pitch that were discussed there also apply for AMPR. In particular, a change in pitch does not result in a change in radiation exposure to the patient, and the use of higher pitch does not result in dose saving. Narrow collimation should be used whenever possible. With 16-section 0.75-mm-collimation CT, the thinnest available reconstruction section width of 0.75 mm is created by using nonlinear weighting functions at the z-axis image-reformation step, at the expense of increased image noise and increased susceptibility to artifacts. As a take-home point, this approach again should only be used for high-contrast applications such as high-spatial-resolution lung imaging. When thin sections are reconstructed as input for 3D postprocessing such as MPR, maximum intensity projection, or volume-rendering techniques, a 1.0-mm section width is recommended as the most suitable trade-off between longitudinal resolution, image noise, and artifacts.

Weighted Hyperplane Reconstruction

The weighted hyperplane reconstruction method, which has been described elsewhere, uses concepts related to AMPR but is derived differently. Similar to AMPR, 3D reconstruction is split

into a series of two-dimensional reconstructions. Instead of reconstruction of traditional transverse sections, convex hyperplanes are proposed as the region of reconstruction. The increasing spiral overlap with decreasing pitch is handled by introducing subsets of detector rows, which are sufficient to reconstruct an image at a given pitch value. At pitch of 0.5625 with a 16-section scanner, the data collected by detector rows one to nine form a complete projection data set. Similarly, projections from detector rows two to 10 can be used to reconstruct another image at the same z-axis position. Projections from detector rows three to 11 yield a third image and so on. In a way, these "subimages" are related to the "book pages" of AMPR. The final image is based on a weighted average of the subimages. By performing parameter optimizations, an optimal balance among various system performance parameters, such as noise, artifacts, and SSPs, can be achieved.

ECG-SYNCHRONIZED SCAN AND IMAGE-RECONSTRUCTION TECHNIQUES

One of the most exciting new applications of multi-detector row CT is the ability to image the heart and the cardiothoracic anatomy without motion artifacts. For ECG-synchronized scanning of the cardiothoracic anatomy, either ECG-triggered sequential scanning or ECG-gated spiral scanning can be used. In ECG-triggered sequential scanning, the heart volume is covered by subsequent transverse scans with a step-and-shoot technique. For each transverse scan, the number of images corresponds to the number of active detector sections. A partial scan data interval is acquired with a predefined temporal offset relative to the R waves of the patient's ECG trace, which can be either relative (as a certain percentage of the R-R interval) or absolute (in milliseconds) and either forward or reverse. Some 16-section CT systems offer gantry rotation times shorter than 0.5 second (eg, 0.42, 0.40, or 0.37 second). In this case, temporal resolution can be as good as 0.21, 0.20, or 0.185 second.

With retrospective ECG gating, the heart volume is covered continuously by a spiral scan. The basic concepts for ECG-gated spiral imaging, such as single-segment and multisegment reconstruction, had already been developed in 1998. The patient's ECG signal is recorded at the same time the CT data are acquired to allow retrospective selection of the data segments used for image reconstruction. Only scan data acquired in a predefined cardiac phase, usually the diastolic phase, are used for image reconstruction. The data segments contributing to an image begin with a user-defined offset relative to the onset of the R waves, similar to ECG-triggered sequential scanning. Image reconstruction generally consists of two steps: multi-detector row spiral interpolation to compensate for the continuous table movement and to obtain scan data at the desired image z-axis position, followed by a partial scan reconstruction of the transverse data segments. The temporal resolution of an image can be improved up to $t_{rot}/(2N)$ by using scan data of N subsequent heart cycles for image formation in a so-called multisegment reconstruction mode, where t_{rot} is the gantry rotation time of the CT scanner. With increased N , better temporal resolution is achieved but at the expense of slower volume coverage: Increased N and slower patient heart rate require a reduction in spiral pitch.

Multisegment approaches rely on a complete periodicity of the heart motion, and these approaches encounter their limitations in patients with arrhythmia or a heart rate that changes during scan acquisition. Multisegment reconstruction may improve image quality in selected cases, but the reliability of good-quality image acquisitions with N -segment reconstruction is compromised with increases in N .

In general, clinical practice suggests the use of one segment at lower heart rates and two or more ($N \geq 2$) segments at higher heart rates. Use of single-segment versus multisegment reconstruction is integrated in the data acquisition process in a variety of ways, depending on the scanner type. One approach consists of automatic division of the partial-scan data segment into one or two subsegments, depending on the patient's heart rate during acquisition ("adaptive cardio volume" algorithm). With a different approach, single-segment partial-scan images are prospectively reconstructed as baseline images, followed by

retrospective two-segment reconstruction for improved temporal resolution in patients with a higher heart rate. Yet another approach is prospective adjustment of the gantry rotation time to the heart rate of the patient to obtain an optimized temporal resolution for a multisegment reconstruction. Again, this approach requires a stable and predictable heart rate during scan acquisition.

Prospective ECG triggering combined with sequential step-and-shoot acquisition of transverse sections has the benefit of smaller patient dose than that of ECG-gated spiral scanning, because scan data are acquired only during the desired heart phases. However, this technique does not provide continuous volume coverage with overlapping sections, and misregistration of anatomic details cannot be avoided. Furthermore, reconstruction of images in different phases of the cardiac cycle for functional evaluation is not possible. Since ECG-triggered sequential scanning depends on a reliable prediction of the patient's next R-R interval by using the mean of the preceding R-R intervals, the method encounters its limitations in patients with arrhythmia. To maintain the benefits of ECG-gated spiral CT but reduce patient dose, ECG-controlled dose modulation has been developed. The major improvements of 16-section CT, compared with established four-section scanners, include improved temporal resolution due to shorter gantry rotation time, better spatial resolution owing to submillimeter collimation, and considerably reduced scan acquisition times. The time to cover the entire heart volume (about 12 cm) with four-section CT at 1.0-mm collimation is about 40 seconds, which is at the limit for a scan requiring patient breath holding. ECG-gated CT of the entire thorax or the aorta is not possible within reasonable scan durations. For a 16-section CT system, the time to cover the entire heart volume with submillimeter collimation is about 15 seconds. With 16-section CT, coverage of the entire thorax (30 cm) can be completed in about 38 seconds at 0.75-mm collimation and in about 19 seconds at 1.5-mm collimation. ECG-gated examinations of extended cardiothoracic anatomy became feasible with 16-section CT, which lends itself to a spectrum of applications where suppression of cardiac pulsation is desired. Typical diagnostic pitfalls caused by transmitted cardiac pulsation can be avoided, such as an artifactual intimal flap

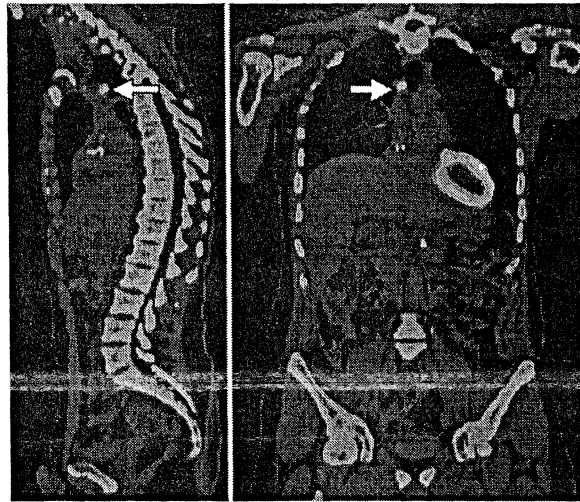
resembling dissection in the ascending aorta. Suppression of cardiac pulsation improves the assessment of paracardiac lung segments and allows confident exclusion of small peripheral pulmonary emboli in segmental and subsegmental arteries. In routine thoracic studies, which are not synchronized to the patient's ECG signal, cardiac motion usually precludes the assessment of coronary bypass grafts.

FUTURE DIRECTION OF MULTI-DETECTOR ROW CT

Sixty four section CT, which has become widely available, enables truly isotropic submillimeter imaging for virtually any application. In the case of cardiac imaging, 64-section CT sets today's benchmark in spatial resolution for noninvasive coronary artery imaging. Motion artifacts in patients with a higher heart rate remain the most important challenge for multi-detector row coronary CT angiography, although diagnostic image quality can be achieved in most cases by administering β -blockers to such patients. Improved temporal resolution is desirable in the future to prevent the need for heart rate control. Increased gantry rotation speed, rather than multisegment reconstruction, appears to be preferable for robust clinical performance. Obviously, substantial development efforts are needed to account for the notable increase in mechanical forces (about 17g for 0.42-second rotation, >33g for 0.3-second rotation) and increased data transmission rates. A rotation time of less than 0.2 second (mechanical force > 75g), which is required to provide a temporal resolution of less than 100 msec independent of heart rate, appears to be beyond today's mechanical limits. An alternative to further increases in rotation speed is to reconsider the scanner concept with multiple tubes and multiple detectors that had already been described in the early years of CT. Owing to its ease of use and its widespread availability, general-purpose CT continues to evolve into the most widely used diagnostic modality for routine examinations, especially in emergency situations or for oncologic staging. CT primarily provides morphologic information; in combination with other modalities, however, functional and metabolic information can also be obtained. Therefore, combined systems for obtaining

comprehensive structural and functional diagnoses will gain increasing importance in the near future.

The combination of state-of-the-art multi-detector row CT with positron emission tomographic (PET) scanners, for instance, opens a wide spectrum of applications ranging from oncologic staging to comprehensive cardiac examinations. The clinical potential of these scanners is currently being evaluated. Reconstruction of the CT images in a sufficient field of view without truncation of anatomic structures (eg, arms) is a prerequisite for adequate attenuation correction of the PET images. An enlarged field of view of up to 70 cm can be realized by extrapolating from the measured CT data. Pertinent algorithms can be found in, for example, Figure shows MPRs from CT images in a 46-year-old man with renal cancer who had undergone nephrectomy and chemotherapy, with PET images superimposed. Areas with increased metabolism are enhanced, and a metastatic mediastinal lymph node can be identified, which supports the notion of PET as adding a "new contrast agent" to CT.



Systems that combine CT and single-photon emission computed tomography are another promising modality. Potential applications are currently being investigated and range from the localization of parathyroid lesions and heterotopic splenic tissue to detection of recurrent nasopharyngeal carcinomas to imaging of aortic prosthesis infection.

CT virtual simulation is gaining increasing importance with a more widespread adoption in 3D conformal and intensity-modulated radiation therapy. With general-purpose CT systems that have a gantry opening with a typical diameter of 70 cm, some patients (eg, women with breast cancer) cannot always be scanned in the treatment position. Such applications, along with interventional procedures and trauma protocols, will be facilitated by CT systems with a larger bore. Recently, concepts have been introduced for four- and 16-section CT scanners with a bore diameter of up to 85 cm and a reconstruction field of up to 82 cm, owing to image reconstruction based on data extrapolation. These systems will probably gain considerable importance in the near future, in particular with regard to the dramatically increasing number of severely obese patients in the Western countries.

For general purpose CT, we will witness a moderate increase in the number of simultaneously acquired sections in the near future. A new generation of CT systems with 32, 40 and—in combination with refined z-axis sampling techniques—64 simultaneously acquired sections are currently being introduced. However in contrast to the transition from single-section to four- and 16-section CT, clinical performance will improve only incrementally with further increases in the number of detector rows. The achievable clinical benefit will have to be carefully considered in the light of the necessary technical efforts and the cost. Clinical progress can more likely be expected from further improvements in spatial resolution rather than from an increase in the volume-coverage speed. In clinical reality, the latter has only rarely been a limiting factor since the introduction of 16-section CT. As soon as all relevant examinations can be performed in a comfortable breath hold of not more than 10 seconds, a further increase in the number of sections will not provide a substantial clinical benefit.

At this point, a qualitative enhancement of CT that allows new clinical applications may again bring substantial clinical progress with, for example, the introduction of area detectors large enough to cover entire organs such as the heart, kidneys, or brain in one sequential scan (approximate scan range, 120 mm). With these systems, dynamic volume scanning would become feasible,

which would open up a whole spectrum of new applications such as functional or volume perfusion studies.

Area-detector technology is currently under development, but no commercially available system so far fulfills the requirements of medical CT with regard to contrast resolution and fast data readout. A scanner with 256 0.5-mm detector elements has been proposed by one manufacturer and appears to be conceptually promising, but this system is still in the prototype stage. Prototype systems by other vendors use cesium iodide–amorphous silicon flat-panel detector technology that was originally used for conventional angiography, which is limited in terms of low contrast resolution and imaging speed. Owing to the intrinsic slow signal decay of flat-panel detectors, rotation times of at least 20 seconds are needed to acquire a sufficient number of projections (≥ 600 projections). The spatial resolution of such systems is excellent, though, because of the small detector pixel size. Excessive dose requirements to date, however, preclude the examination of larger objects. Initial experimental results are thus limited to small high-contrast objects such as joints, the inner ear, or contrast material–filled vessel specimens.

Figure shows a prototype set-up, where a flat-panel detector was incorporated into a standard CT gantry (Somatom Sensation 16; Siemens). The detector covers a 25 x 25 x 18 cm scan field of view, and the pixel size is 0.25 x 0.25 mm, both measured at the center of rotation.

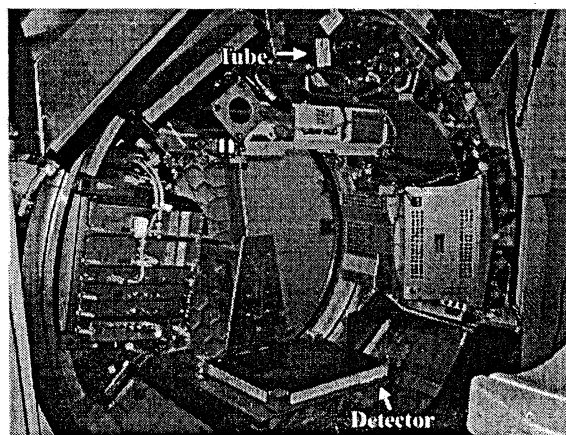
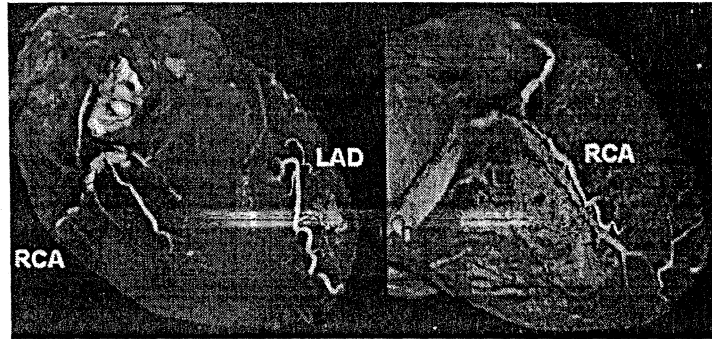


Figure shows volume renderings of a heart specimen (80 kV, 20 mA, 20-second gantry rotation) that demonstrate excellent

spatial resolution, which enables visualization of even very small side branches of the coronary artery tree.



The combination of area detectors that provide sufficient image quality with fast gantry rotation speed will be a promising technical concept for medical CT systems. The vast spectrum of potential applications may bring about another quantum leap in the evolution of medical CT imaging; however such systems will probably not be available in the near future.

POST PROCESSING THE DATA

The raw data acquired by the CT scanner is preprocessed before reconstruction. Numerous pre-processing steps are performed. Electronic detector systems produce a digital data set that is easily processed by a computer. Calibration data determined by the air scans provide correction data that are used to adjust the electronic gain of each detector in array. Variation in geometric efficiencies caused by imperfect detector alignments is also corrected. For example: in fourth generation scanners, the X-ray source rotates in an arch around each of the detectors in the array. As the angle and the distance between the detectors and the X-ray tube changes. These geometric efficiencies are measured during the calibration scans and stored in the computer. They are corrected in preprocessing steps. After the digital data are calibrated. The logarithm of the signal is computed by the following equation

$$\ln(I_0/I_t) = \mu t$$

where I_0 is the reference data, I_t is the data corresponding to each ray in each view, t is the patients overall thickness (in cm) for each ray. And μ is the linear attenuation coefficient (in cm^{-1}).

INTERPOLATION HELICAL:

Helical CT scanning produces a data set in which the X-ray source has traveled in a helical trajectory around the patient. Present-day CT reconstruction algorithms assume that the X-ray source has negotiated a circular, not helical, path around the patient. To compensate for these differences in the acquisition geometry, before the actual CT reconstruction the helical data set is interpolated into a series of planar image data set.

Although interpolation represents an additional step in computation, it also enables an important feature. With conventional axial scanning, the standard is to acquire contiguous images, which abut one another along the cranio-caudal axis of the patient. With helical scanning, however, CT images can be reconstructed at any edge of the scanned volume. Helical scanning allows the production of additional overlapping images with no additional dose to the patient. The sensitivity of the CT images to objects not centered in the voxel is reduced, and therefore subtle lesions, which lay between two contiguous images, may be missed. With helical scanning, in interleaved reconstruction allows the placement of additional images along the patient, so that the clinical examination is almost uniformly sensitive to subtle abnormalities. Interleaved reconstruction adds no additional radiation dose to the patient, but additional time is required to reconstruct the images.

Although an increase in the image count would increase the interpretation time for traditional side-by-side image presentation, this concern will ameliorate as more CT studies are read by the radiologist at computer workstation.

It is important not to confuse the ability to reconstruct CT images at short intervals along the helical data set with the axial resolution itself. The slice thickness dictates the actual spatial resolution along the long axis of the patient.

For example: images with 5 mm thickness can be reconstructed every 1mm, but this does not mean that the image is sampled at 1 mm intervals. To put the example in technical terms, the

sampling pitch is 1mm but the sampling aperture is 5mm. In practice, the use of interleaved reconstruction much beyond a 2:1 interleave yields diminishing returns, except for multiplanar reconstruction or 3D volume rendering.

SIMPLE BACK PROJECTION

Once the image raw data have been pre-processed, the final step is to use the planar projection data sets to reconstruct the individual tomographic images. As a basic introduction to the reconstruction process, assume that a very simple 2X2 "image" is known only by the projection values. Using algebra, one can solve for the image values in the simple case of 4-pixel image. A modern CT image contains approx. 205,000 pixels or "unknowns", and each of the 800,000 projections represent an individual equation. Solving this kind of a problem is beyond simple algebra, and back projection is the method of choice. Simple back projection is a mathematical process, based on trigonometry, which is designed to emulate the acquisition process in reverse. Each ray in each view represents an individual measurement of μ . In addition to the value of μ for each ray, the reconstruction algorithm also "knows" the acquisition angle and position in the detector array corresponding to each ray. Simple back projection starts with an empty image matrix, and the μ value from each ray in all views are smeared or back projected onto the image matrix. In other words, the value of μ is added to each pixel in a line through the image corresponding on the rays path.

Simple backprojection comes from very close to reconstructing the CT image as desired. However a characteristic $1/r$ blurring is a byproduct of simple back projection. Imagine that a thin wire is imaged by a CT scanner perpendicular to the image plane; this should ideally result in a small point on the image. Rays not running through the wire will contribute little to the image ($\mu = 0$). The back projected rays, which do run through the wire, will converge at the position of the wire in the imaging plane, but these projections run from one edge of the reconstruction circle to the other. These projections will therefore "radiate" geometrically in all directions away from a point input. If the image gray scale is measured as a function of

distance away from the center of the wire, it will gradually diminish with a $1/r$ dependency. Where r is the distance away from the point. This phenomenon results in a blurred image of the actual object. When simple back projection is used. A filtering step is therefore added to correct this blurring, in a process known as filtered back projection.

FILTERED BACKPROJECTION RECONSTRUCTION

In filtered back projection, the raw view data are mathematically filtered before back projection onto the image matrix. The filtering step mathematically reverses the image blurring, restoring the image to an accurate representation of the object that was scanned. The mathematical filtering step involves convoluting the projection data with a convolution kernel. Many convolution kernels exist, and the different kernels are used for varying clinical application such as soft tissue imaging and bone imaging.

The kernel refers to the shape of the filter function in the spatial domain, whereas it is common to perform the filtering step in the frequency domain. Much of the nomenclature concerning filtered back projection involves an understanding of the frequency domain. The fourier transform is used to convert a function in the spatial domain into the spatial frequency domain, the inverse Fourier transform is used to convert it back. Convolution is an integral calculus operation.

The difference between filtered back projection and simple back projection is the mathematical filtering operation. Various convolution filters can be used to emphasize different characteristics in the CT images. The Lak filter, named for Dr. Lakshminarayanan, increases the amplitude linearly as a function of frequency and is also called a ramp filter. The $1/r$ blurring function in the spatial domain becomes a $1/f$ blurring function in the frequency domain. Therefore, multiplying with lak filter, where $L(f) = f$, exactly compensates the unwanted $1/f$ blurring, because $1/f \times f = 1$, at all f . this filter works well when there is no noise in the data, but in X-ray images there is always X-ray quantum noise, which tends to be more noticeable in the higher frequencies.

The Shepp-logan filter is similar to the Lak filter but incorporates some roll-off at higher frequencies, and this reduction in amplification at higher frequencies has a profound influence in terms of reducing high-frequency noise in the final CT image. The Hamming filter has even more pronounced high frequency roll-off, with better high frequency noise suppression.

BONE KERNELS AND SOFT TISSUE KERNELS

The reconstruction filters derived by Lekshminarayanan, Shepp and Logan, and hamming provide the mathematical basis for CT reconstruction filters. In clinical CT scanners, the filters have more straightforward names, and terms such as "bone filter" and "soft tissue filter" are common among the CT manufacturers. The term kernel is also used. Bone kernel have less high-frequency roll-off and hence accentuate higher frequencies in the image at the expense of increased noise. CT images of the bone typically have very high contrast (high signal), so the SNR is inherently quite good. Therefore, these images can afford a slight decrease in SNR ratio in return for sharper detail in the bone region of the image.

For clinical application in which high spatial resolution is less important than high contrast resolution. For Example: in scanning for metastatic disease in the liver- soft tissue reconstruction filters are used. These kernels have more role off in higher frequencies and therefore produces images with reduced noise but lower spatial resolution. The resolution of an image is characterised by the modulation transfer function (MTF); The high spatial resolution MTF corresponds to use of the bone filter at small FOV, and the standard resolution corresponds to images produced with the soft tissue filter at large FOV.

CT NUMBERS AND HOUNSFIELD UNITS

After CT reconstruction, each pixel in the image is represented by a high precision floating point number that is useful for

computation but less useful for display. Most computer display hardware makes use of integer images. Consequently, after CT reconstruction, but before storing and displaying, CT images are normalized and truncated to integer values. The number of $CT(x,y)$, in each pixel, (x,y) , of the image is converted using the following expressions:

$$CT(x,y) = 1000 \frac{M(x,y) - \mu_{\text{water}}}{\mu_{\text{water}}}$$

Where $\mu(x,y)$ is the floating point number of the (x,y) pixel before conversion, μ_{water} is the attenuation coefficient of water, and $CT(x,y)$ is the CT that ends up in the final clinical image. The value of μ_{water} is about 0.195 for the X-ray beam energies typically used in the Ct scanning. This normalization results in the CT numbers ranges from about -1000 to $+3000$, where -1000 corresponds to air, soft tissue ranges from -300 to -100 , water is 0, and dense bone and areas filled with contrast agent range up to $+3000$.

CT images are produced with a highly filtered, high KV X-ray beam, with an average energy of about 75keV. At this energy in the muscle tissue, about 91% of X-ray interactions are Compton scatter. For fat and bone, Compton scattering interactions are 94% and 74% respectively. Therefore, CT numbers and hence CT images derive their contrast mainly from the physical properties of tissue that influence Compton scatter. Density (g/cm^3) is a very important discriminating property of tissue, and the linear attenuation coefficient, μ , tracks linearly with density. Other than physical density.

DETECTORS AND DETECTOR ARRAYS.

Xenon Detectors:

Xenon detectors use high-pressure (about 25 atm) non-radioactive Xenon gas, in long thin cells between two metal plates. Although a gaseous detector does not have the same efficiency as a solid one, the detector can be made very thick (e.g.: 6 cm) to compensate in part for its relatively low density. The metal septa that separates the individual xenon detectors can also be made quite thin, and this improves the geometric efficiency by reducing the dead space between the detectors. The geometric efficiency is the fraction of the primary X-rays exiting the patient that strike the active detector elements

The long thin ionization plates of the xenon gas detector are highly directional. For this reason, xenon detectors must be positioned in a fixed orientation with respect to the X-ray source. Therefore, xenon detectors cannot be used for the fourth generation scanners, because those detectors have to record the X-ray source as it moves over a very wide angle. Xenon detectors can be used only for third generation systems.

Xenon detectors for CT are ionization detectors- a gaseous volume is surrounded by two metal electrodes, with a voltage applied across the two electrodes. As X-rays interact with the Xenon gas atoms and cause ionization, the electrical field between the plates causes the ions to move to the electrodes, where the electronic charge is collected. The electronic signal is amplified and then digitized, and its numerical value is directly proportional to the X-ray intensity striking the detector. Xenon detector technology has been surpassed by solid-state detectors, and its use is now relegated to inexpensive scanners.

Solid state detectors

A solid state detector is composed of a scintillator coupled tightly to a photodetector. The scintillator emits visible light when it is struck by X-rays, just as in an X-ray intensifying screen. The light emitted by the scintillator reaches the photodetector, typically a photocathode, which is an electronic device that converts light intensity into electrical pulse proportional to the intensity. This scintillator-photocathode design of solid state CT detector is very similar in the concept to many digital radiographic X-ray

detector systems. However, the performance requirements of the CT are slightly different. The detector size in CT is measured in millimeters., whereas detector elements in digital radiography systems are typically 0.01 to 0.20mm on each side. CT requires a very high fidelity, low noise signal, typically digitized to 20 or more bits.

The scintillator used in solid state CT detectors varies among manufacturers, with CdWO_4 , yttrium and gadolinium ceramics, and other materials being used, because the density and effective atomic number of the scintillators are subsequently higher than those of pressurized xenon gas, solid-state detectors typically have better x-ray absorption efficiency. However to reduce crosstalk between adjacent detector elements, a small gap between detector elements is necessary, and this reduces geometric efficiency somewhat.

The top surface of solid state CT detectors is essentially flat and therefore is capable of x-ray detection over a wide range of angles, unlike the xenon detector. Solid-state detectors are required for fourth-generation CT scanners, and they are used in most high-tier third generation scanners as well.

ARTIFACTS IN CT.

In CT the final image is formed by some mathematical reconstruction process, which make use of X-Ray measurements in reconstructing the density of tissue at every point in the slice. Any error of measurement will reflect as an error in the reconstructed image, which may be a qualitative one or may be less obvious. Such a distortion of the image is called an artifact. Artifacts can be classified according to where their cause stem from - geometric errors, algorithm effects, attenuation measurement error or photon spectrum effects.

Geometric Error.

These are due to the situation in which not all rays measurement occurred along line with the proper orientation in space - a

detector that is not located in its proper position, motion of the patient etc.

Algorithmic effects.

Those caused by the form of recon. algorithm used. Some of the algorithms helps to enhance the image for certain purposes, but this manipulation may often create image that can be misleading. Two effects both of which are due to geometric spreading of the imaged object will be considered

1. Point spread effect: It is noted that the size of an object with high +ve or -ve contrast measured from the scan shows some apparent change in the measurement, and this increases with narrow window.
2. Edge enhancement effect: Arises from a deliberate effort to distort the true density relationship at sharp edges, and is known as edge enhancement. Example : False subarachnoid space in head scan with edge enhancement algorithm.

Attenuation Measurement Errors

These are caused by any malfunction of the detectors that causes an inaccurate attenuation measurement. Even this kind of errors in not completely independent of the patient. Eg: Streak artifacts produced from surgical clips.

There are three main causes of measurement errors.

1. Scattered radiation.
2. Faulty Detector.
3. Faulty x-ray source.

Faulty Detector:

Consider a condition in which one detector in the array is faulty, the resulting ray also became bad. As the source and detector rotate to form subsequent image the same ray will be inaccurate in every view. All these rays are targeted to a circle to form ring artifact. An error of only 0.1 % will cause easily observable circle in the image.

Faulty x-ray source:

If the x-ray output varies during the scan process, an inconsistency in the collected data will immediately occur, since the detectors can't distinguish this change in output. Fixed detector geometry are more sensitive to this output variations. Due to this effects some noise pattern will develop and can be corrected with a separate reference detector which rotate with the x-ray tube in constant geometric relationship. Other possible x-ray source error is momentarily high voltage arching and fluctuations.

Photon spectrum Effect.

This is caused due to the attenuation produced in the x-ray energy spectrum as the beam passes through the patient. It is patient related in the sense that bone and barium produces serious artifacts.

There are two reasons why most CT artifacts appears as streaks.

1. Regardless of the algorithm used any single ray represent a measurement of x-ray transition along a specific straight path through the slice. Thus any error that affects only one ray would appear primarily on those pixels which lies along that ray – as a streak.
2. Rays that shows abrupt change from one view to another. Eg: streak due to patient movement.

Partial Volume Effect.

A voxel represent some finite volume of tissue in the body. If such a voxel contains more than one tissue type then the total content of the voxel are in effect linearly arranged. This effect is significant for relatively small anatomic structure size or for tissue volumes which are rapidly changing in size. These artifacts can be minimized by diminishing the slice thickness although it cannot be eliminated entirely unless beam hardening is also corrected for.

Motion artifacts

When the patient motion occurs the computer has no means to keeping track of where the pixels are in space and which ray path belongs to which row and column. This inconsistency leads to severe streaking and is specially aggravated by the presence of high density structures. The motion not only produces artifactual streaks but as in clinical radiography may cause both a loss of spatial resolu. and a loss of tissue resolu.

Poly chromatic effects.

The best known effect is beam hardening artifact. The basis of this artifact is the fact that in the reconstruction process one attempts to assign an attenuation coefficient value to each voxel within the patient. However attenuation value depends highly on energy of the x-ray beam. As the beam passes through the patient the lower energy photons are preferentially absorbed and we say that beam become harder. Any given voxel within the patient is viewed along different ray path for each different angular projections. The overall effect will be general decrease in the CT number. The visual result of these effect is dark streaks. Eg: inter petrous hypodensity streaks.

Couping artifact.

Seen in the interface of bony structure with adjoining soft tissue due to beam hardening effect. This is evident at the edge of the skull. The ray passing through the periphery of the structure suffers more hardening than those of the centre. So the central region appears darker.

Artifacts related to Spiral CT.**Stair Step artifacts.**

Seen in high contrast longitudinal reformations and is manifested as a disruption of inclined surfaces in regular stair step fashion. It results from the use of interpolation algorithm associated with high contrast interfaces aligned obliquely to the direction of table motion. This artifact is more pronounced when high contrast

structures are more oblique to the direction of patient motion. To eliminate the artifact, both the collimation and the table feed should be less than the longitudinal dimension of the important features on inclined surface and the recon. interval should be less than the table feed.

Break up artifact.

Here the smooth transverse surface of a structure may be seen as irregular with break up at the edges. It is associated with cubic spline interpolation algorithm.

Flow / Contrast Enhancement Artifact

One of the practical problem of spiral CT is that images may be acquired too early after contrast injection. In the liver, unopacified hepatic vein in such instance should not be mistaken as pathological lesion. Filling defects in IVC are extremely common due to unopacified blood and should not be mistaken as thrombus. When thorough IVC delineation is required a second series of delayed images has to be obtained.

Aortic (Ghosting) Artifacts

A more noticeable artifact seen in the ascending aorta and that can simulate aortic dissection. This results from a combination of z - axis blurring and aortic pulsation. Careful inspection images will reveal that the curved thin lucency has extended just beyond the aortic wall in one of the images. This artifacts can be minimized by segmentation of image acquired under 2 sec. Instead of using data throughout 360⁰ cycle, using only data in 60 - 120⁰ rotation will yield images which are not affected by heart movement

HiSpeed CT/i-System specifications

HiSpeed CT/i (GE) is a top of the line helical CT scan system. Using 6.3 MHU x-ray tube which enables 60 sec. Continuous scan with 1 sec. rotation speed.
Generation: Third generation.

X-Ray tube:

Anode - Tungsten Rhenium focal track on a Molybdenum alloy substrate backed by Graphite.
Max. Potential - 140 kVp.
Filament supply - 7 V / 6.5A / 50Hz.
Heat storage : Anode - 6.3MHU..
Focal Spot : Small focus - 0.4 mm x 0.7 mm & 0.6 mm x 0.9 mm
Large focus - 1.2 mm x 1.2 mm
Target angle : 7°

The system automatically select the filament as,
Small : When the technique is 24 KW or less.
Prescription includes detailed, bone or edge recon. Algorithm with KW fall in the range of 24-28.
Large : Technique more than 24 KW.
Technique : 24KW - 28 KW and prescription without bone, detail or edge recon. Algorithm.

KV Choices : 80, 100, 120, 140.
mA Choices : 40 - 400 in 10mA increments.

Exposure time :
Axial Scan - 1, 2, 3, and 4 Seconds.
Cine & Helical - 0.8 sec & 1 Sec.

Scout - 0.26 Sec. - 13.3 Sec. (Scan range 20-1000mm at 75 mm/sec.).

Collimation : 1, 3,5,7,& 10 mm.
Detectors : Hi Light Detectors.
Number: Scanning - 852 & Reference - 12.
Material : Solid Scintillator.
Channel spacing 1mm.
Dose Efficiency: 80% (Generative) and 99% Absorption.

DAS (Data Acquisition System / Analog to Digital Converter)

Channels : 752 Nos.

Sample rate : 984 Hz. Provides 984 views / m Sec.

Tube Control. Regulate : Rotor start and stop time.

Generators : High frequency.

- Comprised of anode and cathode tanks.
- Provides 36 - 48 KW of power.

On Board Computer.

This Computer controls KV, mA and provides mechanism for information input and put put across the slip ring.

Gantry.

Tilt : $+30^{\circ}$ to -30°

Tilt speed : 1° / Sec.

Opening Diameter : 700 mm.

Isocentre to tube distance: 630 mm.

Tube focus to Detector : 1999 mm.

Alignment Light: O/P : $< 1\text{mW}$ / Laser Beam.

Table :

Load Capacity - 180 kg. With +/- 0.25 mm accuracy.

250 kg with +/- 1mm accuracy.

Cradle Travel: 1703 mm max.

Cradle speed : 5mm / Sec. minimum & 75mm/sec. max.

CT.Protocols

Head. – Non traumatic

Routine Head.

- Patient supine.
- Chin tucked down to make the anthropological line vertical.

- Scout lateral view of the whole skull.
- Section planes parallel to supra orbito metal line.
- Thin sections (3-5mm) through posterior fossa.
- 10mm sections above.

Technique: Thin sections in 2 seconds with 200mA & 120 kVp.
10mm sections in 2 seconds with 170mA & 120 kVp.

Facial Bones.

- Patient supine and eyes should be closed to avoid eyeball movements.
- Chin tucked down to make the anthropological line vertical.
- Lateral scout scan.
- Scan sections parallel to IOM line.
- Scan through orbit.
- Slice thickness 3mm.

Technique: 3mm sections in 2 sec. with 200mA & 120kVp.

Coronal.

- Patient supine with neck hyperextended or prone with chin resting in the axial head holder.
- A lateral scanogram is used to select the gantry angulation.
- 3mm images from the orbital apex through the orbit.

Technique: 3mm sections in 2 sec. with 200mA & 120 kVp.

Orbits.

Axial.

- Patient supine & eye should be closed to avoid eye movements.
- Angle of mandible as the zero centering point.
- Lateral localiser scan of the whole skull.
- Scan sections parallel to the IOM line.
- Scan through orbits.
- 3mm slices.

Technique: 3mm sections in 2 seconds with 200 mA & 120 kVp.

Coronal.

- Patient supine with neck hyperextended or prone with chin resting in the axial head holder.
- A lateral scanogram to select gantry angulation.
- 3mm Images from the orbital apex through the orbit.

Technique: 3mm sections in 2 seconds with 200 mA & 120 kVp.

Posterior Fossa.

- Patient supine & eye should be closed to avoid eye movements and radiation to eye.
- Angle of mandible as the zero centering point.
- Lateral localiser scan of the whole skull.
- Scan sections parallel to the IOM line.
- 3mm Sections from foramen magnum to above the third ventricle.

Technique: 3mm sections in 2 seconds with 200 mA & 120 kVp.

Internal Auditory Canals.

Axial.

- Patient supine with OM line perpendicular to the table.
- Lateral scout.
- Scan sections parallel to IOM line.

Technique: 5mm Sections in 2sec. with 170 mA & 120 kVp. to localise the IACs.

After localising the IAC use 1mm scan through the entire IACs using 2sec. 200mA & 120 kVp.

Coronal.

- Patient supine with neck hyperextended or prone with chin resting in the axial head holder.
- A lateral scanogram to select gantry angulation.

- 1mm scans through IAC with smaller FOV.
5mm Sections in 2sec. with 170 mA & 120 kVp. to localise the IACs.
After localising the IAC use 1mm scan through the entire IACs using 2sec. 200mA & 120 kVp.

Temporal Bone.

Axial.

- Patient supine.
- Angle of mandible as zero centre point.
- Lateral Scout scan.
- 1mm Scan to include mastoid and temporal bone.
- Scan sections parallel to IOM line.

Technique: 1mm sections in 2 sec. with 100 mA & 140 kVp or
1sec with 200mA & 140 kVp.
Bone algorithm.

Coronal.

- Patient supine.
- Lateral localiser scan.
- 1mm Scan to include mastoid and temporal bone.
- Scan sections perpendicular to the IOM line.

Technique: 1mm sections in 2 sec. with 100 mA & 140 kVp or
1sec with 200mA & 140 kVp.
Bone algorithm

Sella and Optic Chiasma.

- Patient supine with neck hyper extended.
- Lateral localiser scan.
- 1mm scan through the entire sella.

Technique: 1mm Sections in 2 sec. with 200mA & 120 kVp.
And

3mm sections in 2 sec. with 200mA & 120 kVp. if the sella is large.

Head –Traumatic or Emergency.

Facial Trauma.

- Algorithm – bone and standard for soft tissues.
- 3mm thick direct coronal scan with pitch 1 from front of the nasal bone to behind the sphenoid sinus.
- 3mm Thick axial scans with pitch 1 from below maxillary to above the frontal sinus and include mandible in case of suspected fracture of mandible.
- 1mm helical scans with overlapping for 3D and later reformations, if direct coronal is not possible.

Orbital foreign body.

- Algorithm - detailed.
- 1mm thick helical axial & coronal slices with pitch 1 through orbit.

Cervical Spine – Non traumatic.

Routine (for discs).

- Patient supine, chin extended and the shoulders stretched down.
- Landmark – Sternal notch.
- AP & Lateral scout scans.
- Scans are parallel to and through the disc spaces.
- 1mm or 3mm scans with 2sec. 200mA & 120 kVp.

Thoracic Spine and Chest.

- Patient supine.
- AP & Lateral Scout.
- Slice thickness depending upon the pathology (3 – 10 mm).
Techniques: 2sec. scan with 200mA & 140 kVp.

1sec. scan with 300mA & 140 kVp.

Routine Chest Survey.

- Patient supine.
- Scan from thoracic inlet to adrenals.
- Slice thickness 7 – 10 mm.
- Scan in inspiration (breath hold).

Technique: 1 sec. scan with 240 – 300 mA & 120 – 140 kVp.

HRCT Lung.

- Patient supine.
- Area –whole lung or region of interest.
- 1mm slices with 10mm spacing.

Technique: 2sec. scan with 150 – 200 mA & 140 kVp.
Algorithm. – Lung.

Chest Mediastinum.

- Patient supine.
- Regions.
 - a. Apices to arch – 10mm slices with 10 interval.
 - b. Arch to below carina - 5mm slices with 5mm gap.
 - c. Below carina – 10mm slices with 10mm interval

Technique: 10mm – 1sec. scan with 210 – 240 mA & 120 kVp.
5mm - 1sec scan with 250 – 280 mA & 120 kVp.

Lumbar Spine – For disc.

- Patient supine.
- Iliac crest as landmark.
- AP & Lateral localisers.
- Plan sections through the disc and parallel to it.
- 3mm slices.

Technique: 2sec. scan with 240mA and 140 kVp.

Abdomen Routine.

- Patient supine.
- Landmark – Xyphisternum.
- AP & Lateral Scouts.
- 5 – 10 mm slices.
- Scan at breath hold.
Technique: 1 sec. scan with 240 – 300mA & 120 – 140 kVp.

Pancreas.

- Patient Supine.
- Localise pancreas.
- 3 – 5 mm thick slices.
- 1 sec. scan with 250 – 280 mA & 120 kVp.
- Scan during Breath hold.

Kidneys.

- Patient Supine.
- Localise Kidney.
- 3 – 5 mm thick slices.
- 1 sec. scan with 250 – 280 mA & 120 kVp.
- Scan during Breath hold.

Pelvis.

- Patient supine.
- Landmark – Iliac crest.
- AP & Lateral Scouts.
- 7 – 10 mm Thick slices.
- 1 sec. Scans with 300 mA & 120 kVp.
- Breath hold technique.

CT Angio. Protocols.

Carotid artery.

- Patient supine, neck extended and shoulders stretched down.
- AP and Lateral scout scans.
- Scan thickness 3mm with 50% overlap.

- Pitch 1 - 1.4 as required the area of coverage.
Technique: 1 sec. scans, 250 - 280mA & 100kVp.
Contrast : 90 - 120 cc non-ionic contrast at a rate of 3 - 4 cc / sec.
Scan delay: 8 - 12 sec.

Circle of Willis.

- Positioning as routine axial head scan.
- AP & Lateral Localiser of the skull.
- 1-3mm Thick slices with 1-1.3 pitch.

Technique: 250 – 280mA & 120 kVp . 1 sec. scans with Scan delay 12 – 18 sec.

Contrast: 75 – 125 cc of non-ionic contrast at a rate of 2 – 3 cc/ sec.

Pulmonary Artery.

- Position as for chest scan.
- AP & Lateral Scouts.
- Breath hold should be strictly followed.
- Scan Thickness 5mm with 50% overlap.
- Pitch 1 -1.3.

Technique : 1 sec scan 250 - 280 mA & 120 kVp helical scan.

Contrast : 90 - 120 cc non-ionic contrast at a rate of 3 - 4 cc / sec.

Scan delay : 8 - 12 sec. or smart prep. can be used.

Thoracic Aorta.

- Position as for chest scan.
- AP & Lateral Scouts.
- Scan Thickness 5 - 7 mm with 50% overlap.
- Pitch 1 -1.3.
- Breath hold should be strictly followed.

Technique : 1 sec scan 250 - 280 mA & 120 kVp helical scan.

Contrast : 100 - 125 cc non-ionic contrast at a rate of 2 - 4 cc / sec.

Scan delay : Smart Prep. or 12 - 20 sec. delay.

Coronary artery.

- Position as for chest scan.
- AP & Lateral Scouts.
- Scan Thickness 1 - 3 mm with 50% overlap.
- Pitch 1 -1.3.
- Breath hold should be strictly followed.

Technique : 1 sec scan 250 - 300 mA & 120 kVp helical scan.

Contrast : 100 - 125 cc non-ionic contrast at a rate of 2 - 4 cc / sec.

Scan delay: Smart Prep. or 12 - 15 sec. delay.

Abdominal Aorta.

- Position as for abdominal scan.
- AP & Lateral Scouts.
- Scan Thickness 5 - 7 mm with 50% overlap.
- Pitch 1 -1.4.
- Breath hold should be strictly followed.

Technique : 1 sec scan 250 - 280 mA & 120 kVp helical scan.

Contrast : 100 - 125 cc non-ionic contrast at a rate of 2 - 4 cc / sec.

Scan delay : Smart Prep. or 18 - 25 sec. delay.

Renal Artery

- Position as for abdominal scan.
- AP & Lateral Scouts to localize kidneys.
- Breath hold should be strictly followed.
- Scan Thickness 3 mm with 50% overlap.
- Pitch 1 - 1.2.

Technique : 1 sec scan 250 - 280 mA & 120 kVp helical scan.

Contrast : 100 - 125 cc non-ionic contrast at a rate of 2 - 4 cc / sec.

Scan delay : Smart Prep. or 12 - 20 sec. delay.

Renal Donors Angio.

CT angiography / pyelography may be used in preference to intravenous urography and catheter aorto-renal angiography.

Preliminary non contrast helical CT with 5mm collimation and 1.4 pitch to access the renal calcification.

Begin the CTA at the level of SMA so that scanning through the renal vascular pedicle occurs prior to significant opacification of the renal vein. Extend the scan sequence inferior to the level of common iliac artery bifurcation.

Obtain AP Scout Image 3 - 4 minute after the injection for pyelogram.

Injection : 5cc / sec. for 30 sec.

Scan Delay: Smart prep. Or by mini bolus testing (4cc / sec. for 4 sec.).

Scanning : SMA to iliac bifurcation.

Scan thickness: 3 - 5mm with 50% overlap and pitch of 1- 1.3.

Clinical Applications of CT.

Spiral CT has wider clinical application than initially anticipated. It can be used in head, thorax, abdomen, spinal region as well as for 3D CT angiography.

Thorax.

Volumetric CT data from a spiral acquisition can be used to obtain contiguous overlapping thin sections, which improve the sensitivity for detection and characterization of pulmonary nodules.

Determination of the site of origin of the masses adjacent to the diaphragm is improved by the use of multiplanar images generated from the spiral CT data. Spiral CT also improves all the evaluation and local staging of lung of lung cancer compared to conventional CT, for example sensitivity ratio of pleural involvement increases to a dramatic 92%.

Spiral CT helps in better assessment of the lung parenchyma for the evaluation of interstitial lung diseases and bronchiectasis. It improves detection of lymph adenopathy in anatomically complex location like aorto-pulmonary window and hilar region. Mediastinal vascular anomalies such as dissection, aneurysms, and coarctation can be rapidly, accurately and reliably accessed.

Abdomen.

For the liver, spiral CT has obvious advantages. Spiral CT due to very fast scanning optimizes the contrast dynamics in the liver and has improved detection of metastatic lesion. Both hypovascular and hypervascular metastasis can be detected with better accuracy. Most gastrointestinal neoplasm's are hypovascular, whereas lesions from islet cell tumours, renal neoplasms, sarcomas, lymphomas and melanoma can be hypervascular. Depending upon the type of the primary tumour, spiral CT could be tailored to image the liver either during the arterial phase, (for hypervascular lesions) or during the portal venous phase (at these time there is peak enhancement of the liver parenchyma, maximizing liver to lesion difference) for hypovascular lesions. For suspected hypervascular metastasis a delay time of 25 - 30 sec can be used, whereas for hypovascular metastasis a delay time of 60 sec. is used.

For pancreatic pathologies spiral CT has become the gold standard. Vascular opacification is optimum which is very important for evaluating pancreatitis and malignant tumours of pancreas. In pancreatitis spiral CT can detect the vascular patency and evaluate for pseudo-aneurysms. In pancreatic tumours it is important to opacify the portal vein well to look for enhancement.

In the kidneys, Spiral CT is extremely useful to evaluate and characterize renal masses which are smaller than 1.5 cm in size. Spiral CT with overlapping sections and 3D reconstruction is particularly useful for planning surgical exploration in small indeterminate lesions and for planning resections of large corners in solitary kidneys.

In this initial series, the cortex enhances whereas medulla may still be hypodense. It is important to note that when the kidneys are being evaluated, small tumours of the medulla may not be missed. Some unique application of spiral CT that has been recently advocated is its use in renal colic and for CT urograms. A 40 - 50 sec. Unenhanced spiral CT study from the top of the kidney to the base of the bladder for renal colic is an accounted way of detecting renal calculi.

It is feasible to use data from a contrast enhanced spiral CT to obtain CT renal angiography and to evaluate the renal parenchyma at the same time. Using 3D reconstruction from the delayed images the dense contrast within the collecting system can be used to obtain intra venous urogram like images.

Cranio- Spinal Applications.

For the brain, conventional scan are still preferred for routine indications, as the resolution obtained is slightly better than with spiral acquisition. However in case of un co-operative and pediatric patients, it is extremely useful to use a spiral acquisition. 3D Spiral CT angiography can be used for the carotid bifurcation and intra cranial vascular abnormalities. 3D recon. For evaluation of congenital diseases like cranio stenosis and complex facial trauma are better performed with spiral CT.

Musculoskeletal Applications.

Spiral CT can be used in the evaluation of complex musculoskeletal disease processes like bone or soft tissue tumours, infective processes or complex traumatic processes. In areas where the anatomy is complex, trauma can be reliably and more completely evaluated with 3D recon.(Cranio- vertebral region, acetabular fractures etc.).

CT Angiography.(CTA).

CT angiography is a relatively non-invasive technique to obtain vascular details. If an intravenous bolus of contrast material is administered and timed correctly, images can be obtained at the peak of arterial enhancement with helical scanning. New software timing sequences such as 'smart prep' further improves the quality of CT angiography by allowing accurate timing of the contrast bolus. These images can be post processed to display three dimensionally.

3D CTA gives images which are more or less similar to the DSA images.

3D WORK STATION.

Advantage Windows (AW).

Operating System: Sun OS 5.4 / Type: SUNW, SPARC station
Processor type: Spare/ Processor No: 1

Processor1. Clock : 150MHz / Memory Size: 128MB

No. of Disks: 3

It is a multitasking, multifunctional workstation, based on Sun Microsystems, designed for the display and analysis of the images. AW is networked with MRI, CT and DSA. It is equipped with several software like 3D, Reformat, Denta (Dental CT), BMD & Navigator (Virtual Endoscopy).

3D Analysis Package.

It is a software application designed to be used with AW. It generate reconstructed 3 Dimensional volume and reformatted cut plates on CT and MR series. 3D imaging can be performed using two physical methods: Thresholding & Percentage based. The basic different between the two techniques is that thresholding based software assumes a particular voxel to be of uniform density, where as percentage based software assumes that a voxel contains more than one type of tissue. Advantage windows is a thresolding based software. Here thresolding, erosion, dilation and select object are the four most often used tools for segmenting anatomy for 3D Surface/volume rendering. The reconstructed images can be displayed as Shaded Surface Display (SSD), Maximum Intensity Projection (MIP), High definition MIP (HD MIP), minimum Intensity Projection, Raysum and Integral.

Surface rendering was one of the earliest methods of 3D display and is now incorporated in most commercially available 3D medical imaging packages. In this method, each voxel within the data set is determined to be a part of or not a part of the object of interest. Usually by comparing the voxel intensity to some threshold value, thereby defining the 'Surface' of the object.

With the surface determined, the rest of the data is discarded, Surface contours are typically modeled as a collection of polygons and displayed with surface shading. The resulting image is a simplified and possibly misleading representation of a structure, particularly if the surface is difficult to determine precisely, as is often the case in medical imaging.

By converting data from a volume to a surface, a large portion of the data available is forfeited in exchange for faster, easier computation. This can be an advantage by allowing real-time rendering and thereby enhancing user interactivity, but the usefulness of surface-rendered medical images is generally limited by their inconsistent image fidelity.

Like surface rendering, Maximum intensity projection (MIP) is also commercially available in 3D software packages and has been extensively clinically evaluated, particularly in reconstruction angiographic images from CT and MR data.. The MIP algorithm evaluates each voxel along a line from the viewer's eye through the data set and select the maximum voxel value as the value of the corresponding display pixel. The resulting images are typically not displayed with surface shading or other devices to help the user appreciate the 'depth' of the rendering, making three-dimensional relationships difficult to assess.

If there is another high-intensity material along the ray through a vessel, such as a calcification, the displayed pixel will represent the calcification and will contain no information from the intravascular contrast material. Selection of the heighest pixel value also increases the background mean of the image, which enhances structures such as the kidney and liver, there by decreasing the visibility of vessels in these structures.

What is worse, volume averaging coupled with the MIP algorithm typically produces artifacts: a normal vessel passing obliquely through a volume may look like a string of beads. In spite of its shortcomings, MIP is generally more accurate than surface rendering for CT angiography.

As its name implies, Volume rendering makes use of the entire volume of data rather than just surfaces and so potentially conveys more information than a surface model.

Volume rendering techniques sum the contributions of each voxel along a line from the viewer's eye through data set. This is done repeatedly to determine each pixel value in the displayed image. Because the entire data set is incorporated into the resulting image, computers that are more powerful are necessary to do volume rendering at a reasonable speed.

We view volume rendering as the most advanced form of 3D image processing available creating accurate, clinically useful images. Volume rendering is just now being incorporated into commercially available software. With general availability and continued increase in computer power, volume rendering will likely become the most important technique for 3D medical imaging.

Denta scan (Dental CT)

It is a software programme, which allows us to evaluate the teeth, and teeth bearing portion of the jaws in a manner differ from routine axial sections and sagittal or coronal reconstructions.

The software is constructed such a way that multiple, contiguous axial sections are fed in to it to yield panoramic(OPG like) and para axial reconstructions. Para axial reconstructions are perpendicular to the plane of the tooth and are the only image, which shows all four cortices of the mandible well.

The technique: Axial images of 1 to 2 mm thickness with 1mm gap are obtained with a low kV setting of 90-110 and a low mA setting of 120-180, Use a matrix of 512 X 512 and a high definition bone algorithm with a display FOV of 15cm . The mandible and maxilla have to be scanned separately. A bite is placed in the mouth to adequately separate the mandible and maxilla. For mandible place the patient so that the mandible is

perpendicular to the floor plane. For maxilla place the patient so that the maxilla is perpendicular to the floor plane.

Bone Mineral Densitometry (BMD)

Quantitative computer tomography (QCT) is the technique used to measure the trabecular BMD in the lumbar spine and trabecular and cortical bone in the fore arm. These measurements have been used to provide normal reference values and to assess diseases that affect both bone mineral density and body composition. BMD measurements is currently the most accurate method for predicting patients at a risk of osteoporotic fracture. These measurements are also useful for monitoring the response to therapy for prevention of bone loss and fracture.

The QCT-5000 bone densitometry system allows highly automated vertebral bone mineral density measurements. The hardware consists of a calibration phantom, a QA Torso phantom, phantom cushion and bolus bag.

Technique : Taking a lateral scout image and axial scans through 3 or 4 vertebrae of the patient while they are laying on the calibration phantom. Care should be taken to perform the scans with a standard mA, Scan time, Slice thickness, Scan FOV and Table height.

NAVIGATION (Virtual Endoscopy)

The navigator package generates 3-dimensional volumes of hollow structures on advantage windows system. The multiple screen displays the position and direction of the navigator, in 3 planes. A mouse and keyboard driven navigator provides luminal views in real time.

Virtual endoscopy or fly-through methods that combine the features of endoscopic viewing and cross sectional volumetric imaging may provide and advances in diagnosis. Virtual endoscopy presentations of image data enables the operator not

only to explore the inner wall surfaces but also to navigate inside the virtual organs extracted from CT Scan or MR images.

Imaging requirements - All routine CT or MR imaging techniques that provide high-resolution cross-sectional images can be post processed to obtain 3D re-constructions. CT scanning for virtual endoscopy is performed using a continuously rotating helical CT system. Typical helical CT data acquisition parameters are 120 kVp, 70-165 mA, 20-40 sec exposure, 5-3mm collimation, 5-6mm/sec table feed (pitch of 1.2), and 512 X 512 matrix. Reconstruction parameters include 180 linear interpolation, standard reconstruction kernel, and 3mm table incrementation. For virtual colonoscopy studies, after oral colonoscopy preparation, patients are scanned after ingestion of oral contrast agent (1000 ml of 2% Barium Sulfate suspension) or, alternatively, air insufflation can be applied. Glucagon may have a role in improving image quality by reducing bowel peristalsis. The optimal CT technique for virtual colonoscopy has not yet well established. For CT abdominal angiography studies, helical CT scans are obtained after IV injection of contrast agent.

For MR imaging, T1 weighted 3D volumetric gradient-echo sequences provide isotropic voxels that are applicable for 3D rendering- For vessel fly-through method, 3D Time-of-flight MR angiography is preferred over phase contrast angiography because of the better delineation of vessel wall.

Leksell Stereotactic System.

Introduction

In 1920 Clarke, who interestingly was not a surgeon himself, suggested the use of mechanically directed instruments for surgery in the depth of the brain. However, the first intracerebral stereotactic operation was not performed until 1947 when Spiegel and Wycis introduced "stereoencephalotomy" for the treatment of intractable pain,

The Leksell Stereotactic instrument was introduced and used for the first time in 1949. Although continuous technological

improvements and many valuable modifications have been made, the original basic principle of the instrument, with the target in the center of the semicircular arc, has been maintained since that time.

Stereotaxy is often considered the simplest and safest approach to the deep areas. It is a highly suitable technique for making deep functional lesions and often the only desirable method for biopsies, for the treatment of deep-seated tumors such as cranio pharyngiomas and cysts of the foramen Munro and for the evaluation of deep hemorrhages. Implantation of radioactive seeds for interstitial irradiation necessitates the precision offered by stereotaxy in order to comply with dose-planning requirements.

With the development of the Leksell Stereotactic Gamma Knife in 1968, which is based on the same principles as the open technique, the number of indications for stereotactic treatment, now also by radiosurgery, has increased further. With this technique conditions such as A V M s, acoustic tumors, pituitary adenomas, pineal tumors, as well as other benign and malignant tumors can be treated without opening of the skull. The very high precision of the gamma Knife lends itself particularly well to small and critically located targets. Radiosurgery in functional disorders and in the treatment of vascular malformations in the brain stem are good examples for such targets.

The combination of stereotactic surgery and computerized tomography (CT) has been very fruitful and, with magnetic resonance imaging (M R I) new dimensions have opened up.

The Leksell micro - Stereotactic System.

The principal components of the Leksell Micro Stereotactic system are the coordinate frame and the semicircular arc.

The Co-ordinate Frame

It has the shape of the rectangle, which is 25mm high, 206mm

wide and 236mm low. It is made of light weight aluminium and has four feet which connect it to the x-ray equipment, the CT and MR adapters as well as the May field frame fixation. The rectilinear coordinate system has its origin outside the frame, superior, posterior and lateral to it on the right side. It conforms to the X, Y and Z noman- clature used in CT and MRI scanning. The coordinate scales on the frame indicate positive values only, Thus for instance, confusion between right and left is precluded.

The frame is provided with six posts for fixation to the patient. The frame is provided with two earplugs, which are used as an aid for the orientation of the frame on the patient's head, prior to its fixation.

The Stereotactic Arcs.

The semi circular arc is composed of the arc itself, to axes with side rings and sliding instrument carrier. Its design is such that when using needles, electrodes or other instruments whether the standard Leksell length, or active part will always enter in the center of the arc, i.e. at the selected target point. This gives the surgeon the unique opportunity to place the burr hole anywhere on the skull. The arc is positioned on to the coordinate frame according to the target coordinates obtained from a MR or CT scan or from an X-ray film.

C.T. PERFUSION

C.T Perfusion is used to measure the heamodynamic properties such as CBf,CBv,MTT and time to peak(TTP).The principle of these method involves continous acquisition of data from a particular area performed during the first passage of a ppherally administrated contrast meadium through the cerebral vasculature.From the images obtained in this way, the shortterm changes in the density of the brain tissue can be

presented in the form of a time density (TD) curve for every volume element of the slice being examined

Analysis of the resultant TD curve is used to calculate the blood hemodynamic properties.

CT Perfusion, commonly use either a non-diffusible tracer (iodine) or diffusible tracer (xenon).

Xenon CT used as an accurate technique in CT perfusion but involves inhalation of stable xenon by the patient, with possible side effects and necessitates expensive and complex equipment.

Non-diffusible tracer (iodine) is the commonly used tracer in PCT. In this two region of interest (ROI) are placed on a non-contrast enhanced image and contrast enhanced CT scan was done. This will produce a TD graph. Later analysis of this TD curve with perfusion software produce hemodynamic parameters. MDCT Technique allows imaging different slices at the same time.

Knowledge of cerebral blood flow (CBF) alterations in cases of acute stroke could be valuable in the early management of these cases. Among imaging techniques affording evaluation of cerebral perfusion, perfusion CT studies involve sequential acquisition of cerebral CT sections obtained in an axial mode during the IV administration of iodinated contrast material. They are thus very easy to perform in emergency settings.

Viability of the cerebral parenchyma is dependent on cerebral blood flow (CBF). Complex autoregulation processes ensure the adjustment of regional CBF to local energetic needs, determined by the activity level of local neurons. CBF alterations are encountered in association with a variety of pathologic conditions, the most frequent being strokes. Cerebral infarcts occur when CBF values are <10 to 15 cc/[100 g x min], whereas penumbra, relating to reversible cerebral ischemia, happens with CBF between 15 and 20 cc/[100 g x min]. Present indications for a thrombolytic therapy rely on the time interval between the beginning of symptoms (inferior or superior to 3 hr) and the native cerebral CT findings. Knowledge of a quantitative map of

CBF, indicating the severity and potential reversibility of neuronal damages, would perhaps allow for the clinical use of the theoretical thresholds mentioned above; thrombolysis achieved when the penumbra prevails over the infarcted area might not only be more profitable but furthermore might decrease the risk of intracranial bleeding. Different imaging techniques are now available to evaluate CBF, notably stable xenon CT and perfusion CT studies. Stable xenon CT relates to dynamic CT scanning during inhalation by the patient of a gas mixture containing stable xenon and oxygen. Alveolar xenon accumulation is measured end-tidally by a thermoconductivity analyzer and assumed to be equal to arterial xenon concentration. Stable xenon is a diffusible gas that progressively pervades cerebral blood and neurons on a well-balanced basis. Its radiopacity relates cerebral increase of CT units on successive cerebral CT sections to an increase of the parenchymal concentration of stable xenon. Stable xenon CT data analysis is realized through applying the equilibrating indicator model, which relies on the Fick principle:

$$\frac{dQ(t)}{dt} = \text{CBF} \cdot (C_a(t) - C_v(t)) \quad (1)$$

where $C_a(t)$ and $C_v(t)$ designate the instantaneous arterial and venous concentration of the indicator at time t , whereas $Q(t)$ designates the amount of indicator in the local vascular networks in connection with time t . The equilibrating indicator model supposes a balance between the venous, $C_v(t)$, and the parenchymal, $C_b(t)$, concentrations of the indicator:

$$C_b(t) = \lambda * C_v(t)$$

where λ relates to the cerebral parenchyma-blood partition coefficient of the indicator. Combination and development of equations 1 and 2 result in the supporting equation of the equilibrating indicator model:

$$C_b(T) = \lambda \cdot K \cdot \int_{t=0}^T C_a(t) \cdot e^{-K \cdot (T-t)} \cdot dt$$

with

$$K = \frac{CBF}{\lambda}$$

Stable xenon CT necessitates excellent collaboration from the patient, as well as specialized and expensive equipment. It may occasionally be responsible for a decrease in the respiratory rate, headaches, nausea, vomiting, and convulsions.

Perfusion CT studies involve sequential acquisition of cerebral CT sections achieved on an axial mode, during the IV administration of iodinated contrast material. Perfusion CT data consist of contrast enhancement profiles obtained at each pixel, the latter relating linearly to the time-concentration curves of the contrast material. Analysis of these curves is realized according to the central volume principle. The regional cerebral blood volume map is inferred from a quantitative estimation of the partial volume averaging effect, completely absent in a reference pixel at the center of the large superior sagittal venous sinus:

$$CBV = \frac{\text{area under the curve in a parenchymal pixel}}{\text{area under the curve in the reference pixel}}$$

where CBV stands for cerebral blood volume with a correction factor to take into consideration that iodinated contrast material is restricted to the plasma phase of the blood. The impulse function and the related mean transit time maps result from a deconvolution of the parenchymal time-concentration curves by a reference arterial curve. Finally, combination of cerebral blood volume and mean transit time at each pixel leads to a CBF value

$$CBF = \frac{CBV}{MTT}$$

where MTT stands for mean transit time.

The imaging protocol for stable Xe gas perfusion

six contiguous 10-mm sections located on the cerebral hemispheres. This protocol was repeated two times before and four times during an inhalation of a gas mixture containing 28% stable xenon, 25% oxygen, and 47% air, to a total of 6 min. The obtained data were analyzed on a computer with a post-processing software.

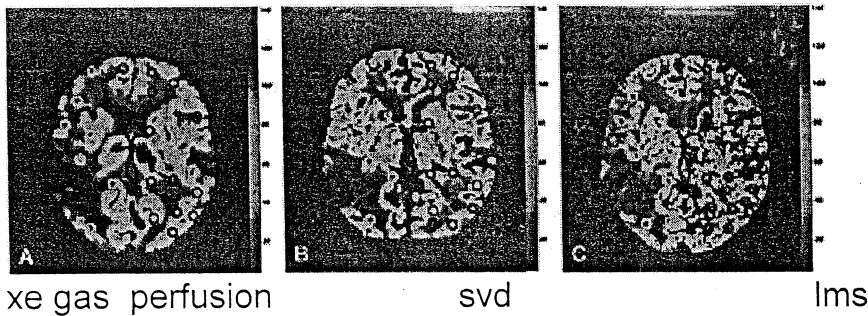
The imaging protocol for perfusion CT

two adjacent levels among the six examined ones were selected at the level of the basal nuclei. Twenty-five successive 10-mm CT sections were obtained every 2 s at these two adjacent levels, with a total acquisition time of 50 s. Acquisition parameters were 80 kVp and 200 mAs. CT was initiated 2 s before the IV administration of 1.3 cc/kg of iohexol by means of a power injector at a rate of 5 cc/s. The delay before injection of the contrast material allowed for the acquisition of baseline images without contrast enhancement. This perfusion CT protocol involved an additional radiation dose of 291 mSv. Regarding the stochastic effects of radiations, this dose, once redistributed on the entire cerebral volume, amounted to 29 mSv, which is inferior to the dose of a standard cerebral CT examination (40–60 mSv).

Stable xenon CT results and perfusion CT data were transferred to a workstation for computer processing and comparison.

Perfusion CT rely on the central volume. There are two different deconvolution methods: the singular value decomposition (SVD) and the conventional least mean square (LMS) methods. In both technique, the reference artery was automatically selected by the software (thus avoiding interobserver variability) as the pixel with the shorter time to peak on the corresponding time-concentration curve in the area drawn by the operator, either around the anterior cerebral artery (ACA) or around the middle cerebral artery (MCA).

Perfusion CT studies have the same complications as that of CT angio, whereas stable xenon CT examinations were responsible for a decrease in respiratory rate.



Stable xenon CT relies on the equilibrating indicator model, which indicates that the balance of concentrations of a diffusible indicator between blood and cerebral parenchyma is more or less rapidly realized in the various cerebral areas, according to the importance of the corresponding local CBF. Knowledge of the blood (or alveolar) and parenchymal concentration curves leads to the balance rates and to the CBF values for each pixel. On the other hand, perfusion CT data analysis is realized through the central volume principle. The latter describes the behavior of an iodinated contrast material bolus crossing the cerebral capillary networks, with subsequent modifications of contrast enhancement profiles. Iodinated contrast material is limited to blood vessels, at least at first pass and in healthy cerebral parenchyma.

Perfusion CT studies of CBF are not time-consuming and are well tolerated. They verify classical contraindications regarding the IV administration of iodinated contrast material. On the other hand, the IV administration of nonionic iodinated contrast material can reasonably be achieved even in patients who have suffered acute stroke.

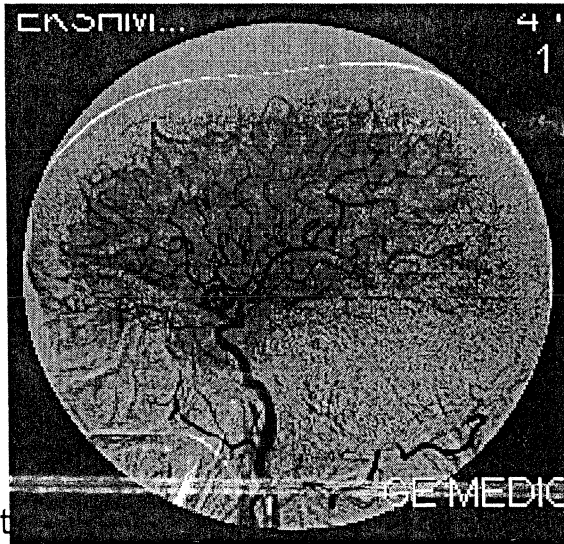
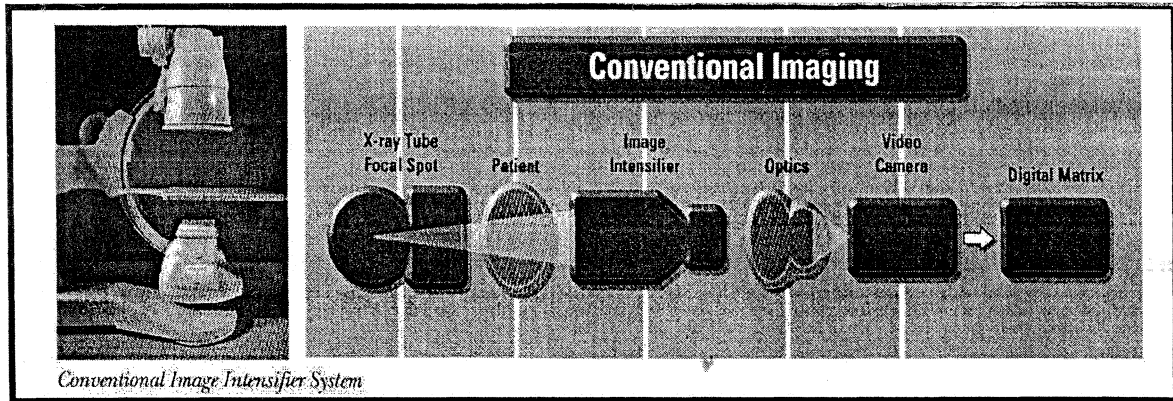
Perfusion CT studies do not require any special equipment. Even with a multi-detector CT unit, perfusion CT coverage of the brain is inferior to that of stable xenon CT, because only two 10-mm cerebral CT sections can be examined. This is mainly because of the much quicker kinetics of iodinated contrast material compared with that of stable xenon CT. However, the purpose of our study was to compare the corresponding sections in perfusion CT and stable xenon CT and not brain coverage.

Perfusion CT studies were performing at low kVp (80) rather than high kVp (120), allowing for a statistically significant increase in contrast enhancement. This results from the more important perfusion of gray matter compared with that of white matter. Moreover, after contrast enhancement, radiographic interaction with soft tissues at 80 kVp relates to the photoelectric effect, due to the 33 keV K-edge of the iodine included in the contrast material. On the other hand, radiographic interaction before contrast material administration is mainly due to the Compton effect. Although the use of 80 kVp involves a lower photon flux, it does not result in a statistically significant increase in noise, allowing 80 kVp images to be used in perfusion CT analysis. Finally, performance of perfusion CT examination at 80 kVp, keeping milliamperes constant, lowers the radiation dose by a factor of 2.8.

With stable xenon CT, there is no concern regarding the choice of a reference artery, because the arterial time-concentration curve is assumed to be the same as that measured in the end-tidal breathed air. However, this leads to other pitfalls, notably for patients with chronic respiratory disease or with cardiac or vascular shunts.

DIGITAL SUBTRACTION ANGIOGRAPHY(DSA)

DSA is the generic term for any digital radiographic method for implementing subtraction. DSA offers the advantage of image manipulation, plus the instantaneous viewing of the images, and sensitivity to low contrast images, with smaller amounts of contrast medium injected, are all desirable technical considerations. DSA has many applications for low-contrast examinations. DSA can be very helpful in patients who are at risk for a high rate, high volume contrast examination because less contrast media is needed to obtain diagnostic images. Film framing rates on the order of 3 to 4 fps is desirable.



The various subtraction techniques are;

1. MASK SUBTRACTION
2. DUAL ENERGY SUBTRACTION
3. TIME INTERVAL DIFFERENCING
4. TEMPORAL FILTERING

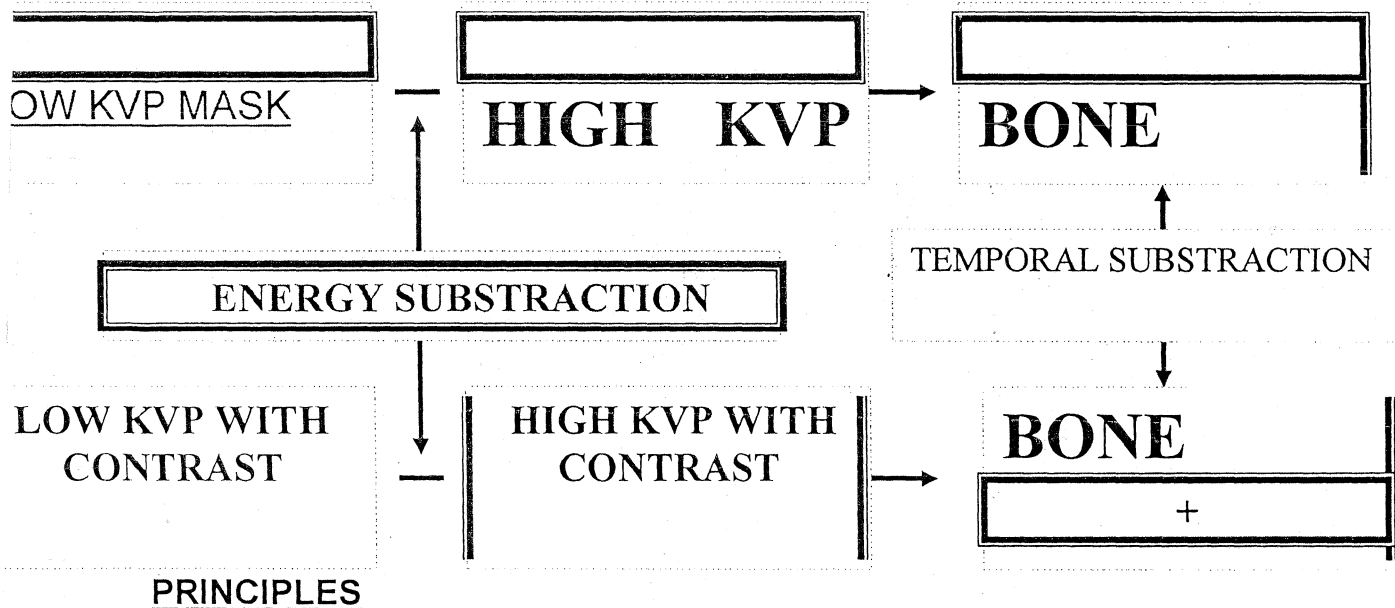
MASK SUBTRACTION

Mask subtraction is the DSA technique, which closely resembles film screen angiography. The patient is prepared and catheter is placed under fluoroscopy control. Contrast media is then injected and exposures made, starting before contrast media is

expected to arrive the vessel of interest and extending past the time when it is expected to have cleared the vessel. Subtraction of pre contrast mask image from contrast containing image is the basis of this mask subtraction technique.

DUAL ENERGY SUBTRACTION

In dual energy subtraction, two images are taken in a very short period, during which time there is no change in the patients. These two images are obtained by making exposures with different x-ray energy spectra. E.g.; from a high kvp exposure and a low kvp exposure.



The differential attenuation between bone and soft tissue at low kvp is much greater than their differential attenuation at high kvp. The bone contrast in the high kvp film is reduced much more than the soft tissue contrast. In the regions that do not contain bone, the soft tissues cancel but bone containing regions do not cancel completely. Thus the final images consists of only bone plus the inevitable noise.

PROBLEMS WITH DUAL ENERGY SUBTRACTION

All the problems associated with radiographic imaging in general and DSA in particular are still present. Additional problems that may be encountered are;

1. The high kvp image still has some bone with in it when the soft tissue densities are eliminated by subtraction, some of the bone densities are eliminated by subtraction, and some of the bone density is also removed.
2. A more complex x-ray machine is required, as an x-ray generator capable of switching kvp and mAs rapidly is needed to overcome problems with anatomic motions.
3. Due to differential beam hardening of the two beams in the different thickness of bone and S/T in the body there is improper subtraction of bone.
- 4.

OTHER DUAL ENERGY SUBTRACTION

1. K-SHELL SUBTRACTION
2. HYBRID SUBTRACTION

TIME INTERVAL DIFFERENCING

This has seen some application in cardiology. The technique is closely related to simple mask subtraction. In simple mask subtraction an early image is chosen as a mask and this mask is subtracted from each succeeding image of the series of subtracted images. In TID technique, a new mask is chosen for each subtraction and each subtracted image is the difference between images separated by some fixed interval of time.

TEMPORAL FILTERING

In this technique, several of the early frames are added to obtain a mask and several of the later frames added to obtain a contrast containing image and the two are subtracted. This reduces the image noise with out an increase in tube mA being required for the same. The obvious advantage of the technique is the reduced tube loading required as compared to single mask subtraction. The disadvantage is that a large number of individual frames must be handled for temporal filtering.

There are two catheterization laboratories; one equipped with seimens Polydoros 100 & DIGITRON 3 and other with GE Advantx DLX-LCV

Equipment Parameters:

Seimens Polydoros 100

Seimens Polydoros 100 is a high frequency X-ray generator with microprocessor control. The equipment is a bi plane C-arm System with cine attachment. It has DSA provision to.

DIGITRON 3

Digitron 3 is a fully digital DSA unit designed for cardiac work as well as routine intraarterial and intravenous subtraction angiography. It is a real time image acquisition and processing system.

Image post processing features are mask selection, electronic shutter, Grey-scale inversion, image re-registration, horizontal/vertical (pixel shift), image zoom, land scape road map, edge enhancement, image integration up to sixteen images, ROI's, image filtering, cardiac analysis etc.

GE Advantx DLX-LCV **SYSTEM SPECIFICATIONS**

MAINCOMPONENTS

1. X-RAY TUBE
2. IMAGE INTENSIFIER
3. GENERATOR
4. TV -CAMERA
5. COMPUTER CONSOLE

X-ray tube

MAX 6 HEXAGRAPH(SIX FOCAL SPOT)
 3-FILAMENTS → 1.1, 0.6, 0.3
 BIASED FOCAL SPOTS → 0.30b, 0.20b, 0.15b

Image intensifier

TYPE	SIZES
DUAL FIELD	6"
TRIPLE FIELD	9"
QUADRUPLE FIELD	12" / 16"

X-ray generator – LFX generator

100KW, 12 PULSE
 ROTOR CONTROLLER- TIRC 10,800 rpm

HV TRANSFORMER

Pulse bias tank – control grid

PICK UP TUBES

Vidicon
 Primicon, saticon (Philips)
 Plumbicon, newicon, chimicon

PARAMETERS SELECTION

Dose, Focalspot, FOV, MA, KVP

TRIAL → System estimates the exposure factors

Kvp → 60-120

FRAME RATES → 12.5-25 F/S-CINE

VASCULAR MAX → 8.3 F/S

DOSE VASCULAR CARDIAC

A	100	4
B	250	7
C	500	7
D	1000	15

FLUORO DOSE

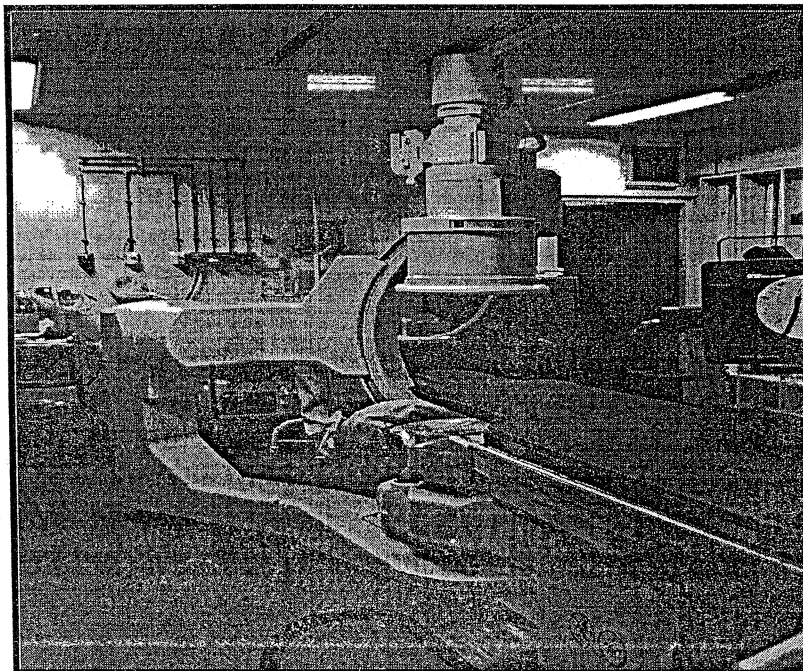
MAX KV –120 KVp, CONTROLLED ABC

DOSE 10R/M - max dose at tabletop

Modes → low – med -high

. The system offers facilities such as last image hold (LIH), auto injection of contrast, edge enhancement, noise filtering, roadmap etc, Image processing include different subtraction modes, pixel shift, gray scale inversion, remasking, image filter, image integration, integrated mask, maximum opacification, temporal scale, landscaping, shutter, stenosis analysis, cardiac analysis and annotation.

L-ARM



CONSOLE



PART A : Equipments & Accessories

DSA

1. Operation

2. Components of the Digital Imaging System

3. Methods of manipulating and archiving the DSA Image

1. Operation

DSA stores images electronically and with the aid of a computer, subtracts images in real time. Real-time digital imaging permits 30 images per second. The images can be manipulated to change the window levels and contrast, a function that is invaluable in low contrast examinations. The DSA system operates at a fluoroscopic milliamperage level on the average of 10 to 20 mA, depending on the equipment being used. The image is processed as the run is being performed, so that subtraction image can be viewed and the exposure terminated once the area of interest is viewed. What occurs is that the image from the image intensifier is converted from analog

data into digital information and processed by the computer for display and storage.

2.Components of the Digital Imaging System

A television image chain is one of the most important links in the digital system. The light produced by the image intensifier is proportional to the amount of remnant x-ray photons. The image intensifier (II) input phosphor is coated with an x-ray sensitive phosphor, such as cesium iodide (CsI). The television chain produces an electronic video signal at the output phosphor in proportion to the amount of light, but exceedingly intensified. The television camera (Plumbicon, CCD, or Saticon) then forms an image by electronically scanning a target or semiconductor. The video signal produced is sent to the digital processing unit Digital Subtraction Angiography System

Characteristics needed in the TV chain of a DSA system include low lag and low noise. Plumbicon TV camera tubes are usually used for DSA because of their intrinsic low lag. Excessive lag would produce "ghosting" Ghosting occurs when the image viewed actually contains information belonging to the previous images. Studies can be greatly affected by this artifact. The signal to noise ratio (SNR) is the signal voltage ratio to the noise voltage. Conventional fluoroscopy requires a 200:1 SNR. Digital requires about 1000:1 SNR. Generally noise can mask or hide the signal. Because subtraction studies are usually performed on low -contrast studies it is especially important that the amount of electronic noise must be kept to a minimum, much more so than in standard film screen radiography or fluoroscopy. Digital radiography images can be degraded by noise from the fluctuations in x-ray photon intensity (quantum mottle) and noise from the television chain (electronic noise). Techniques have been devised to help overcome the problem of noise in DSA . Frame integration averages together several frames, thereby diminishing the effects of electronic and quantum noise.

A light diaphragm or filtering device is used to control the amount of light from the image intensifier that reaches the television camera tube. Without the use of apertures, either not enough or too much light (saturation) is allowed to reach the television camera.

The analog-to-digital converter (ADC) is the component that converts the video images from the TV chain into digital form. The video signal has a voltage that varies during the scan, representing the number of x-ray photons that are received by the image intensifier. Measurement of the voltage of each individual pixel of every line results in conversion of the video image into a digital image. So it is easy to see that the size of the matrix will greatly influence the image resolution.

Television scan modes are a component of the actual television monitor and are of various types.

Interlaced: the 525-raster or line frame is scanned in 1/30th of a second. This means each 262.5 line field is scanned every 1/60th of a second. The interlaced mode reads the information in two separate fields and then combines, or interlaces, the information. This mode is adequate to avoid the perception of flicker. The human eye can detect flicker on a television monitor below 50 pulses per second. The major drawback of interlaced mode is that certain images are discarded, so the patient receives some unnecessary x-ray exposure.

Progressive: Every line is read in order, rather than in field - interlaced manner. This method involves the TV camera scanning being paused while an exposure is made so that it is not as rapid as interlaced but eliminates motion.

Slow: This method is inadequate for real - time DSA. It has application in the 1024x1024 pixel digital fluoroscopy imaging, with a 1050 line TV frame. The slow scan mode is limited to 7.5 frames per second. Real-time continuous digital fluoroscopy requires 30 frames/second. Pulsed digital fluoroscopy typically allows for only 1 to 6 images per second but could not use the slow scan mode. These images have less noise but require high mA.

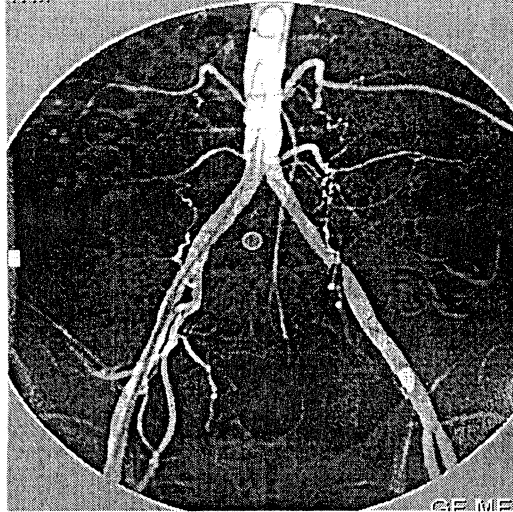
3. DIFFERENT METHODS FOR IMAGE MANIPULATION

Window and Level Adjustments:

DSA allows for post-procedural manipulation of images. The simplest way to manipulate an image is to vary the window level, which

controls the density or vary the contrast by adjusting the window width.

LANDSCAPING



A small amount of the mask image is added back to the subtracted image, but not so much that the subtracted image is overwhelmed. Landscaping is usually done to assist the radiologist in preparation of invasive procedures.

Edge Enhancement:

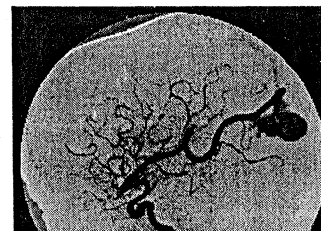
Edge enhancement is intended to increase the visibility of small structures with moderate to high contrast. The basic goal is to discard most of the information about the large structures and keep the information relates to small structures. Edge enhancement is accomplished by subtraction of low pass filtered image from the original which yields an image in which edges and small structures remain. Unfortunately noise also prominent in this edge enhanced image.

NOISE SMOOTHING

BEFORE FILTERING



AFTER FILTERING



It is an attempt to decrease the visual prominence of noise so that low contrast objects of moderate to large size may be better appreciated. The technique operates by reducing the statistical fluctuation in each pixel by averaging the pixel with its closest neighbours. The visual prominence of noise is suppressed through this process. The disadvantage is that resolution is decreased.

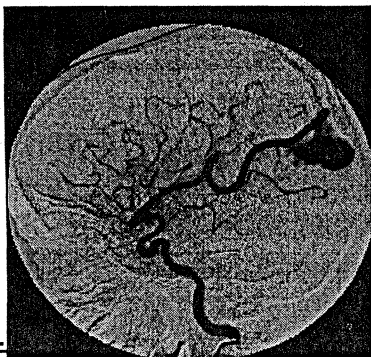
MASK RE-REGISTRATION

It's sometimes useful when there is yet patient motion between the time the mask is taken and the time that contrast material arrives in the vessel of interest.

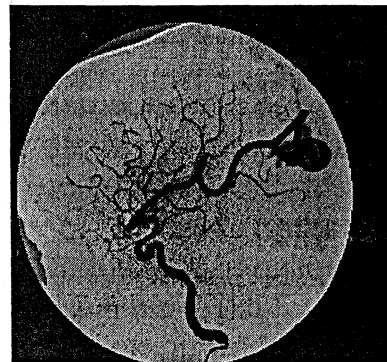
The re-registration of mask may be accomplished either;

1. MANUALLY (VIEWER-CONTROLLED)
2. AUTOMATICALLY (COMPUTER CONTROLLED)

BEFORE RE-MASKING



AFTER RE-MASKING



The patient is 3-dimensional and mask re-registration operations are effective only for motions confined to a plane;

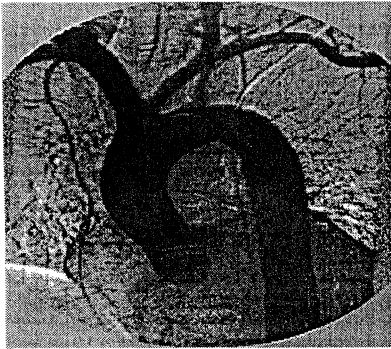
- Upper structures project on to different lower structures when the patients roll on to one side as compared to being flat on his

back. Re-registration cannot exactly compensate for this type of motions.

- Motion of structures within the patient (especially bowel gas and larynx) also changes the relative positions of prominent structures, and again re-registration cannot compensate exactly.

IMAGE ZOOM :

WITH OUT ZOOM



WITH ZOOM



Image zoom allows for magnification of an area of interest but however it does not increase visibility of detail.

Image Reregistration or Remasking :

These two tools can be helpful to reregister or remask the image, which is helpful in eliminating motion artifacts. Misregistration, which occurs when the mask and the series images are not aligned, can be corrected through this process.

Pixel Shifting:

This is a post processing technique. Mainly two types of pixel shifts are there;

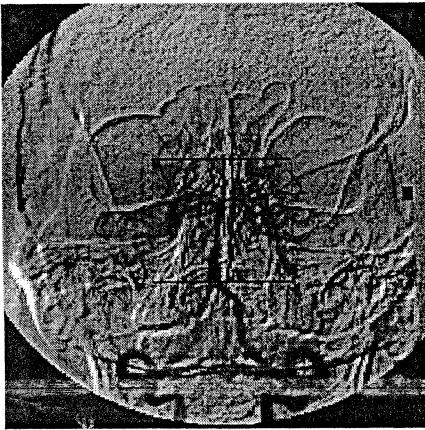
1. AUTOMATIC PIXEL SHIFTING
2. MANNUAL PIXEL SHIFTING

AUTOMATIC PIXEL SHIFTING

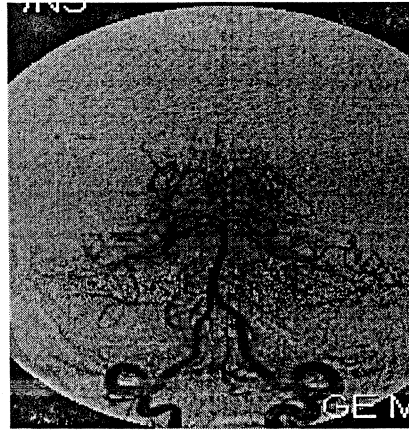
The automatic pixel shift is used to perform the registration on either the left or right side of the image using a ROI positioned by the operator; the calculation in the ROI shall always be limited to the useful part of the image. The box

should be positioned so that it covers a ROI containing vessels and subtraction artifacts, which disturb the visibility of vessels. The calculated shift would be more effective if the edges of the horizontal and vertical bone structures are present in the ROI. If the shift is greater than 7.5 pixels, the system unable to register the image and ROI is needed to reposition.

WITH OUT PIXEL SHIFTING



AFTER AUTOMATIC PIXEL SHIFTING

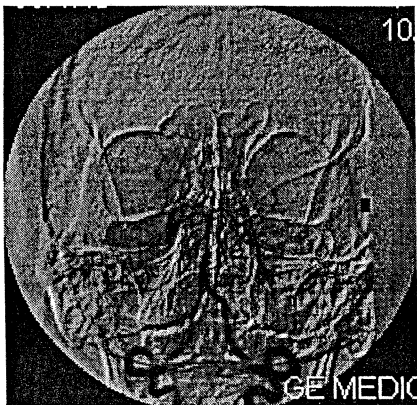


MANUAL PIXEL SHIFTING

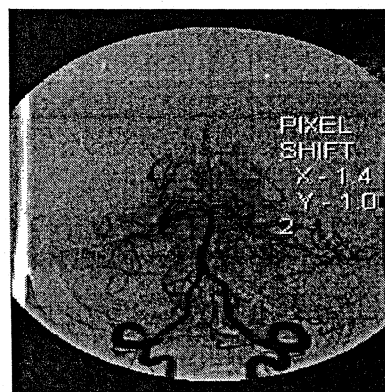
The joystick can be used to shift the image to reduce the registration error. The manual pixel shift adjustments let you to register subtracted images when movement occurred between mask selection and image acquisition.

The pixel shift is equal to 0.1 pixels in 512 matrix and 0.2 pixel in 1024 matrix.

WITH OUT PIXEL SHIFT

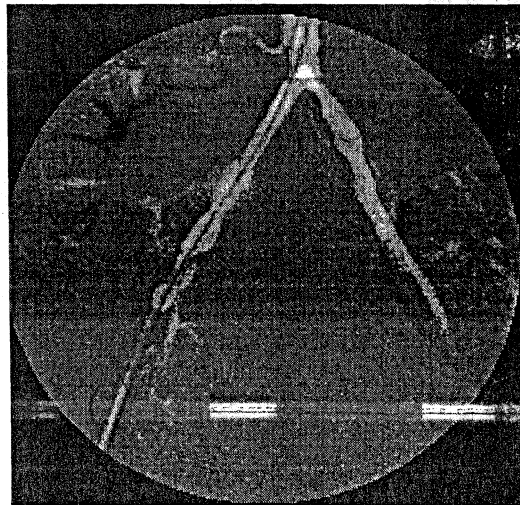


WITH PIXEL SHIFT



Road Mapping:

Road mapping is a means of tracking or tracing with small amounts of contrast material injected. The vascular path is remembered by the system, and a print can be made or it can be subtracted from a image with no contrast media. Road mapping that provides a continuous real-time DSA image is widely used and is extremely helpful for angioplasty, embolization and difficult or superselective procedures. The catheter, as well as the contrast filled artery, appears on the monitor, and acts as a guide for the catheter.



MAXIMUM OPACIFICATION

Lets the operator view the maximum density reached by each pixel in a range of images. This function is useful to show on a single image the vessel opacified at their higher intensity.

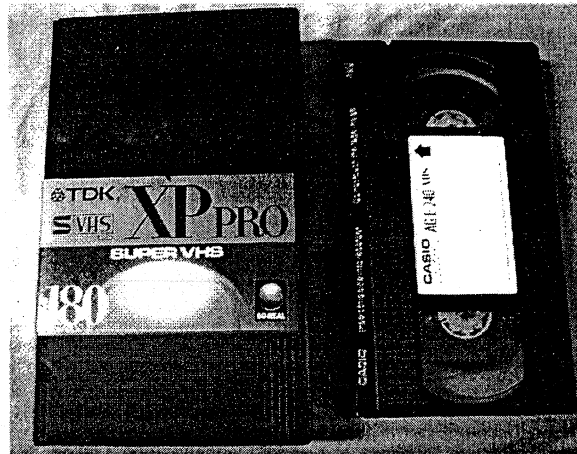
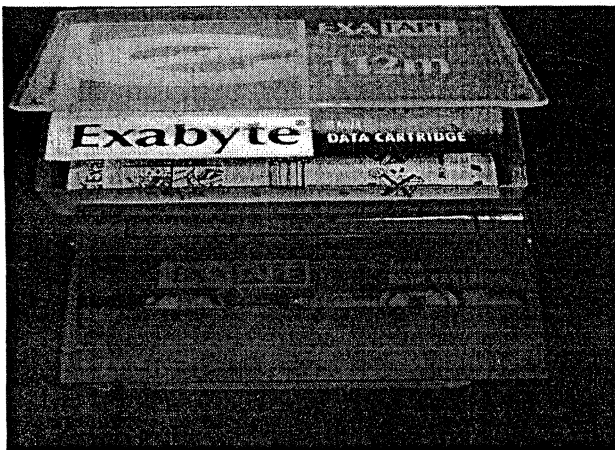
INTEGRATED MASK

It creates an average image from a range of images; the resulting image is used as a mask.

INTEGRATED IMAGE

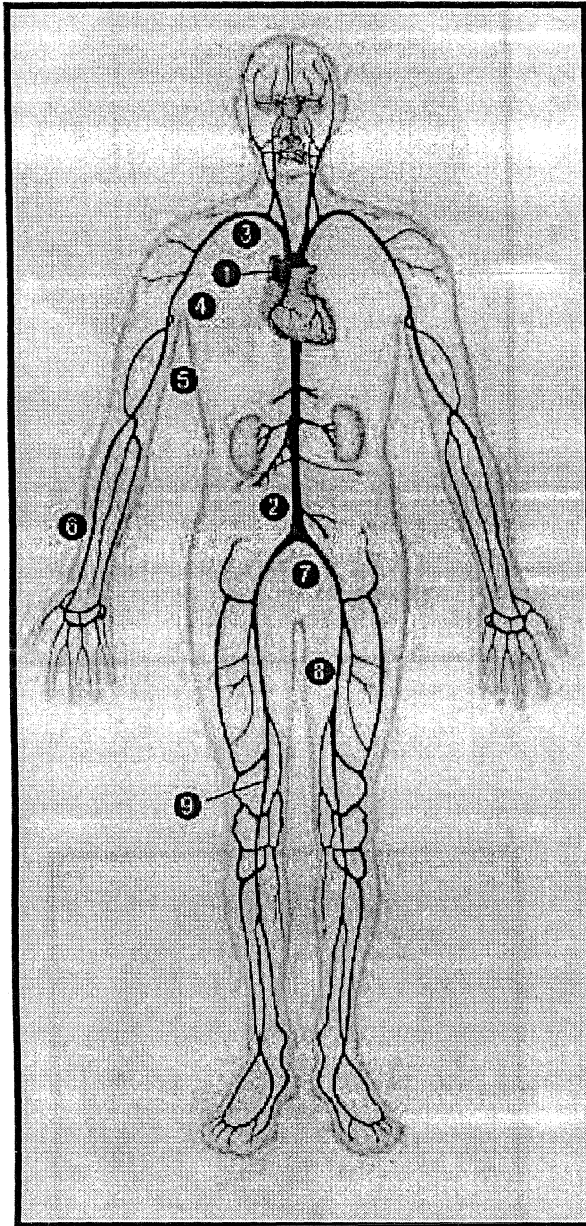
It creates an average image from a series of images. The improvement in SNR makes it suitable for poorly injected sequence

Archiving of DSA Images:



This is possible on video tape recorders and digital disk. Final hard copies of digital images are printed on either multiformat or laser cameras.

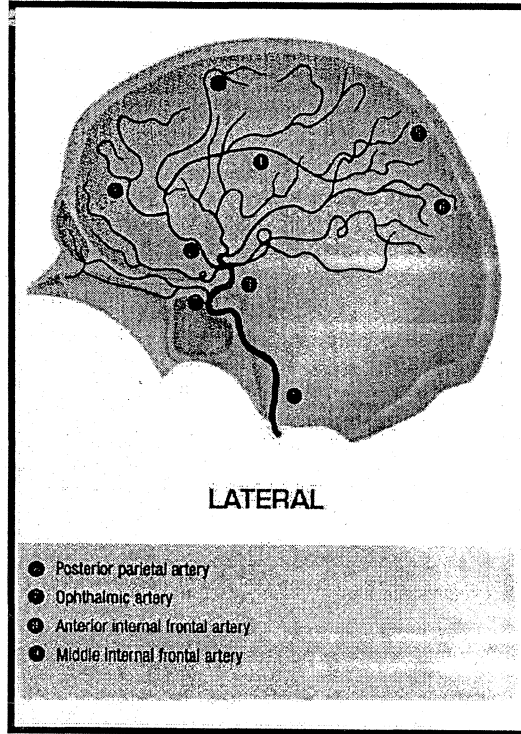
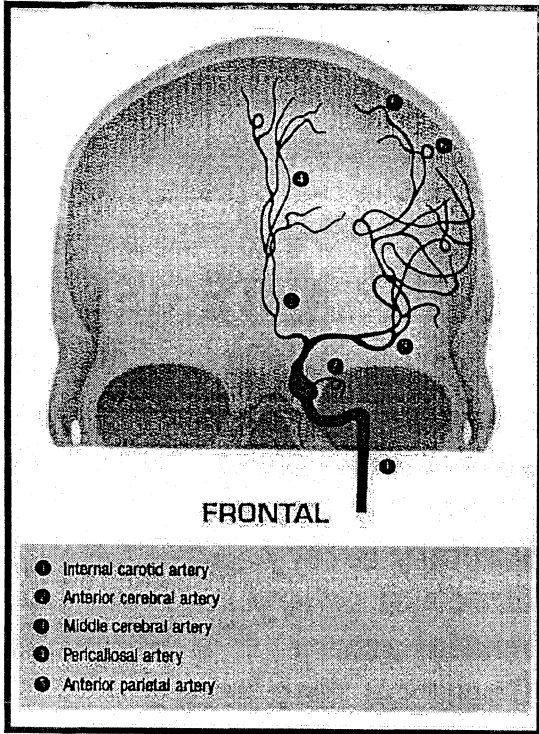
HUMAN CIRCULATORY SYSTEM



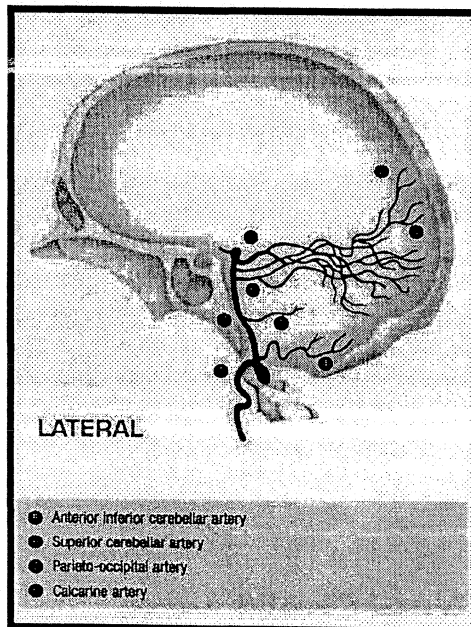
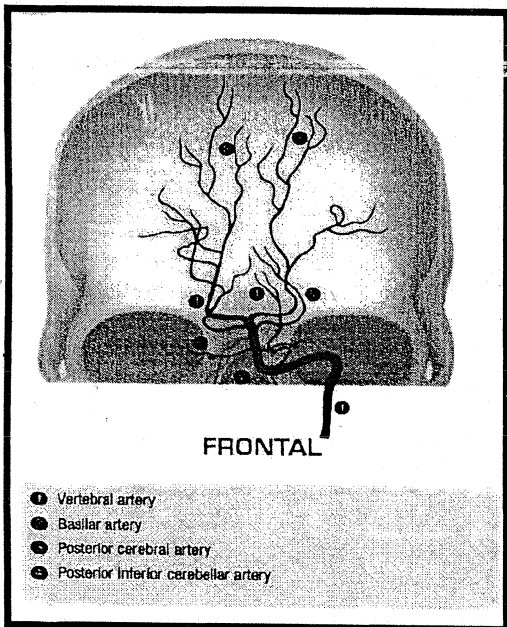
ARTERIES

1. Ascending aorta
2. Abdominal aorta
3. Subclavian artery
4. Axillary artery
5. Brachial artery
6. Radial artery
7. Common iliac artery
8. Femoral artery
9. Popliteal artery

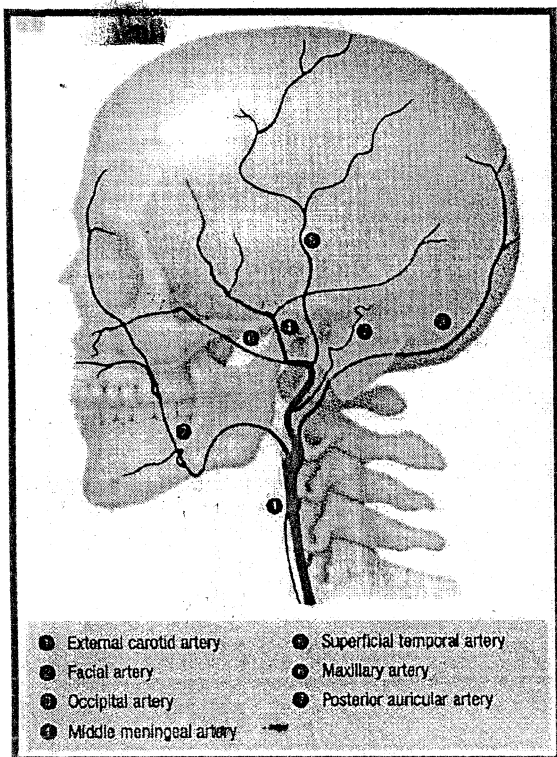
Internal Carotid Artery



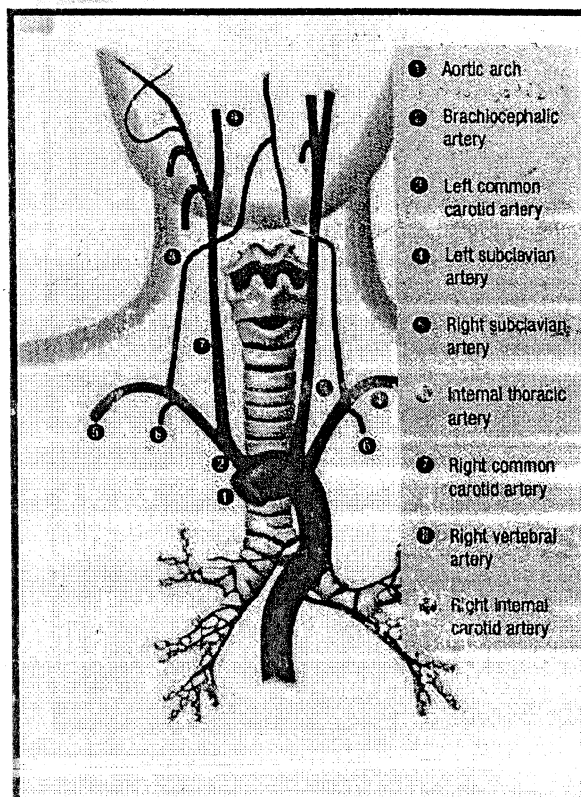
Vertebral Artery



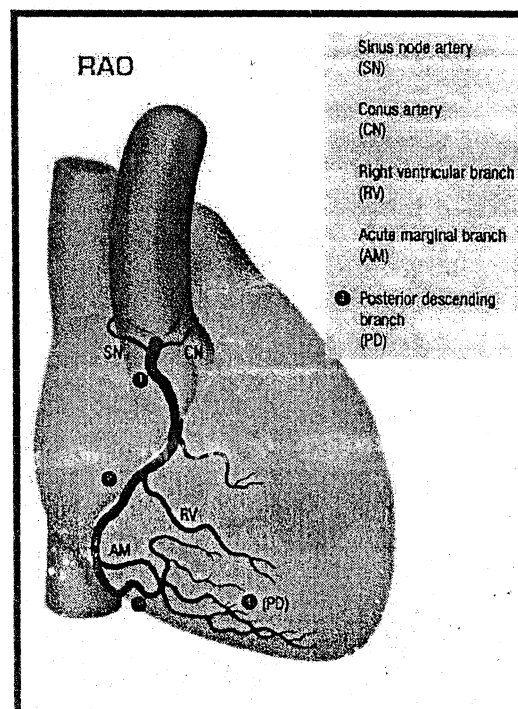
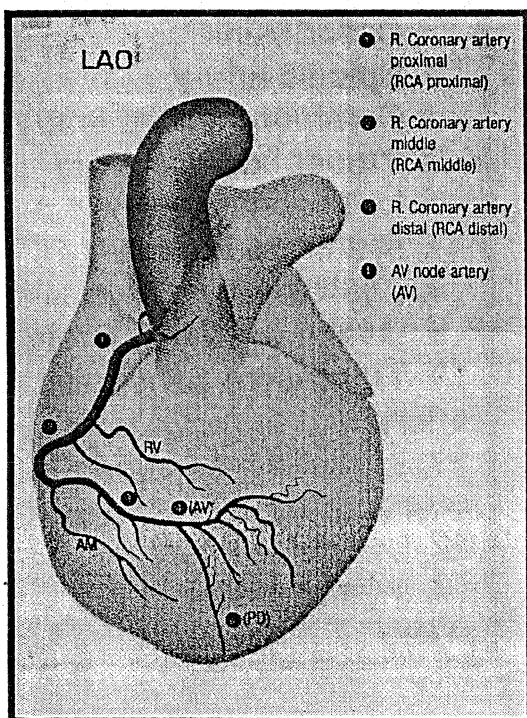
External Carotid

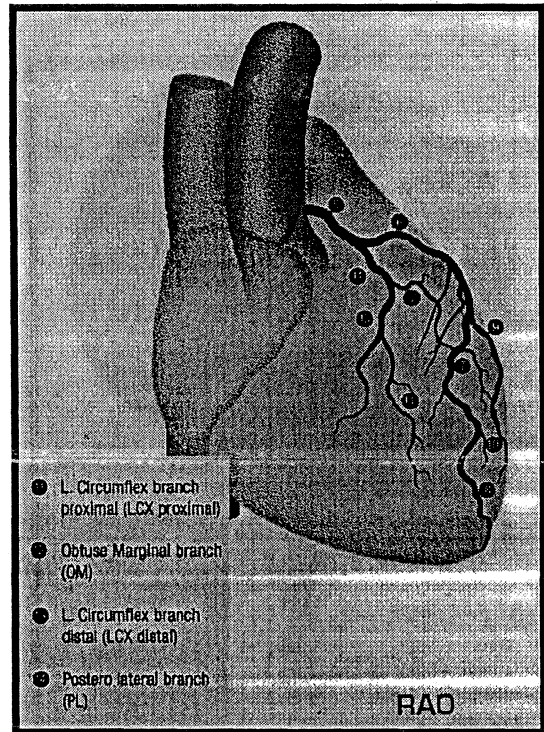
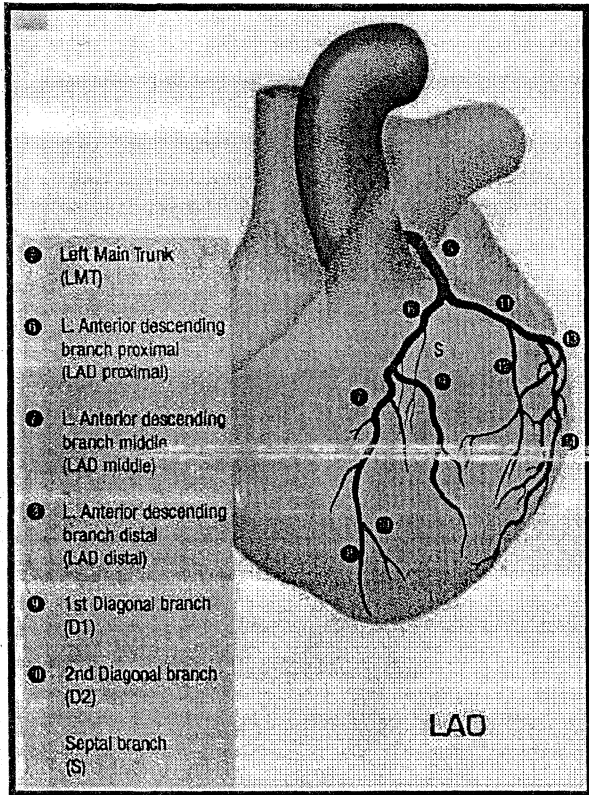


Neck vessels

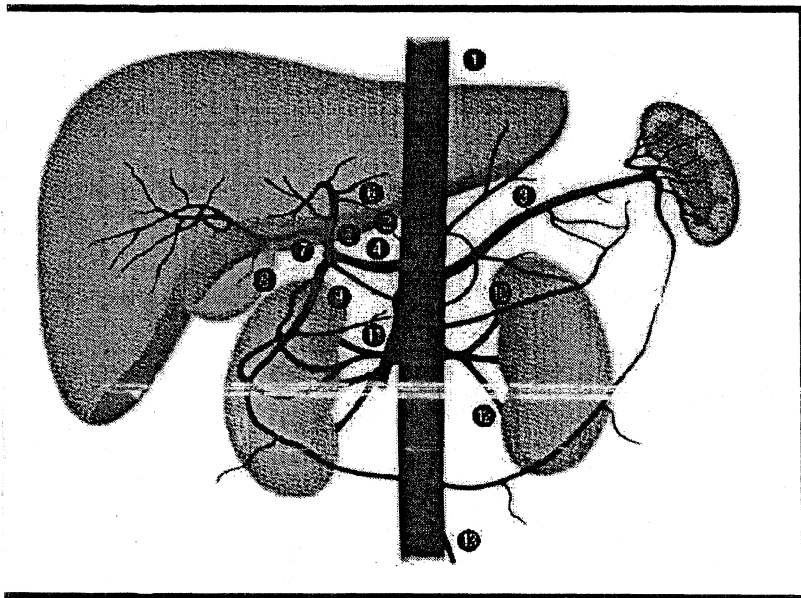


Right Coronary Artery



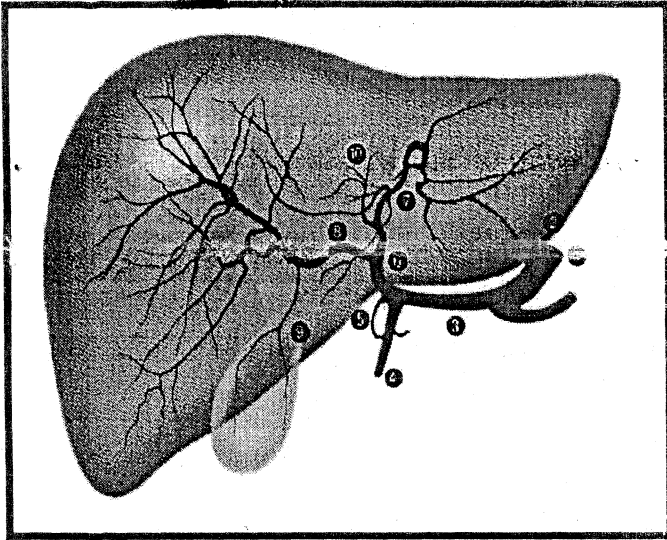


Abdominal



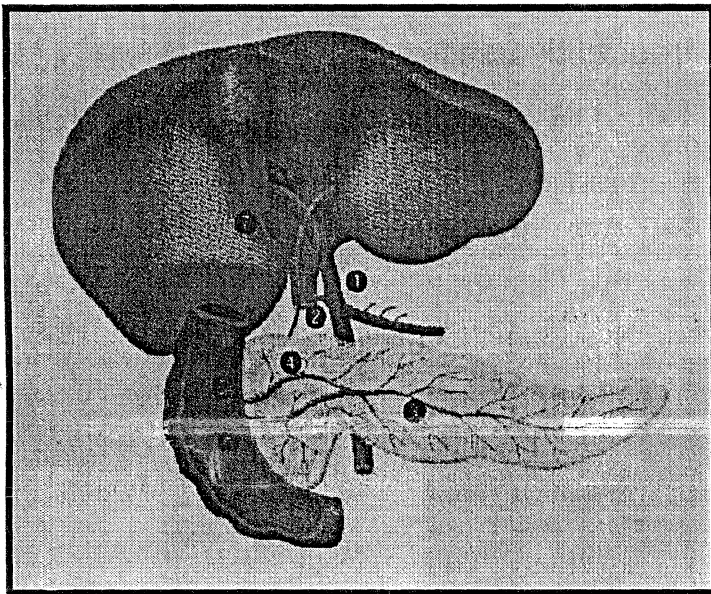
1. Abdominal aorta
2. Celiac trunk
3. Splenic artery
4. Common hepatic artery
5. Proper hepatic artery
6. Left hepatic artery
7. Right hepatic artery
8. Cystic artery
9. Gastro duodenal artery
10. Transverse pancreatic artery
11. Superior mesenteric artery
12. Left renal artery
13. Inferior mesenteric artery

Hepatic Arteries



1. Celiac trunk
2. Left gastric artery
3. Common hepatic artery
4. Gastro duodenal artery
5. Right gastric artery
6. Proper hepatic artery
7. Left hepatic artery
8. Right hepatic artery
9. Cystic artery
10. Middle hepatic artery

Portal Vein and Pancreas



1. Portal vein
2. Common bile duct
3. Pancreatic duct
4. Accessory pancreatic duct
5. Lesser duodenal papilla
6. Greater duodenal papilla
7. Cystic duct

INTRODUCTION TO CATHETER TECHNIQUE

The basic technique of arterial catheterization is applicable to veins also.

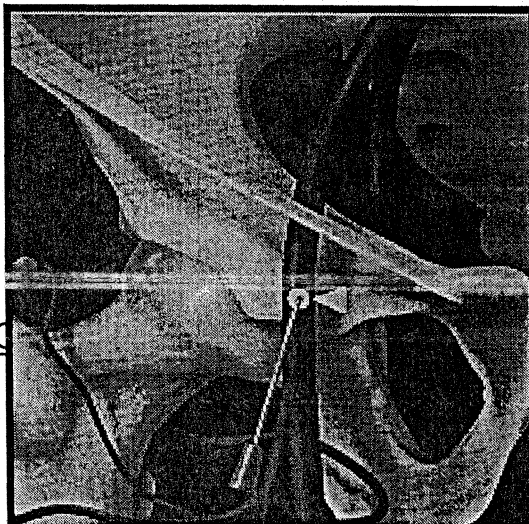
PATIENT PREPRATION

1. The patient will need to be admitted to hospital, as careful preparation before and observation for 24 hours after the procedure will be required.
2. The arrangements should be made to discontinue anticoagulants.
3. The radiologist may wish to see the patient on the ward prior to the examination. This will be the time when:
 - a. The procedure is explained and questions answered
 - b. Informed consent is obtained
 - c. The patient is examined, with special reference to blood pressure and peripheral pulses as a base line for post angiographic problems.
4. The puncture site is shaved.
5. Pre medication - pethedine
phenergan

PUNCTURE SITES

1. Femoral artery – the most frequently used.
2. Axillary artery.
3. Brachial artery
4. AORTA- by using Teflon sleeved needle for trans lumbar aortography the sleeve can be directly antegradely or retrogradely in the aorta.

FEMORAL ARTERY PUNCTURE

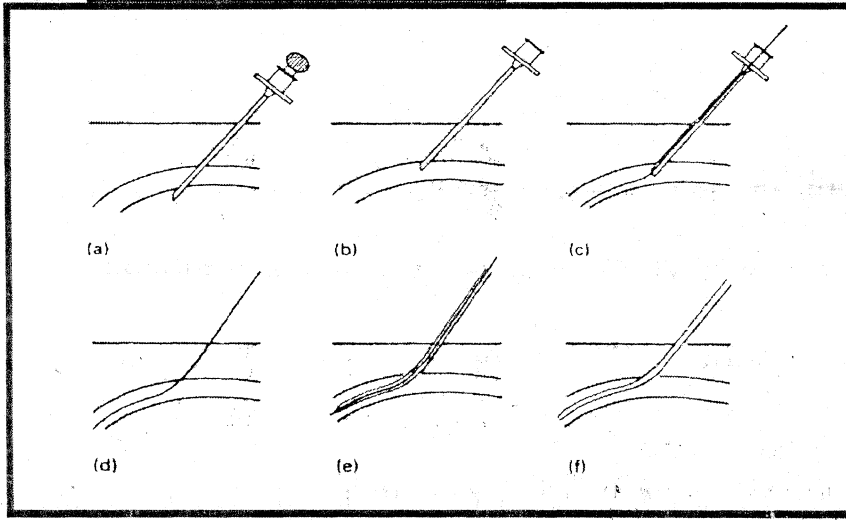


CONTRAINDICATIONS

1. Severe aortic, iliac or femoral artery arteriosclerosis.
2. Blood dyscrasis.
3. Femoral artery aneurysm.
4. Marked tortuosity of the iliac vessels may preclude further advancement of the guide wire or catheter. In such a case axillary artery puncture or direct puncture, e.g. of a carotid artery, will be necessary.

PUNCTURING TECHNIQUE

SELDENGIER METHOD



- a. Both walls of the vessel punctured.
- b. Stillette removed.needle with drawn so that the bevels is with in the lumen of the vessel and blood flows from the hub.
- c. Guide wire inserted through needle
- d. Needle with drawn, leaving guide wire in situ
- e. Catheter threaded over wire.
- f. Guide wire withdrawn.

TECHNIQUE

1. The patient lies supine on the x-ray table. Both femoral arteries are palpated and the easiest is selected for puncture. If the pulsation are equal, the right side is used

2. Before beginning, the appropriate catheter and guide wire are selected and their compactability checked by passing the guide wire through the catheter and needle.
3. Using aseptic technique, local anaesthetic is infiltrated either side of the artery down to the periosteum. A 5mm transverse incision is made over the artery to avoid binding of soft tissue on the catheter.
4. The actual point of puncture of the femoral artery must be considered. The femoral artery arches medially and posteriorly as it becomes the external iliac artery. Attempts to puncture the artery cephalad to the apex of the arch will result either in failure to puncture the artery or puncture of the artery deep in the pelvis at a point where haemostasis cannot be secured by pressure. Puncture distal to the arch may result in catheterization of the profunda femoris artery. Correct puncture is made at the apex of the arch with the needle directed to the 45° cephalad to the skin surface and slightly medially. The needle is then in direct line with the lumen of the artery .
5. The artery is immobilised by placing the index and middle fingers of the left hand on the either side of the artery and the needle is held in the right hand . The needle is advanced through the soft tissues until pulsation are felt transmitted through its tip. If the pulsations are localized to one side or other, the needle is either medial or lateral to the artery and should, therefore, be re-positioned. Both walls of the artery are punctured with a stab. The stilette is removed and the needle is slowly withdrawn until pulsatile blood flow from the end of the needle indicates a satisfactory puncture. poor flow may be due to:
 - a. Femoral vein puncture
 - b. The end of the needle lying sub intinally
 - c. Hypo tension → due to vasovagal reactions during the puncture.
 - d. Atherosclerosis
 - e. Polycythaemia
6. When good flow is obtained the guide wire is inserted through the
Needle and advanced up the artery using fluoroscopy to follow its travel. It should not be forced . When it is in the

descending aorta the needle is withdrawn over the guide wire keeping firm pressure on the puncture site to prevent bleeding. The guide wire is then wiped clean with a wet swab and the catheter threaded over it. The catheter is advanced up the descending aorta, under fluoroscopic control and when in a satisfactory position the guide wire is withdrawn.

6. The catheter is connected via two-way tap to a syringe of heparinised saline and after aspiration of air bubbles is flushed.
7. At the end of the procedure and the withdrawal of the catheter compression of the puncture site maintained for 10 min. if continued becomes a concern, consideration should be given to neutralizing the effect of heparin by giving protamine sulphate.

AXILLARY ARTERY PUNCTURE

Indications – as for femoral artery puncture but as this approach is associated with a higher incidence of complications it should only be used if femoral artery catheterization is unsuccessful.

Contraindications

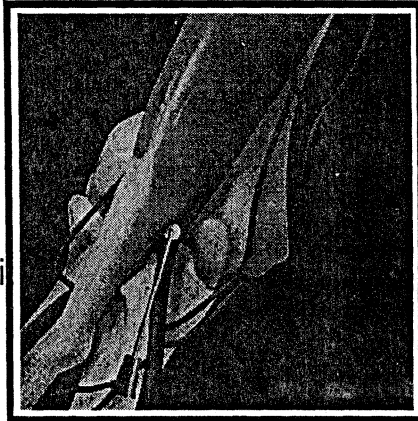
1. Atherosclerosis of the axillary or subclavian arteries.
2. Subclavian artery aneurysm

Technique

1. The patient lies supine on the x-ray table with his/her arm fully abducted. Peripheral arm pulses and blood pressure are compared in each arm. The puncture point is just distal to the axillary fold and using aseptic technique the area is infiltrated with local anesthetic.
2. A small incision is made in the skin, 1-2cm distal to the point of arterial puncture.
3. The needle is directed more horizontally than in the femoral approach and along the line of the humerus.

4. Following satisfactory puncture the remainder of the technique is as for femoral artery catheterization.

BRACHIAL ARTERY PUNCTURE



Left brachial artery is used for angiographic procedures.

INDICATIONS

- Angiography when both ilio femoral arteries severely diseased or occluded.
- Common femoral artery region infection or recent surgical graft.

CONTRAINDICATIONS

- Uncorrectable bleeding diathesis.
- Known occlusion of brachial artery.

•

GETTING STARTED

THINGS TO CHECK

- Check Brachial, Radial, Ulnar Pulses In Arm.
- Left arm is used because only cross-vertebral artery on way to aortic arch.
- On the right side cross right vertebral and carotid ostia as well as left carotid and left subclavian.

General complications of catheter technique

1. Due to anesthetic
2. Due to contrast medium

3. Due to the technique

Due to the anesthetic:- 1. General anesthesia
2. Local anesthesia:

The maximum adult dose of lignocaine is 0.2g. Anesthetic lozenges contribute to the total dose.

Symptoms are of paraesthesia and trembling which may progress in to convulsions, cardiovascular and respiratory depression and death.

Treatment is symptomatic.

Due to the contrast medium

ALLERGIC AND IDIOSYNCRATIC

Non fatal reactions are much less common with intra-arterial injections than with intravenous injections.

TOXIC

1. Hot feeling : This is localized to the region supplied by the injected artery. Injection of conventional contrast media may cause severe pain typically when injected in to the arm, the artherosclerotic lower limb or the external carotid artery . symptoms are due to the high osmolarity and it is for this reason that the new low-osmolar contrast media are preferred.
2. A large dose delivered to a specific organ may have a chemotoxic effect.
 - a. Coronary arteries – pure sodium or meglumine salt produce impaired myocardial contractility and ECG changes. Addition of calcium to the contrast medium improves contractility but also increases the subjective feeling of the warmth.
 - b. Cerebral arteries - Sodium salts are more neurotoxic than meglumine salts.
 - c. Spinal cord – direct injection to a lumbar or intercoastal artery which feeds the artery of the Adamkiewicz, or diversion of contrast medium in

to the spinal vascular bed in low aortic obstruction, can result in spinal cord damage.

d. Kidneys – Acute renal failure is a rare complication and is more likely associated with

- Pre-existing renal disease, diabetes or myelomatosis.
- Decreased renal perfusion (hypotension, dehydration, low-output cardiac failure)
- Large volumes of contrast medium
- Recent administration of nephrotoxic drugs.

Due to technique:

LOCAL

1. Haemorrhage/haematoma.
2. Arterial thrombus. May be due to:
 - a. Stripping of thrombus from the catheter wall as it is withdrawn
 - b. Trauma to the vessel wall.

Factors implicated in increased thrombus formation are:

 - a. Large catheters
 - b. Excessive time in the artery
 - c. Many catheter changes
 - d. Inexperience of the radiologist
 - e. Polyurethane catheters, because of their rough surface.

The incidence is decreased by the use of:

 - a. Heparin-bonded catheters
 - b. Heparin – bonded guide wires
 - c. Flushing with heparinised saline
3. Infection at the puncture site
4. Damage to local structures, especially the brachial plexus during axillary artery puncture.

5. False aneurysm. Rare. Present as a pulsatile mass at the puncture site, usually 1-2 weeks after angiography and is due to the communication between the lumen of the artery and a cavity with in an organized haematoma. Treatment is surgical.
6. Arteriovenous fistula. Rare.

DISTANT

1. Peripheral embolus from the stripped catheter thrombus. Emboli to small digital arteries will resolve spontaneously; emboli to large arteries may need surgical embolectomy.
2. Atheroembolism. More likely in old people. J- shaped guide wires are less likely to dislodge atheromatous plaques.
3. Air embolus. May be fatal in a coronary or cerebral artery. It is prevented by:
 - a. Ensuring that all taps and connectors are tight.
 - b. Always sucking back when a new syringe is connected.
 - c. Ensuring that all bubbles are excluded from the syringe before injecting.
 - d. Keeping the syringe vertical, plunger up, when injecting.
4. Cotton fibre embolus. Occurs when syringes are filled with from a bowl containing swabs prevented by,
 - a. Separate bowls of saline for flushing and wet swabs, or
 - b. A closed system of perfusion.
5. Artery dissection, due to entry of catheter guide wire or contrast medium in to the subintimal space. It is recognized by resistance to the movement of the guide wire or catheter or increased resistance of injection of contrast medium. The risk of serious dissection is reduced by:
 - a. Floppy J- shaped guide wires
 - b. Pig tail catheter
 - c. A test injection prior to a pump injection

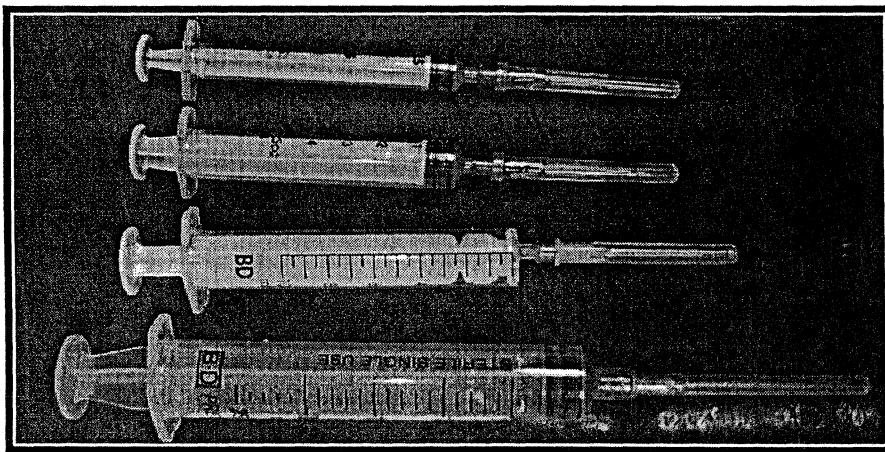
d. Careful and gentle manipulation of the catheters.

6. Catheter knotting
7. Catheter impaction
8. Guide wire breakage

MATERIALS USED FOR ROUTINE ANGIOGRAPHY AND INTERVENTIONS

1. SYRINGES

Syringes used for flushing of the microcatheter, angiography, and injection of the embolic agents must have luerlocks to appropriately secure them to the hub of the microcatheter. The barrel and sturdy enough for repeated use. Most operators use either 1cc or 3cc syringes because adequate injection rates are difficult to obtain with larger volume syringes. The 1cc syringes are made up of clear poly carbonate barrels and plungers that are provided in several colors.



2. Micro puncture Needles

Used to reduce complications in patients when they are at a risk for bleeding, or to prevent accidental puncture of nearby nerves or vital

organs. They are also useful in arteries that are prone to spasm or in patients with grafts or poorly palpable arteries

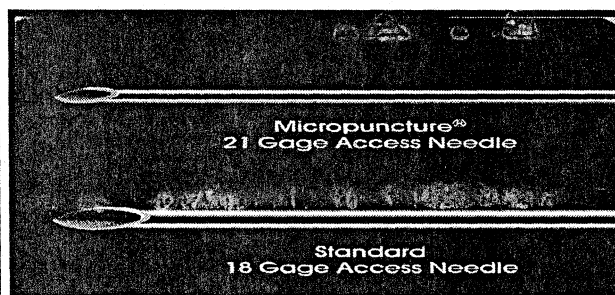
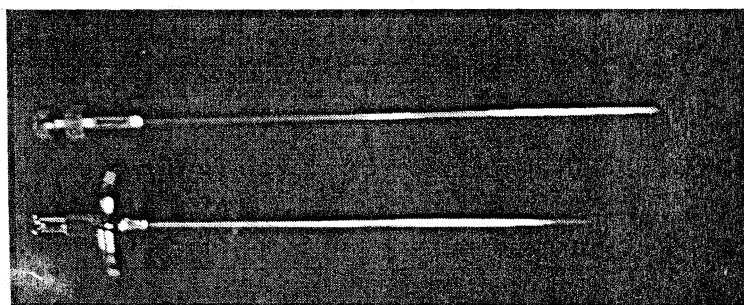
One- Piece or single-walled needles are multipurpose needles that consist of a beveled needle with a base plate; there is no stylet. Two- Piece or double-walled needles consist of an outer cannula with an inner stylet and obturator. Examples include the two piece Seldinger or modified Potts needle.

Three-piece or sheath needles consist of a cannula, stylet and sheath. The stylet is in the cannula, and this system is inside the sheath. Examples include the three-piece Potts Cournand.

NEEDLES

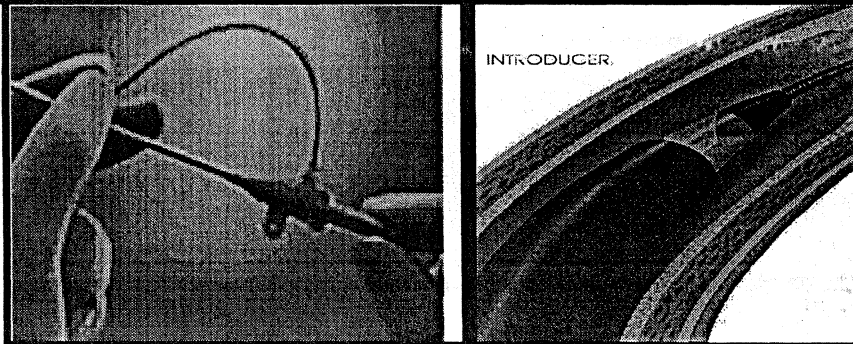
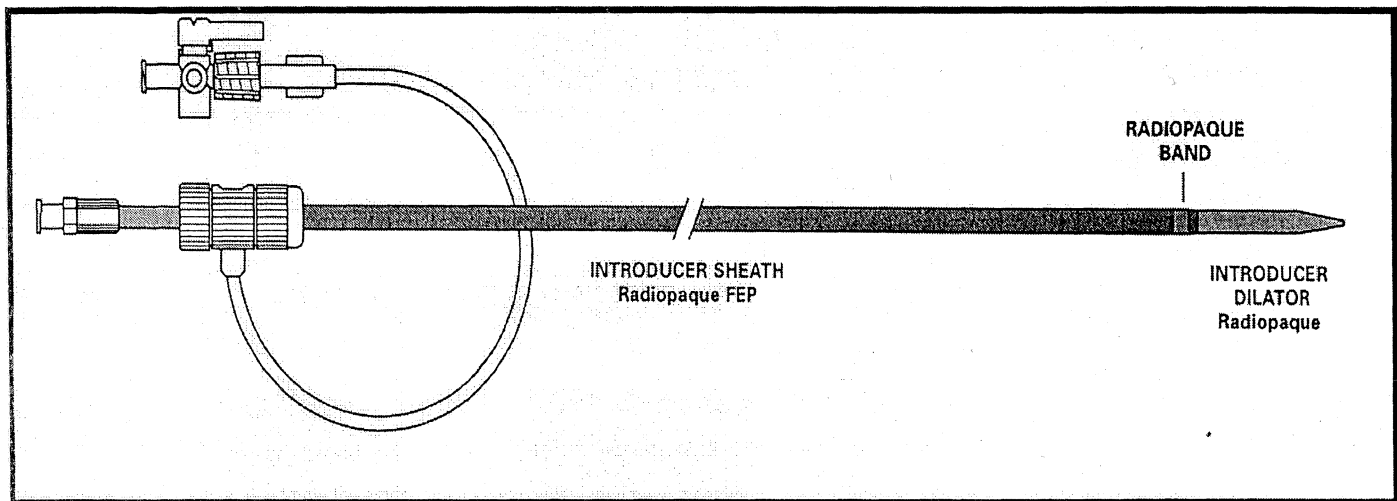
- Single Wall
- Seldinger Type
- Modified Cournand Type

SELDINGER NEEDLE



Arterial needles generally are 2 1/8" or 3 1/4" long and come in a variety of gauge sizes. The 16 to 20 gauge needles are generally used, with the 18 gauges being the standard. All needles used in angiography are thin-walled in comparison to hypodermic needles, which are not. This allows the passage of the guide wire through the needle. A variety of needles are also available for special interventional procedures. Percutaneous arterial needles are available in a variety of designs.

INTRODUCER SHEATH



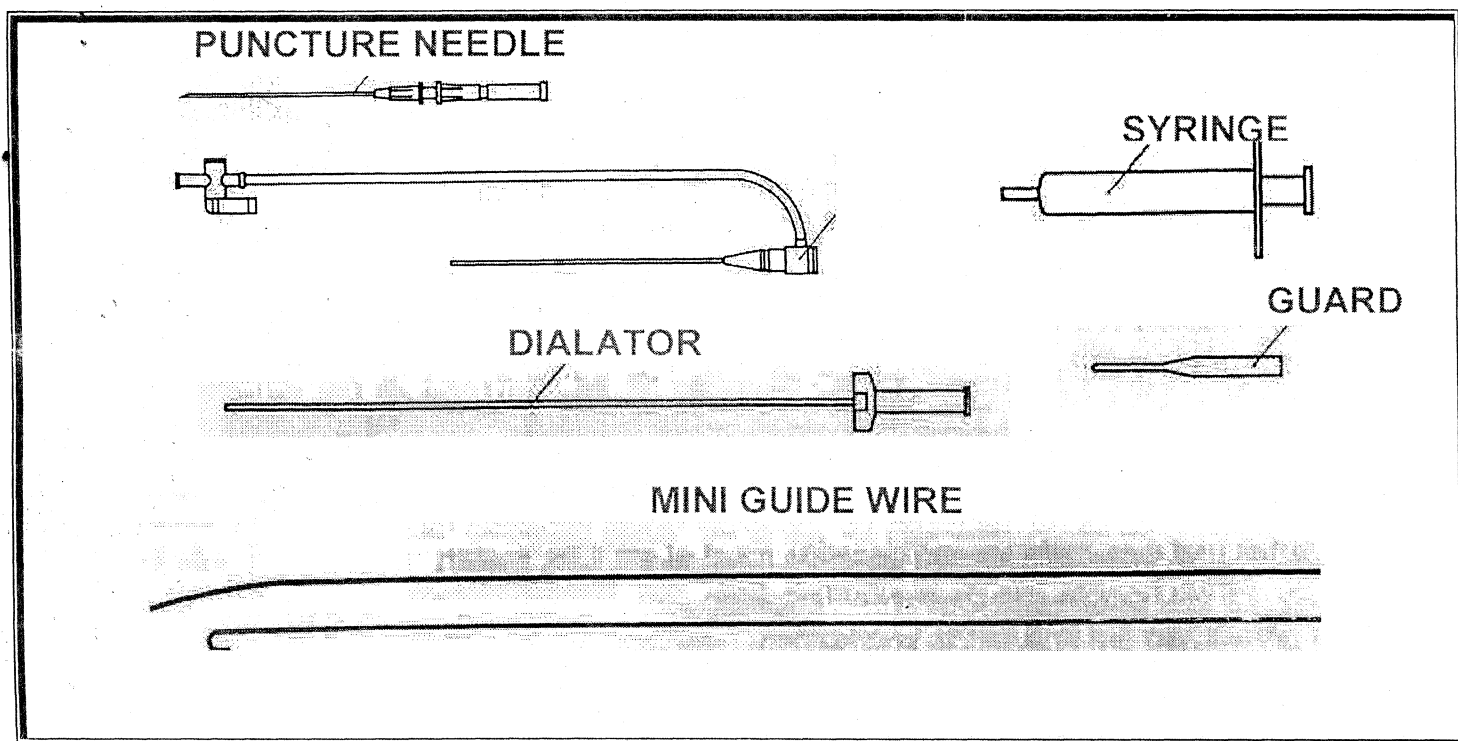
Sheaths provide smoother and safer catheter introduction during procedures that require multiple catheter exchanges or arduous manipulation of the catheter. Sheaths usually come with safety features such as a sideport for heparin and or contrast installation and for obtaining pressure measurements. There should also be a hemostasis valve to prevent air aspiration and blood backflow. Sheath also come with a interlocking dilator to aid in the insertion of the sheath, once the sheath dilator is in the artery, the interlocking hub can be detached and the dilator portion can be removed

For most interventional procedure, after arterial access has been obtained, the placement of sheaths offers significant advantages.

1. Repeated access to the artery
2. Ease of catheter maneuverability.
3. Lessened bleeding.

The placement of a sheath of appropriate length may also allow the operator to negotiate tortuous anatomy only once. There are numerous sheaths available in from various manufactures.

INTRODUCER SET



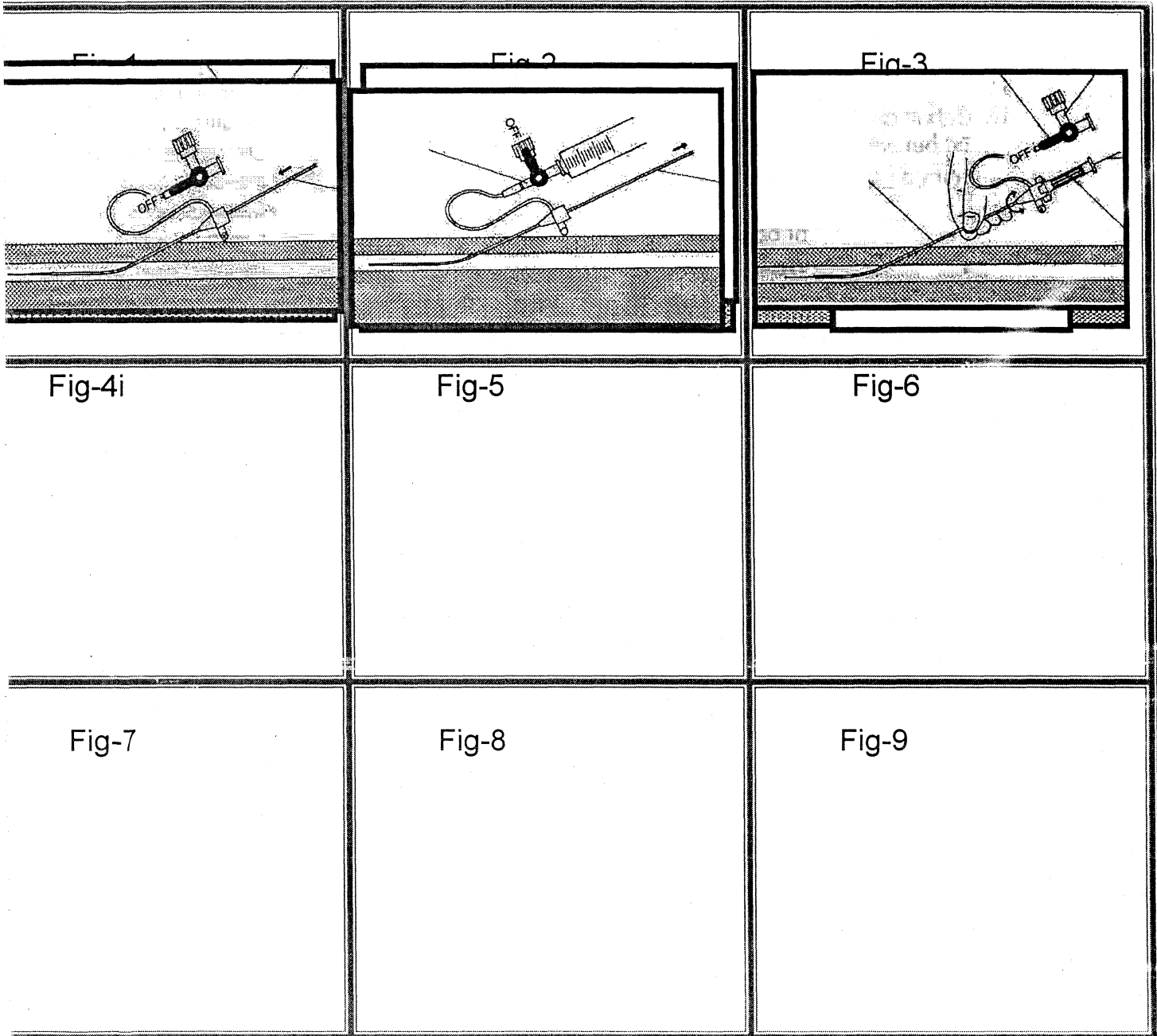


Fig-4i

Fig-5

Fig-6

Fig-7

Fig-8

Fig-9

INSTRUCTION FOR USE

1. Make a small incision at the puncture site with surgical knife.
2. Insert an entry needle to puncture the blood vessel. Then remove the inner metallic needle (fig-1) reinsert the inner needle of the entry needle in to the plastic cannula. Dispose the removed inner metallic needle safely being careful to avoid infections.
3. Before inserting angled/spring mini guide wire, set the guide inserter at the plastic cannula hub for easy insertion. Insert the flexible end of the mini guide wire through the plastic cannula in to the vessel. (Fig-2).
4. Remove the plastic cannula over the mini guide wire.
5. Connect flushing line to the 3-way stopcock of the introducer sheath. Fill the sheath assembly completely with heparanized saline, removing all air.
6. Prime the dilator using the syringe with heparanized saline.
7. Insert the vessel dilator fully in to the sheath. The female hub of the sheaths connects with the male hub of the dilator, and locks in place by means of grip. (Fig-3)
8. Insert the dilator and sheath together over the mini guide wire, and in to the blood vessel (fig-4)
9. Unlock the dilator hub from the sheath hub by bending the dilator hub downward. (Fig-5)
10. Slowly remove the dilator and mini guide wire, leaving the outer sheath in the vessel (fig-6). If injection or sampling is necessary at this point, remove the mini guide wire, and use the dilator hub as an injection port before removing it.
10. Insert the catheter in to the valve center of the sheath.
11. Insert a catheter through the sheath and in to the blood vessel, and advanced to the desired location (fig-7). When

exchanging catheters, remove the used catheter and repeat the same step.

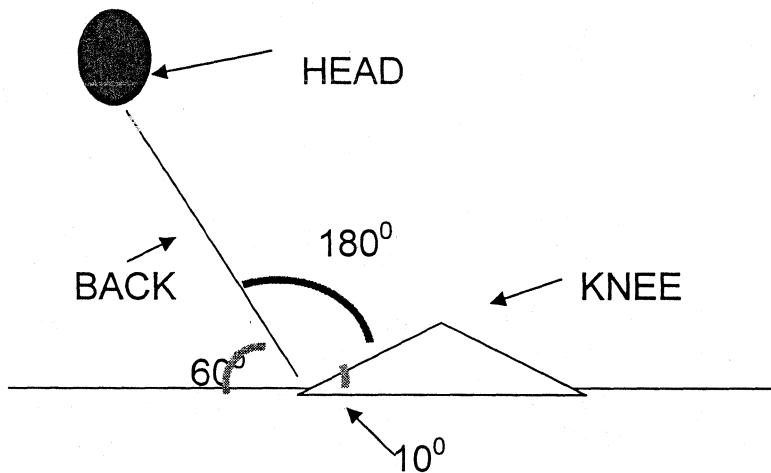
12. When inserting, manipulating or withdrawing a catheter from the sheath always hold it in place .to temporarily suture the sheath

11. After the intended procedure is completed, remove the catheter and then the sheath.



12.

CONTRAINDICATION

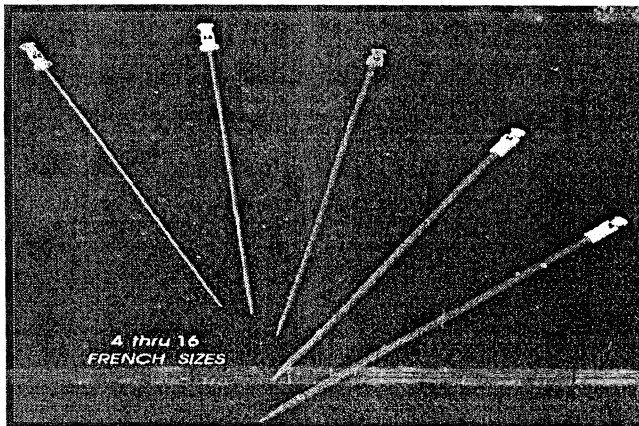
1. Patient must not exceed a 60° angle position post procedure.
2. An introducer obturator must be inserted in to the sheath before a patient can sit up.



VARIOUS SIZES COLOUR

4F (1.35mm)	
5F (1.65mm)	
6F (2.0mm)	
7F (2.3mm)	
8F (2.7mm)	
9F (3.0mm)	
10F (3.3mm)	
11F (3.7mm)	

DILATORS

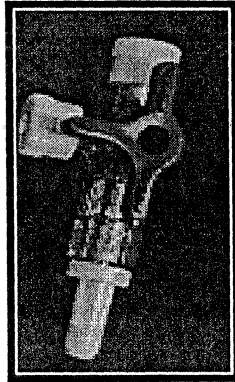


Dilators are thick-walled plastic tubing with a tapered end that provide a tract from the skin surface to the vessel, assuring smooth catheter entry. The dilator is most commonly used when a guide wire is in a vessel but catheter placement is difficult, when a vessel is heavily scarred, when plaque is present at the puncture site, or in the presence of a graft. In the case of dilation of a graft, the interventionalist will find it necessary to overdilate , by going up one French size

ADAPTERS

Provides a means of connection between instruments; classified as male and female; the female adapter can only be connected to a male connector and the male adapter to a female connector.

STOPCOCKS



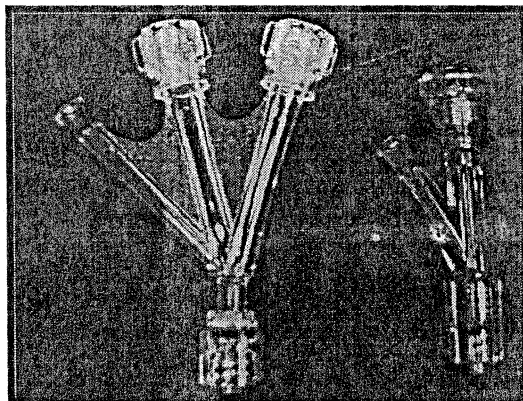
Are valve attachments to control the passage of fluid (blood , contrast, parenteral fluids) that opens and closes with the turn of a lever

MANIFOLD



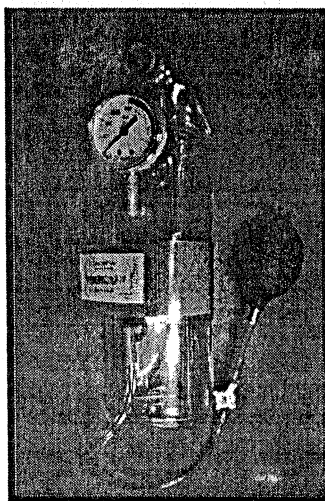
A series of stopcocks placed in either a Y formation or a linear formation. The linear formation allows for the discreet regulation of individual stopcocks.

TUOHY-BORST



It is an adapter that provides a continuous flush in a coaxial system between the two systems.

PRESSURE SALINE BAG



CONTRAST

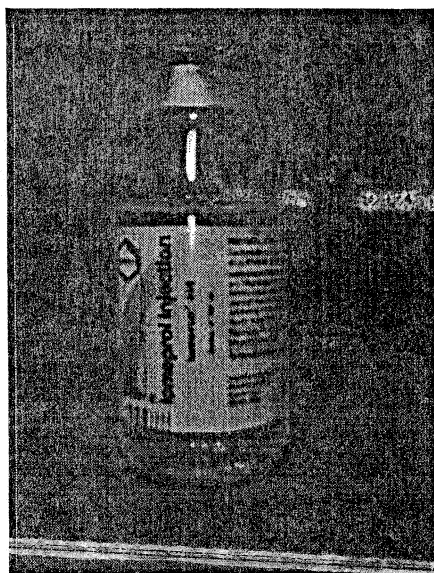
The contrast material used for interventional procedure should be non-ionic for several reasons.

1. Perforation or extravasation can occur at any time.

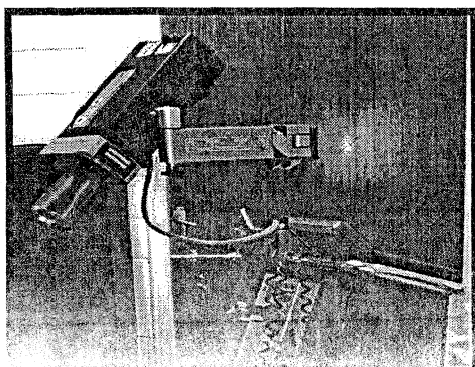
2. non-ionic contrast media are less toxic to the CNS tissue and less likely to provoke seizures.
3. Non-ionic contrast has been shown to cause less renal failure in those patients with underlying renal disease, and it seems reasonable that in other patients there may also be less renal toxicity with non-ionic contrast usage.
4. In the awake patient, the non-ionic contrast cause less movement on injection , allowing for superior images.

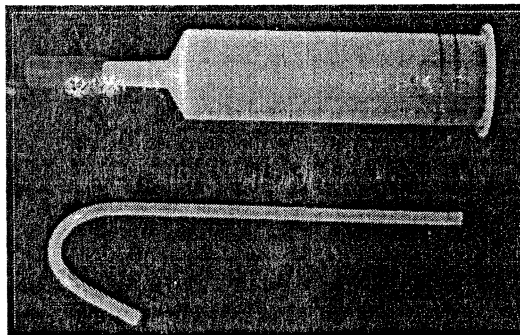
A drawback of non-ionic contrast has been the possible increase in clot formation , which has been somewhat controversial.

Another suggested drawback has been the greater cost of non-ionic.



Automatic Pressure Injectors





The automatic power injector most commonly used today allows for specific volume of contrast to safely be delivered at a precise flow rate, regardless of the variables. Using the electromechanical injector, the parameters can be set at a control panel, directing the amount (ml) of contrast media to be delivered over a set period of time (per second). For example, 15 ml/sec for 4 seconds would be a total volume of 60 ml. The components of the pressure injector include: Syringe Removable; usually sterile disposable type Heating Device Reduces viscosity of the contrast media High Pressure Mechanism Electromechanical system with a motor drive connected to a screw, that drives the plunger piston plate and transmit motion to the syringe Control Panel Allows for programming and display of the injection parameters, i.e., volume flow rate, PSI, injection delay, film delay, linear rise, reset/abort button, visible and audible warning alarms.

Functions of an automatic injector include:

Flow rate - the rate at which the contrast media is going to flow. Variable introduce into flow rate include catheter length, diameter and the number of side holes, as well as contrast viscosity.

Volume - the desired amount of contrast you want to deliver Linear rise- the time it take to reach the desired flow PSI (pounds per square inch)- the pressure at which you inject the contrast media. The pressure should never be higher than the catheter can accept or higher that the vessel can tolerate. The smaller the vessel, the lower the pressure.

Injection delay - the filming will start and the injection, according to the delay set Filming delay- the injection will start first, and then the filming will begin.

Automatic pressure injectors are equipped with multiple safety features to prevent inadvertent damage to the patient or damage to the catheter.

Function monitoring devices : A flashing red light, audible warning or a printed message to notify the operator that a setting has been omitted or that there is some mechanical malfunction

Volume limiting device: The maximum limit on the amount of contrast administered is set on the control panel, but a backstop lever is located on the head of the injector. Pressure limiting device: Sets a maximum pressure (PSI) to be induced, prevents catheter breakage or recoil and vessel damage; ensures safe use with low-pressure catheters.

Accelerator regulators : Allows the drive motor to propel over an exact duration of time, to guard against recoil

Rate rise control: Allows gradual rise in (PSI) instead of an instant surge or pressure all at once

Additionally, the radiographer must take care to properly load the syringe. With the syringe pointing up straight up, clear all air bubbles from the syringe. Automatic injectors have loading and unloading buttons to draw up and empty contrast. When hooking up to the catheter, point the injector head down. In the event any air bubbles are in the syringe, they would rise away from the catheter. A high pressure connector is usually used to connect to the catheter . This allows for better visualization of any air bubbles. The radiologist will direct the radiographer to first advance contrast. Once the connection is made, aspirate back blood and check for any air bubbles in the connecting tube. Care should be taken not to aspirate too much blood, as this will dilute the contrast and could possibly cause clot formation in the injector syringe. Before the injection begins, the radiologist and radiographer should confirm the parameters selected. Once the injection begins, the radiographer must be prepared to terminate the injection, in the event of an emergency .

Injectors in use:

Angiomat 6000

Vol. Range : 0.1ml – 9.9 ml in 0.1ml increments .
: 1ml – 150ml in 1ml increment.

Flow rate range : 0.01ml – 40ml/sec in 0.01ml increments.

Duration range : 0 – 10/255sec in 0.01sec increment.

Pressure limit : 127 to 1200 psi in 1psi increment.

Medrad Envision CT injector
EOM 700 E

Fluid delivery performance:

Vol. Range : 1ml to max. syringe capacity (125ml)
Flow rate range : 0.1 – 9.9ml /sec in 0.1ml/sec increments
Duration range : 1sec – 20:50Min for 125ml syringe.
Pressure limit : 25 – 300 psi in 1psi increments
- 1.7 – 20.7 Bar in 0.1 bar increments

CATHETERS

Catheters are long, hollow tubes that supply an avenue or passageway for contrast media, embolizing materials or therapeutic medications or instruments. Catheters come in a variety of sizes that are measured in the term called French as general rule: 3F to 5F for children and 4F to 9F for adults. This term French refers to the outer diameter of the catheter.

Catheters also have a variety of shapes which include single and multiple curves. Catheters are designed to be either radiolucent to view bubbles or radiopaque for better visualization. Memory refers to a catheters ability to retain its shape. Torque refers to the catheters ability to twist and turn. Another term use to discuss catheters is Tractability this refers to how well a catheter can track over a guide wire Ordinary catheters cannot normally be directed into a CC-fistula or into the most distal external and internal carotid artery branches that may be supplying an AVM. So micro catheters are used they range in the order of 2.5fr. to 3.5fr. These are made of highly flexible silicone. These catheters may be either flow directed or may require the use of a guide wire.

Units Useful for Angiographic Catheters & Guide wires

1" = 25.4mm

1mm = 3F = 0.039"

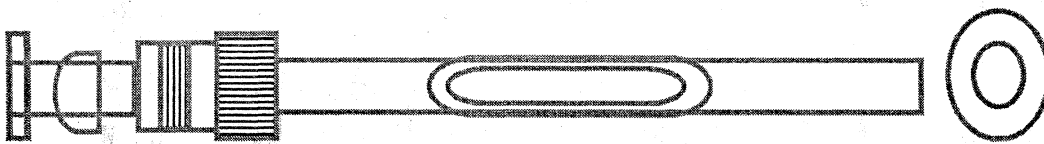
1F = 0.013"

French FR	Sizes mm
1F	1/3mm
2F	2/3mm
3F	3/3mm

Example 9F catheter = 3mm Diameter

Angiographic Catheter

Non-Braided Catheters



Catheter Construction
section

Cross

Types:

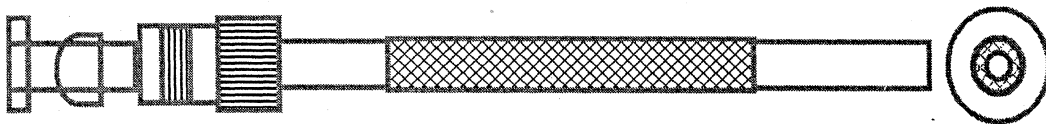
Radio opaque Polyethylene – has a medium coefficient of friction and less stiffness than TFE. It does absorb some moisture. Polyethylene tubing contains Bismuth Salts for radio opacity. Suitable for selective and super selective angiographic studies.

Thin wall radio opaque Nylon – has low friction coefficient and is kink resistant. Bismuth salts for radio opacity. Suitable for maximum high – flow administration of contrast medium.

Radio opaque TFE – has a lower coefficient of friction and higher burst point than polyethylene of similar wall thickness and out side diameter. TFE is stiffer than polyethylene. It doesn't absorb moisture. Bismuth salts are used for radio opacity. Suitable for non-selective angiographic studies. (! Never sterilise TFE by irradiation; complete disruption results.

Radio opaque vinyl – is more flexible than standard polyethylene material. Suitable for selective and sub selective angiographic studies when standard catheter can't be advanced over wire guide to desired position.

Braided Catheters



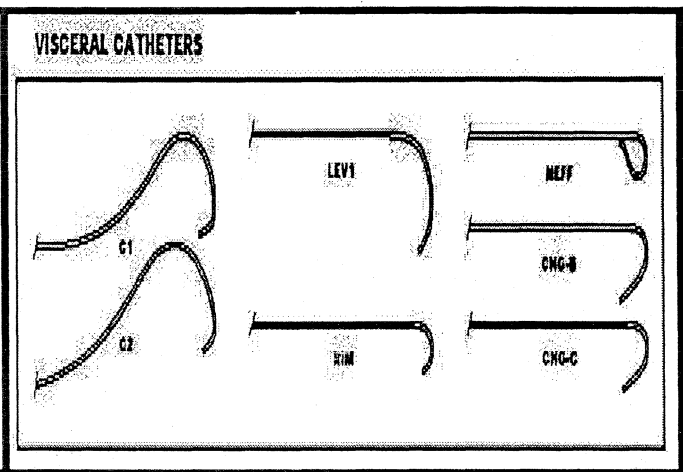
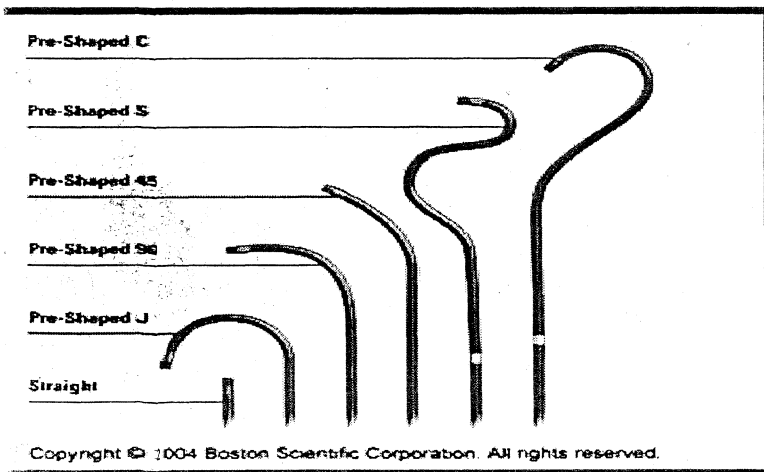
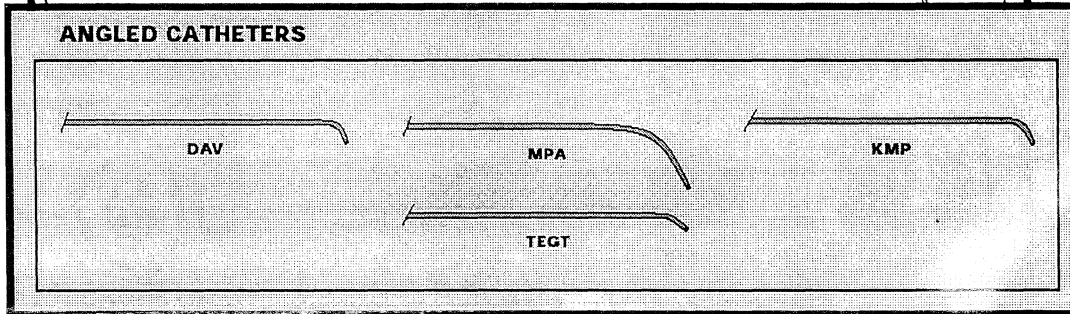
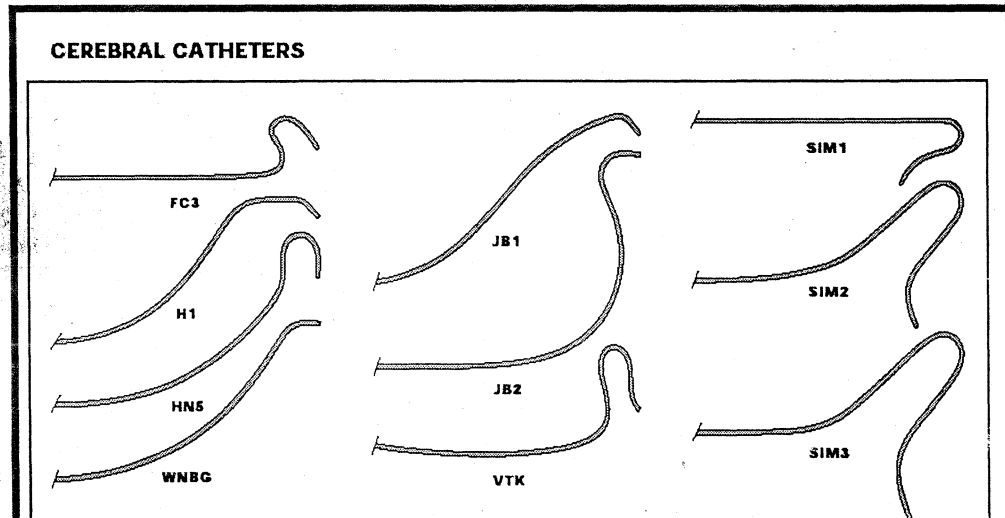
Cross section Catheter construction

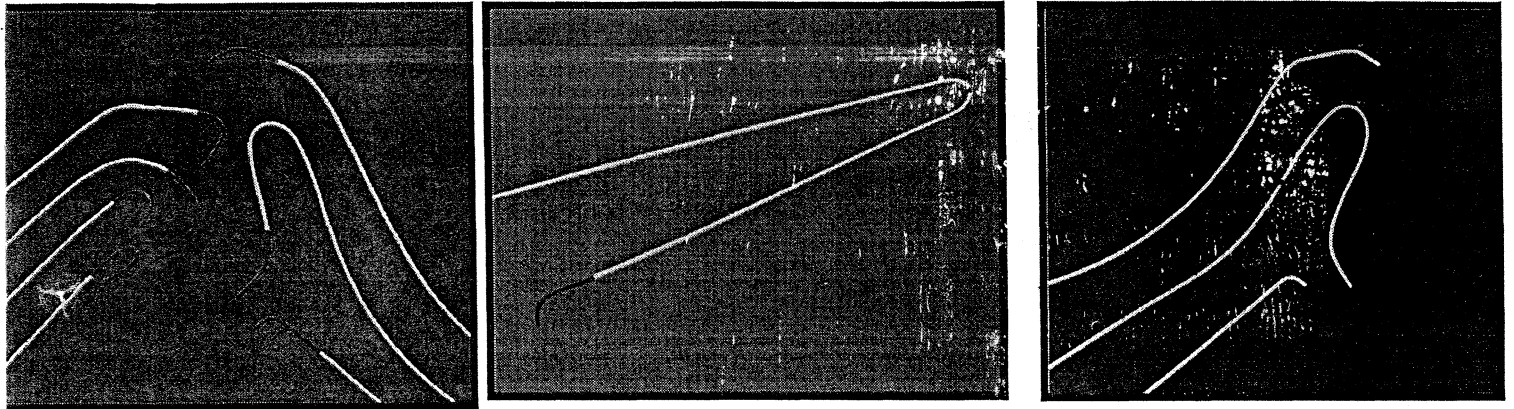
Types:

Thin wall radio opaque polyethylene with torque control – has stainless steel braided construction within catheter shaft which imparts torque control to catheter tip. Bismuth salts are used for radio

opacity. Suitable for high flow administration of contrast medium in selective angiographic studies.

Thin wall Radio opaque Nylon with Torque control – has low coefficient of friction surface characteristics; incorporates stain steel braided construction with catheter shaft which imparts torque control to catheter tip for a One to One response. Bismuth salts are used for radio opacity. Suitable for high flow administration of contrast medium in selective angiographic studies.

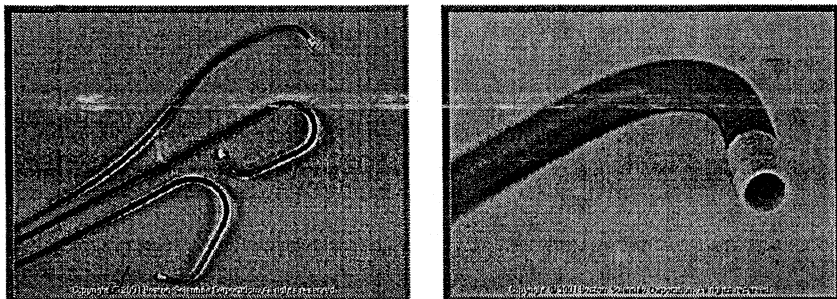




GUIDE CATHETERS

Used in interventional procedures for introducing microcatheters. In selecting the size of the micro catheter, the operator must always remember the goals of the procedure. This requires matching of the guide catheter inner lumen with the outer diameter of the micro catheter. In addition one must also keep in mind the inner lumen of the micro catheter, so that appropriate embolics can be delivered. In selecting the guiding catheter, the stiffness and shape of the tip are important. The guide catheter frequently will move with respiration and with placement of the micro catheter. This can cause the tip of the guide catheter to dig in to the intima, which seems to be occurring more frequently with the pre shaped models. The stiffness of the guide catheter shaft also is important, because the micro catheters can exert force against the guide catheter and push it from the desired vessel. However, too stiff a catheter may make initial placement difficult and may also lead to undesired straightening of vessels and sometimes-intimal damage. The length of the guide catheter is important in very tall individuals, in whom a 90-cm length may be inadequate for appropriate positioning of the guide catheter.

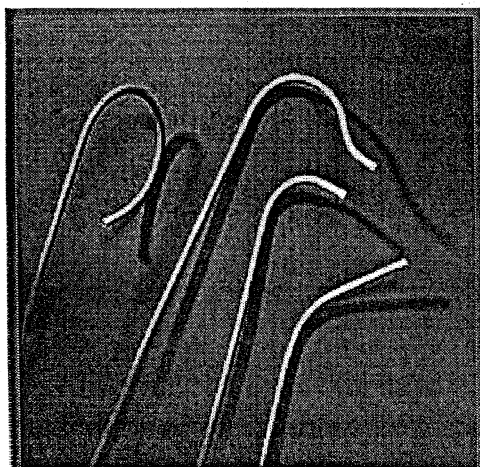
Mach 1™ Guide Catheter



COAXIAL DESIGN LUMAX® FLEX GUIDING CATHETERS

Used as a support catheter for angioplasty catheters and other devices for intervention. The inner coaxial catheter is tapered to a .038-inch end hole to facilitate percutaneous placement of the guiding catheter, without use of an introducer sheath, into the vessel. Device movement through the catheter into the vessel is enhanced by a large ID and low friction lumen surface. Catheter construction provides excellent support without softening during procedural use. LuMax® Flex tip allows atraumatic engagement of target vessel while providing high radiopacity for precise positioning. Supplied sterile in peel-open packages. Intended for one-time use

VISTA BRITEV TIP



MICROCATHETERS: - All of the commercially available micro catheters are constructed of polyethylene and are hydrophilically coated. Many micro catheters will contain braided materials, which improves flexibility, pushability, and trackability of the microcatheter. The braided construction lessens the incidence of micro catheter kinking or ovalizing as it traverses bends. This braid feature can also

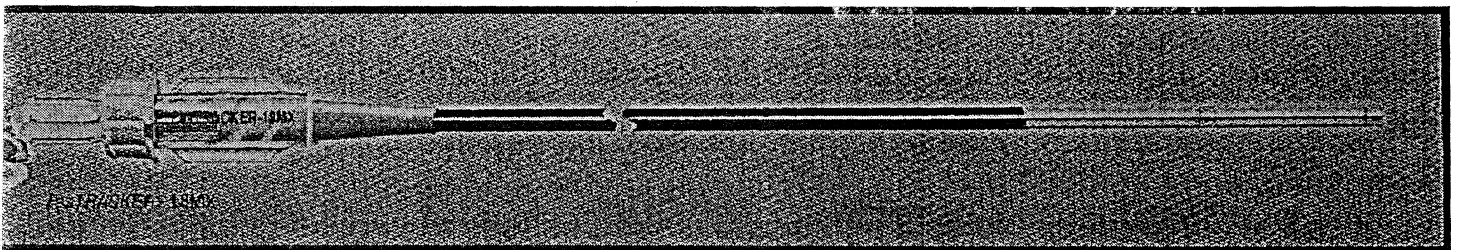
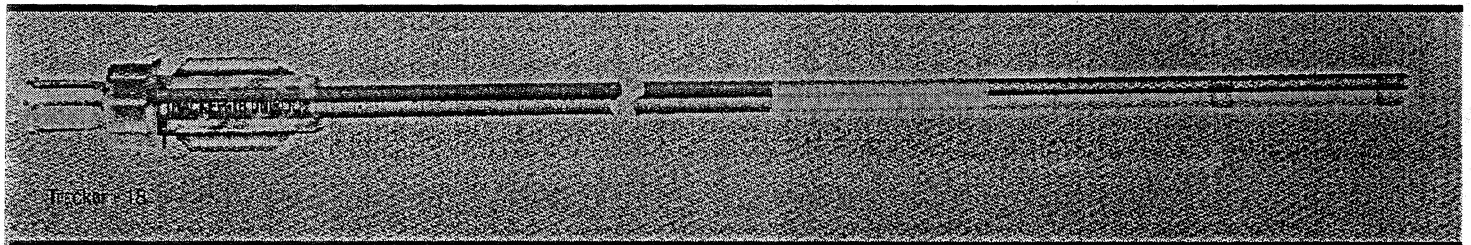
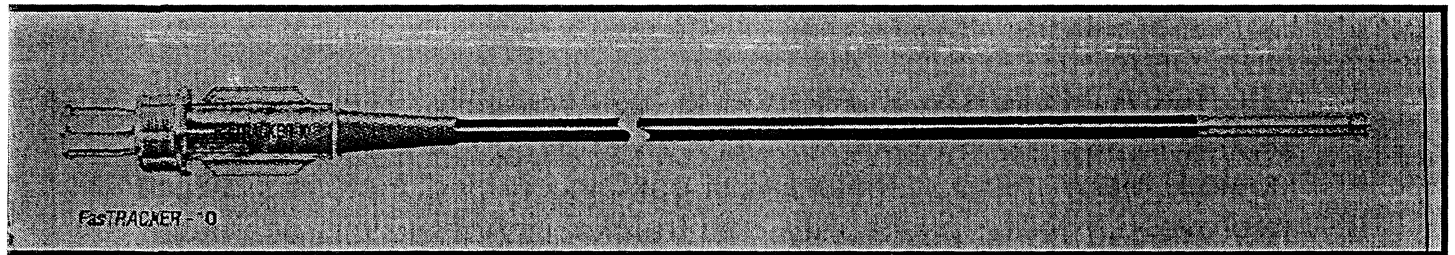
MANUFACTURES	OUTER DIAMMETER			INNER DIAMMETER		
	F	mm	Inch	F	mm	Inch
BALT	3F	0.95	0.040	1.6	0.60	0.023
	4.5	1.50	0.059	3.00	1.00	0.039
	5F	1.67	0.066	3.60	1.20	0.047
	6F	2.00	0.079	4.5	1.50	0.059
	7F	2.33	0.093	4.8	1.60	0.063
	8F	2.67	0.104	6.00	2.00	0.079
	9F	3.00	0.118	7.2	2.40	0.094
	10F	3.33	0.133	7.8	2.60	0.102
NYCOMED	5F	1.67	0.066	3.68	1.23	0.04797
	6F	2.00	0.079	4.44	1.48	0.05772
	7F	2.35	0.092	5.28	1.76	0.06864
	8F	2.68	0.105	6.00	2.00	0.079
	9F	3.00	0.118	6.99	2.33	0.093
	10F	3.35	0.131	7.80	2.60	0.102
	6F	2.00	0.079	4.92	1.64	0.064
CORDIS VISTA BRITE TIP	6F	2.00	0.079	4.92	1.64	0.064
	7F	2.35	0.092	6.00	2.00	0.078
	8F	2.68	0.105	6.768	2.256	0.088
	9F	3.00	0.118	7.53	2.51	0.098
	10F	3.35	0.131	8.46	2.82	0.110

cause the microcatheter to move forward and suddenly to retract as the guide wire is removed. Most

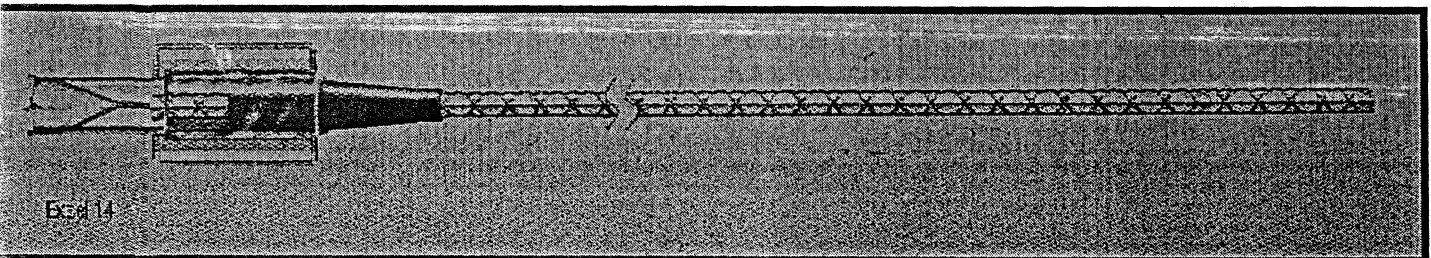
Currently available micro catheters have similar performance characteristics. All the catheters have a marker at the tip, and most are available in a two-marker variation for the deployment of coils.

NON BRAIDED

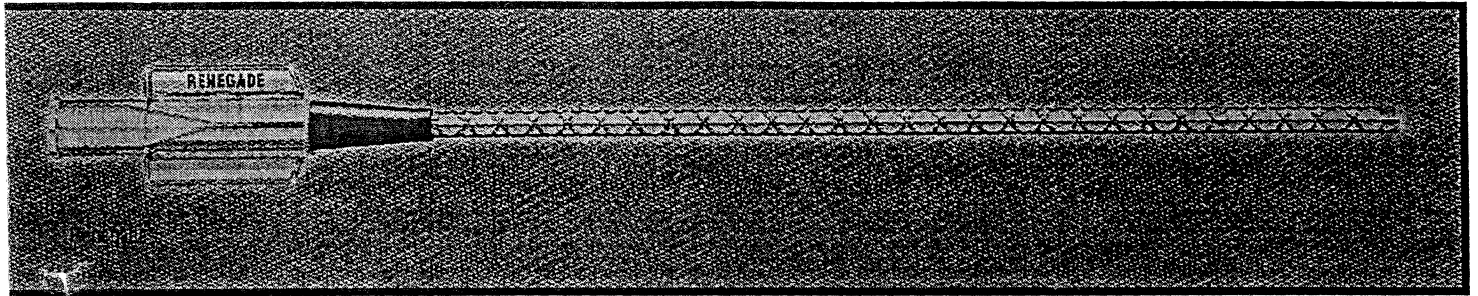
FasTRACKER-10



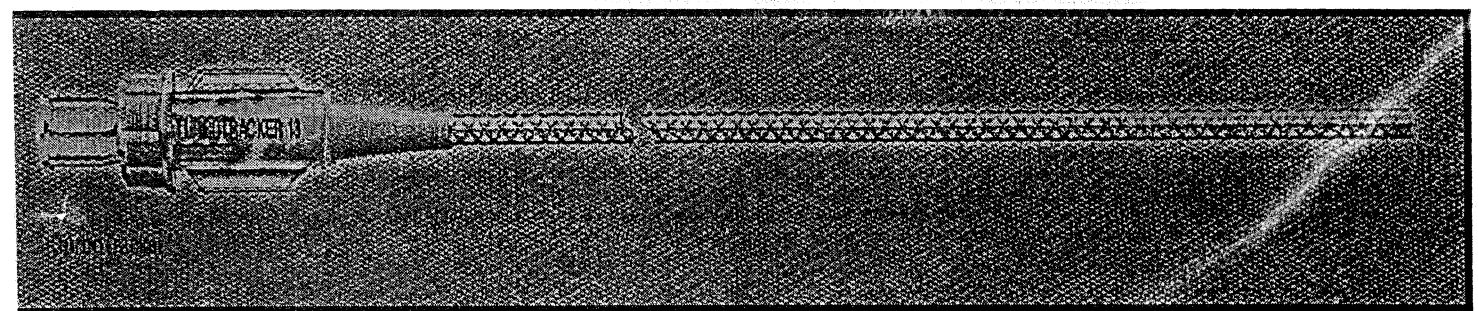
Excel-14



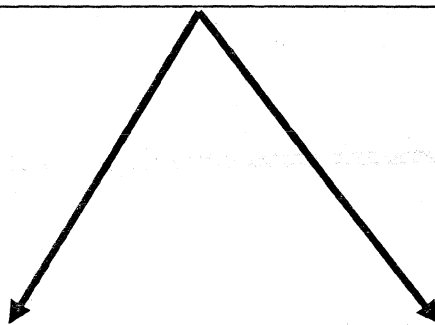
RENEGADE



TURBOTRACKER



MICROCATHETERS



OVER THE WIRE
MICRO CATHETER

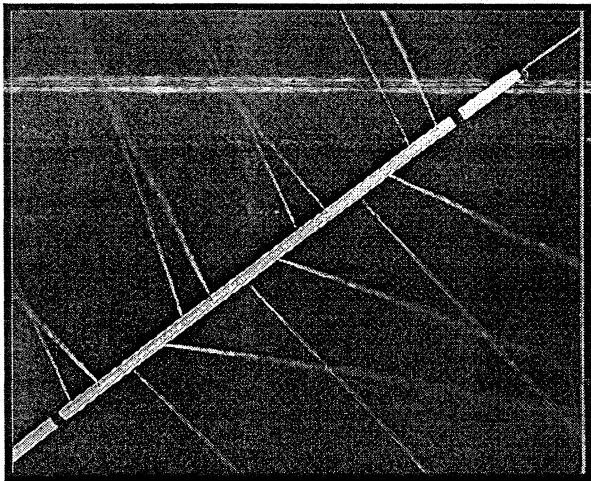
FLOW GUIDED CATHETER

OVER THE WIRE MICROCATHETERS

Used for the infusion of thrombolytic agent.

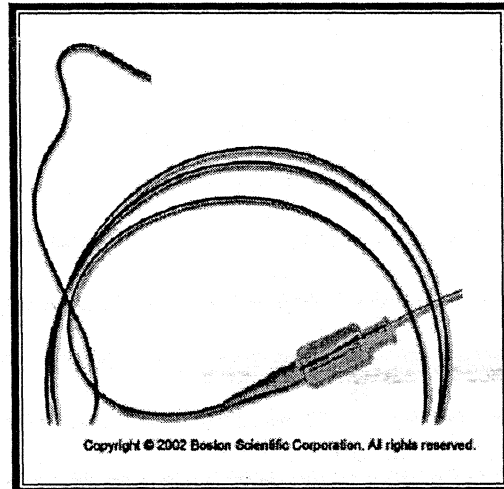
MICROSOFT STREAM(BOSTON),

Which requires a 0.042-inch lumen guide catheter. This is a 150 cm hydrophilic-coated 0.021-inch inner diameter microcatheter that has three side holes in the distal 2cm. These side holes are smaller proximally and larger distally to promote equal distribution of flow.

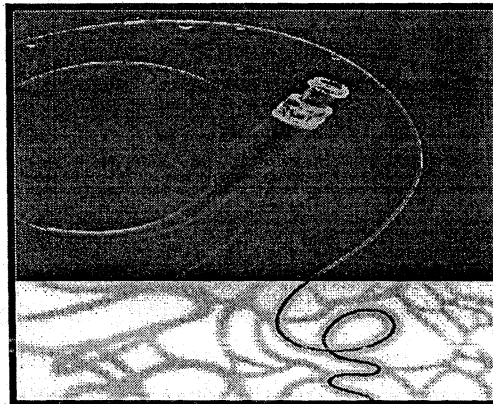


In-Time Infusion catheter (Boston)

A 160 cm hydrophilic coated microcatheter to which 17mm of the distal portion of a transend micro guide wire have been added. This wire tip is shapeable and eliminates the need for a micro guide wire. The distal 1cm of the microcatheter, which is just proximal to the radiopaque micro guide wire tip, has six side holes for the infusion. The In-Time microcatheter may be delivered through a 0.035-inch lumen catheter.



MICROFERRET -18



Used for diagnostic or interventional procedures in small vessels or super selective anatomy including neuro, peripheral or coronary applications. The microferret-18 catheters tapers from a proximal 3.0 French to a distal 2.4 French diameter. The catheter features a three-step multiple stiffness polyethylene catheter shaft to facilitate placement in tortuous or super selective vascular anatomy. The distal soft, flexible portion resists kinking and is available in 12 or 20cm length. The rigid proximal and intermediate portions facilitate pushability. Slip-coat hydrophilic coating on the distal two segments of the catheter provides an extremely lubricious surface for enhanced maneuverability. The catheters accept a wide variety of embolisation devices including .018 inch microcoils and appropriately sized PVA

embolisation foam. The catheter is available with one or two platinum radiopaque band.

COOK → MICROFERRET-18

Working length	150
Proximal OD French	3.0
Proximal ID(IN.)	0.018
Braid material	NONE
Distal OD French	2.4
Distal ID (IN.)	0.018
Tip marker OD French	2.4
Markers	1or2
Minimal guide (in.)	0.038
Maximal wire OD (IN.)	0.016
Inner lumen construction	Polyethylene

Prowler microcatheter(braided)

The prowler micro catheters are also available in a preshaped 45-degree, 90-degree angle, J-tip. The preshaped curves keep the operator's fingers from the steam, and the microcatheter seems to maintain their shape longer .At times, as mentioned earlier, the braided catheter will retract as the guide wire is removed. Similarly braided catheters have a tendency to suddenly move forward .

CORDIS

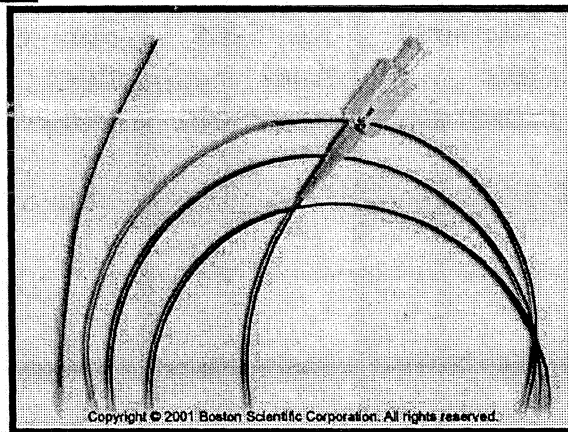
	PROWLER	PROWLER	PROWLER
RAPID			
	10	14	PLUS
TRANSIT			

Working length	150/50	150/20	150/20	150/50
Distal flexible segment	170/50	170/50	150/45	170/50
Proximal OD French	2.3	2.3	3.0	3.0
Proximal ID INCH	0.015	0.0165	0.021	0.021
BRAID MATERIAL	Platinum coil distally Stainless steel proximally	Platinum coil distally Stainless steel proximally	Platinum coil distally Stainless steel proximally	Platinum coil distally Stainless steel proximally
DISTAL OD FRENCH	1.7	1.9	2.3	2.3
DISTAL ID INCH	0.015	0.0165	0.021	0.021
TIP MARKER OD FRENCH	1.7	1.9	2.3	2.3
MARKERS	1or2	1or2	1or2	1or2
MINIMAL GUIDE INCH	0.035	0.035	0.042	0.042
MAX WIRE OD	0.012	0.014	0.018	0.018
INNER LUMEN CONSTRUCTION	PTFE	PTFE	PTFE	PTFE
DEAD SPACE	0.32	0.35	0.50	0.50

FLOW GUIDED MICROCATHETERS

These are very flexible hydrophilic-coated catheters that are primarily designed to deliver liquid embolics such as glue, onyx, and dehydrated alcohol, PVA (less than 500 μ m) can be administered through these microcatheters as well.

SPINNAKER ELITE



Developed specially for flow directed applications, the spinnaker elite flow directed microcatheter might be used for regional infusion of diagnostic agents and vascular occlusion with berenstein liquid coil-10. The flow-directed spinnaker elite (Boston) is not approved for use with glue or other liquid agents, which would seem to be its purpose.

FEATURES

ADVANCED COMPOSITE SHAFT CONSTRUCTION

Segment- specific, composite shaft construction is designed to deliver the most desirable performance attributes in each shaft segment.

Extra flexible distal segment

1. Atraumatic, unbraided distal segment promotes flow directed navigation and vessel selectively.
2. Steam shapeable tip.
3. Offered n 1.5F and 1.8F OD and a choice of distal segment lengths.

Responsive proximal shaft

1. High durometer
2. unbraided shaft, body for support, control, and pushability.
3. smooth mid shaft transitions facilitate advancement through tortuous vasculature.

Hydrolene- hydrophilic polymer surface

Ultra-lubricious outer surface facilitates advancement and manipulation of the spinnaker elite with in the guide catheter and through challenging anatomy.

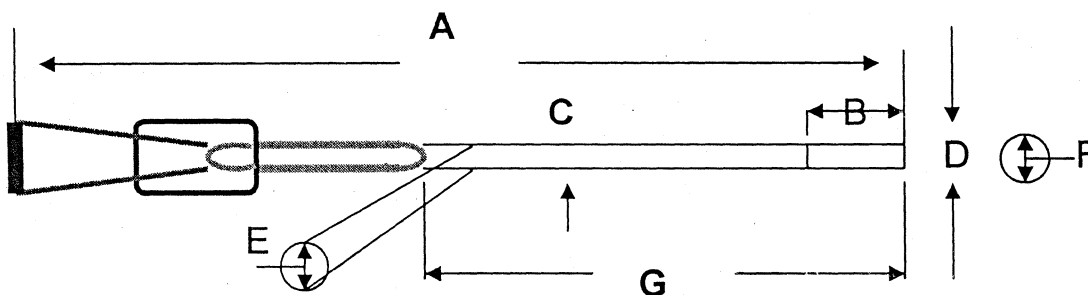
Radiopaque distal tip marker

Platinum distal tip marker provides superb fluoroscopic visualization.

Optional stylet

An optional stylet may be used to facilitate introduction of the spinnaker elite in to the guiding catheter.

Spinnaker Elite

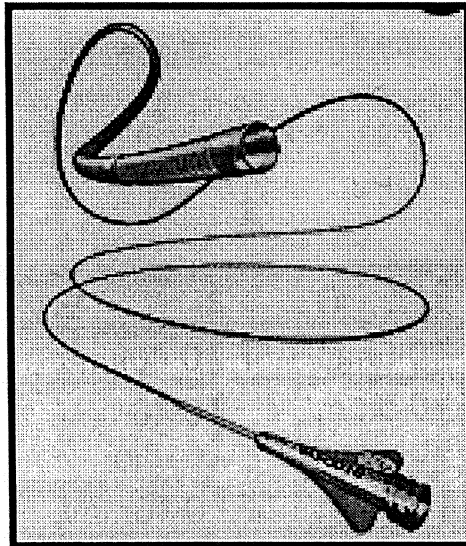


1.5F	A	B	C	D	E	F	G
S	166cm	15cm	1mm/3F	0.5mm/1.5F	0.46mm/.018in.	0.28mm/.011in.	160cm
L	166cm	20cm	1mm/3F	0.5mm/.011in.	0.46mm/.018in.	0.28mm/.011in.	160cm
XL	166cm	30cm	1mm/3F	0.5mm/1.5F	0.46mm/.	0.28mm. 011in.	160cm

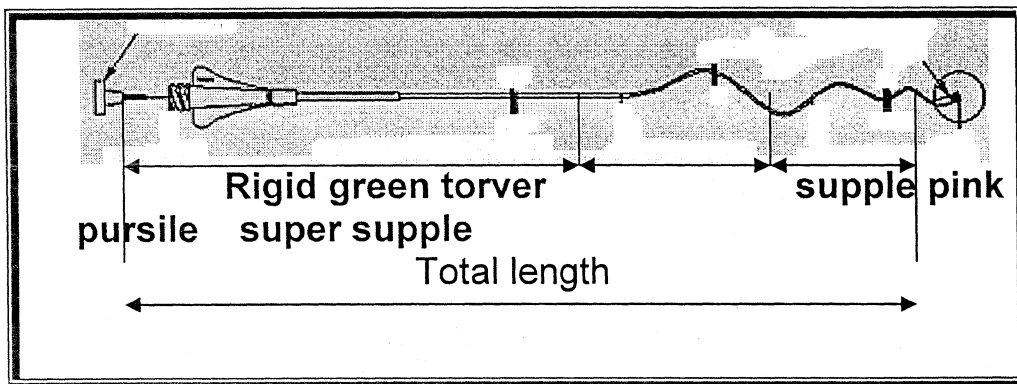
018in.

1.8F	A	B	C	D	E	F	G
S	166cm	10cm	1mm/3F	0.6mm/1.8F	0.46mm/ 018in.	0.28mm/ 011in.	160cm
L	166cm	20cm	1mm/3F	0.6mm/1.8F	0.46mm/ 018in.	0.28mm/ 011in.	160cm
XL	166cm	30cm	1mm/3F	0.6mm/1.8F	0.46mm/ 018in.	0.28mm. 011in.	160cm

BALT MAGIC



MAGIC catheters are designed for general intravascular use. They may be used for the controlled, selective regional infusion of therapeutic agents or embolic materials into vessels. The MAGIC catheter is intended to facilitate access through distant, tortuous vasculature. Progressive suppleness ranging from a super supple distal shaft to a rigid proximal shaft allows the catheter to be advanced by the physician. The rigid proximal shaft allows torque control to facilitate the advancement of the catheter. The MAGIC catheter tip (ring) and shaft are radiopaque.



LENGTH (CM)

Reference	Total	Green TORVER	Pink PURSIL	White PURSIL	REMARK
MAGICSTD	155	120	25	10	TIP 1.8F
MAGICMP	165	120	25	20	
MAGICOLIVE	155	120	25	10	Olive TIP
MAOLIVEMP	165	120	25	20	1.8F
MAGIC 1.5F	155	115	25	15	TIP 1.5F
MA 1.5F MP	165	120	25	20	
MA 1.5F OLIVEMP	165	120	25	20	OLIVE TIP 1.5F

LENGTHS (CM)

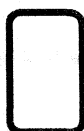
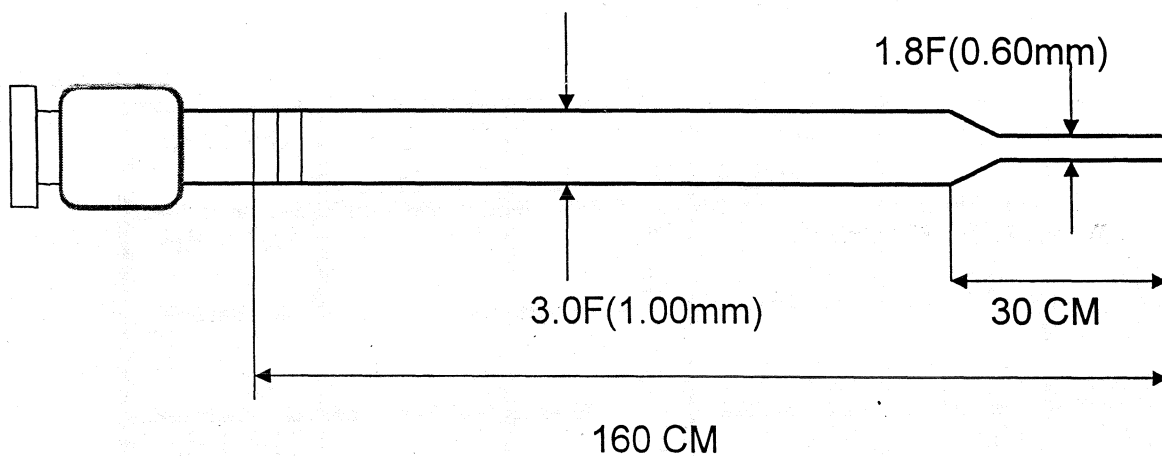
Reference	Total	Green TORVER	Pink PURSIL	White PURSIL	Turquoise PURSIL	REMARKS
MAGIC 1.2F	165	120	30	3	12	TIP 1.2F
MAGIC 1.2FM	165	120	30	12	3	TIP 1.2F
MA1.2F OLIVE	165	120	30	3	12	OLIVE TIP 1.2F

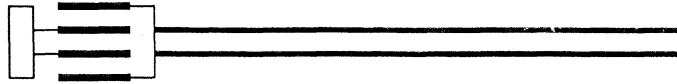
PARTICLE INJECTION MAGIC CATHETERS

Lengths (cm)

References	Total	Green TORVER	Pink pursil	White pursil	REMARKS
MAGIC3F	155	120	-	35	Particules up to 1000µ

REGATTA (CORDIS)





WORKING LENGTH	160/10
DISTAL FLEXIBLE SEGMENT	160/20
PROXIMAL OD FRENCH	3.0
PROXIMAL ID (INCH)	0.021
DISTAL OD FRENCH	1.8
DISTAL ID (INCH)	0.014
TIP MARKER OD FRENCH	1.8
MINIMAL GUIDE (INCH)	0.042
MAXIMUM WIRE OD (INCH)	0.010
INNER LUMEN CONSTRUCTION	N/a
DEAD SPACE	0.27
BURST PRESSURE	150

MICRO THERAPEUTICS

Rebar-10

Rebar-14

Rebar-18

Working length	153	153	153
Distal flexible segment	170	170	170
Proximal O.D French	2.3	2.4	2.8
Proximal ID(INCH)	0.015	0.017	0.021
Braid material	Stainless steel	Stainless steel	Stainless steel
Distal OD French	1.7	1.9	2.3
Distal ID(INCH)	0.015	0.017	0.021
Tip marker OD French	1.7	1.9	2.3
Markers	2	2	2
Minimal guide(inch)	0.035	0.035	0.035

Maximum wire OD(in..)	0.012	0.014	0.018
Inner lumen construction	PTFE	PTFE	PTFE
Dead space	?	0.27	0.27

TARGET

Working length	150/6	150/15	150/10	150/3
Distal flexible segment(cm)		150/7.5	150/20	155/15
Proximal OD French	2.6	2.4	3.0	2.6
Proximal ID(IN.)	0.019	0.017	0.021	0.016
Braid material	Stainless steel	Stainless steel	Fiber Weave	None
Distal OD French	2.0	1.9	2.5	2.0
Distal ID(IN.)	0.019	0.017	0.021	0.015
Tip marker OD French	2.0	1.95	2.5	2.6
Markers	1or2	1or2	1or2	1or2
Minimal guide(in.)	0.042	0.035	0.038	0.038
Maximal wire OD(IN.)	0.016	0.014	0.018	0.010
Inner lumen construction	PTFE	PTFE	PTFE	PP/EVA proximal PE/EVA distal
Dead space(cc)	0.35	0.30	0.47	0.30

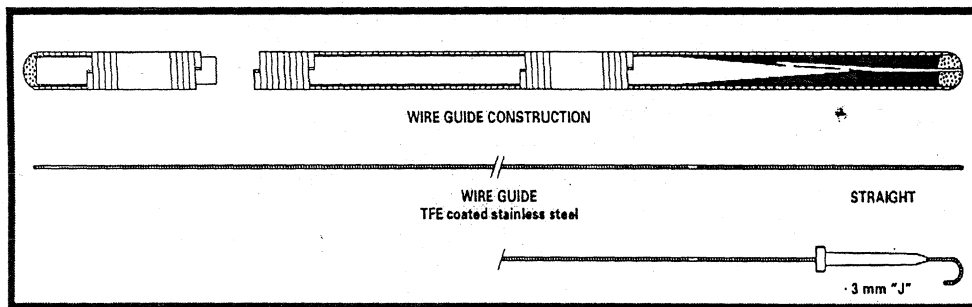
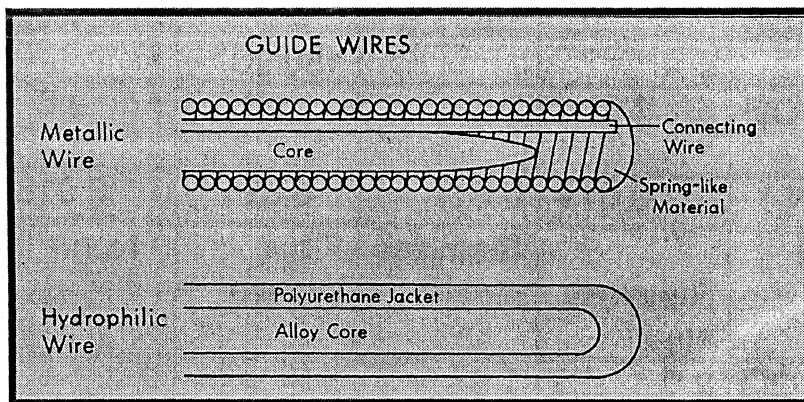
Excelsior

Excel

Renegade

FasTracker-10

Guide Wires:



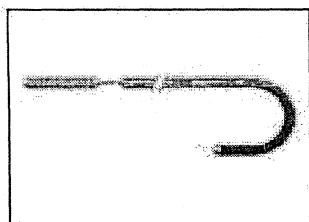
- Guide wires are thin and are constructed of stainless steel wire.
- Guide wires are very flexible and vary in size and length.

GUIDEWIRE USED FOR DIAGNOSTIC ANGIOGRAPHY

The length of guide wires vary from 100 to 300 cm and sizes range from 0.025 " to 0.045 " in diameter. Guide wires may be coated with Teflon, polyurethane, polyethylene or a hydrophilic coating (Glide Wires) to lower the friction coefficient and prevent damage to the vessel walls. Guide wires may also treated with heparin to reduce the chance of blood clotting and emboli formation.

CONSTRUCTION.

The usual guide wire has mainly two parts one is the coil and the other is the core or mandrel. Spinning a thin strand of round wire around a metal tube or rod usually makes the coil. The coil is advanced over a core or mandrel and fixed at the ends by a soldered bond. At the flexible end or tip a safety ribbon is soldered and runs the length of guide wire adjacent to the core.



All guide wires have an inner safety wire, which is bonded or welded at both ends to prevent separation in the event of guide wire breakage and to contain the coils of the spring wire during manipulation. The wire tips are welded and polished for strength and smoothness to minimize vessel trauma. We usually refer guide wire

length in centimeters and the diameters both in inches and in millimeters.

Material.

Guide wires are all constructed of fine stainless steel. Some guide wires are coated with Teflon (PTFE-Poly Tetra Fluoro Ethylene) to reduce the friction coefficient of the guide wire within the catheter. Care must be taken when a Teflon coated guide wire is passed through a metal needle or catheter hub, as the Teflon coating can become abraded and flake off.

Types.

Guide wires basically are of two shapes, straight and J curve and may have either a movable core or a fixed core.

Straight guide wires.

The straight guide wire has a straight flexible tip generally 3cm and is appropriate for passage through linear vessels. Since both the proximal (stiff) and the distal (flexible) ends of the straight guide wires appear the same, it is imperative to check the ends before insertion to ensure that the flexible end is inserted. Some straight wires are constructed such that both the ends are flexible to avoid this complication.

J-curve guide wires.

The J-curved guide wire generally has a 1.5, 3, or 6 mm-curved radius at its tip. It is particularly used for passage through tortuous vessels. Many percutaneous catheter introducer sets now supply a 3mm curve J curve wire with the kit. The larger 6mm curve wire is often appropriate for negotiating a tortuous femoral vein. An even larger 15mm radius J-curve is for cases of extreme vessel tortuosity.

A multipurpose, double end guide wires offer a long 10cm flexible straight tip at one end with a small 1.5 mm J curve on the other end. J-curve guide wires are packaged with a small plastic sleeve that when advanced over the tip of the guide wire straightens the guide

wire and permits it to be introduced into the hub of a needle or catheter.

Fixed core guide wires.

Fixed core guide wires have a straight inner core (mandrel) that is fixed at one end and terminates usually 3cm from the tip. The core provides necessary stiffness to the body of the coiled wire, while allowing flexibility at the tip. Variations in the tapered length 10 or 15 cm of the mandrel alter the degree of tip flexibility and improve the transition from the flexible distal coil to supported mandrel. This is an important feature in PTCA guide wires. Both straight and J-curved guide wires are available with a fixed core.

Movable core guide wires.

Guide wires that have the straight core attached only at the distal end are designed to allow movement of the tip of the core for the purpose of increasing or decreasing the length of the flexible tip. This is accomplished by not welding the proximal end of the mandrel core to the end of the guide wire, but instead securing the inner mandrel to a 5cm handle that is attached to the distal end of the wire. A variation in the tapered length of the core (5cm, 10cm, 15cm) alters the degree of tip flexibility or in case of J-curve wires the radius of the floppy wire. The ability to increase or decrease the length of the wire's flexible tip is desirable for traversing local areas of tortuosity. On occasion, re-advancement of inner mandrel may be difficult. This may be due to bending or kinking of the UN supported flexible tip. In general the inner mandrel should never be forced against a resistance, if this occurs the entire guide wire should be removed.

Core-flex guide wires.

Core-flex guide wires have a solid stainless steel mandrel with a safety wire constructed of spring coil soldered to the end. Although the outside diameter of this guide wire is 0.018inch, the guide wire offers the same support and rigidity as an 0.035 inch guide wire because of its solid steel mandrel. All usual PTCA wires are Core-flex wires. The rigidity of the guide wire permits percutaneous cannulation

with a smaller 22-gauge cannula rather than a standard 18-gauge cannula.

The Exchange guide wires.

The exchange guide wire is a straight wire unique only in its length of 260 or 300 cm. It is used exclusively for catheter exchange purposes as it is long enough to allow the removal and insertion of a catheter while the wire tip remains in the selected chamber. This is often used for exchanging catheters in LV if difficulty was encountered in crossing the aortic valve. Exchange guide wires are also used in coronary angioplasty.

Open-end guide wire/catheter.

This new type of Teflon wire has an open end and a removable inner mandrel. In this way it can function both as a guide wire and with the mandrel removed as a catheter. This guide is of greatest use in traversing tortuous or stenotic arteries during peripheral angioplasty. Pressures can be monitored during this procedure through the lumen of the guide wire.

Dimensions of the guide wires.

The guide wires for CAG are available in sizes ranging from 0.018 to 0.038 inches in diameter. The guide wire most commonly used for coronary angiogram is 0.035inch guide wire as it will support and passes through 6,7 and 8 Fr. catheters. Ideally the catheter tip should fit closely over the guide wire, with no excess space between the catheter and the guide wire. This promotes the smooth insertion of the catheter into the vessel.

Guide wires are available in lengths varying from 50 to 300 cm. The length of the specific catheter to be used dictates the guide wires length. The guide wire should be at least 20cm longer than the catheter to allow insertion of the catheter over the wire with some wires always exposed at the hub for manipulation and retention. Usually for coronary angiogram a 145cm wire is used.

Desired features of a guide wire.

Important considerations in guide wire selections include *stiffness, flexibility, and smoothness*. These features are important for both patient safety and improved operation.

Stiffness.

The stiffness of the guide wire is a function of both the core and the diameter of the wire. The wire must be sufficiently stiff to be pushed forward without collapsing back on itself or coiling within the vasculature. Increased stiffness can be obtained by use of a larger guide wire, if possible or use a guide wire with heavy-duty larger mandrel.

Flexibility.

The degree of flexibility of the guide wire is related to the core as well as to the single helix of stainless steel wire around it.

Smoothness.

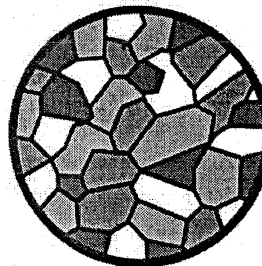
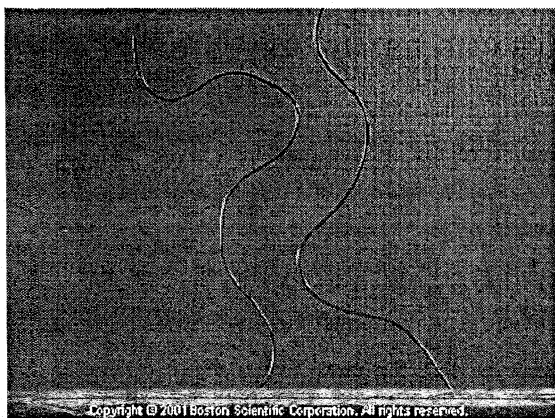
The smoothness of the wire is a result of its surface. Certainly the Teflon coated wires are smoother and produce less friction when passed through catheters. Some guide wires are treated with benzalkonium heparin to reduce thrombogenicity

MICROGUIDE WIRE

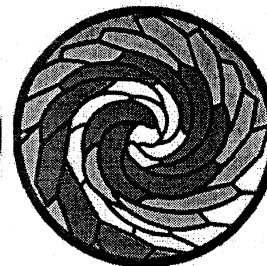
The selection of a micro guide wire can be difficult because there is the trade-off between increased flexibility and decreased ability of the microcatheter to pass over the microguidewire. Increased stiffness of the microguidewire may make traversing numerous curves very difficult. Most micro guide wires have a stain less steel core; a few have a nitinol core. This core provides most of the torquing. A few have polymer outer coatings and nearly all have hydrophilic coatings. The core diameter is so small that microguidewires are not radiopaque unless the manufactures have added platinum coils to the distal segments, distal radiopaque polymer coatings, or gold-tipped,

markers. Some microguidewires tip are shapeable, and others are pre shaped by manufactures. The softness of the tip, although important when the microguidewire has been deployed several centimeters, is moot when the micro guide wire is first emerging from the microcatheter because the microcatheter buttresses even the floppies tip to make it almost needle like on its initial emergence from the catheter. Almost all microguidewires have a hydrophilic coating that improves performance. The smallest micro guide wires for neuro interventional use are 0.08 inch in diameter; other ranges up to 0.018 inch in diameter.

TRANSEND STEERABLE GUIDE WIRE



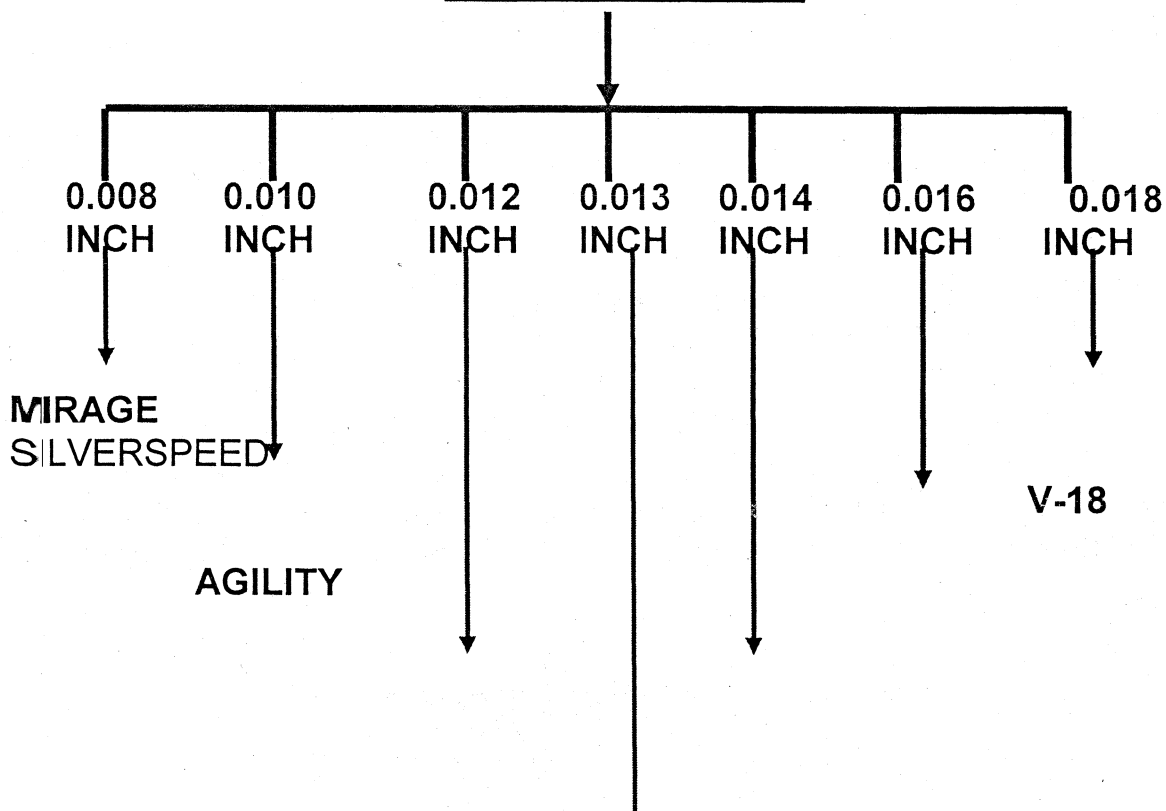
Conventional stainless steel alloy core construction



Proprietary SCITANIUM alloy core construction

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MICRO GUIDE WIRE



SPEED TRANSEND SILVER
 SILVER SPEED HEDLINER

HEDLINER AGILITY
 SILVER SPEED
 STRATEGY
 TRANSEND
 FasDasher
 CHOICE

MIZEEN

CORDIS

AGILITY-10 AGILITY-14

Wire diameter proximal (inch)	0.009	0.014
Wire diameter distal (inch)	0.010	0.014
Length (cm)	195	205,350
Core material	Stainless steel	Stainless steel
Outer jacket distal	Hydrophilic on metal	Hydrophilic on metal
Radiopaque Length (cm)	10 PT COIL	10 PT COIL
Hydrophilic coating Length (cm)	158	170
Shapeable tip	Yes	Yes
Reshapeable tip	No	No
Flexible length	35	42
Tip designs	Straight	Straight

Reshapeable tip	No	No	No
Flexible length	30	35	35,40
Tip designs	Straight	Straight	Straight

TARGET

Transend 10
HEDLINER

TRANSEND

FasDasher-14

MIZZEN

EX-.014

Wire diameter proximal (inch)	0.01	0.014	0.014	0.013	0.012 0.016
Wire diameter distal (inch)	0.010	0.014	0.014	0.013	0.012 0.016
Length (cm)	205	182,205	195	195	200
Core material	STAINLESS STEEL	STAINLESS STEEL	STAINLESS STEEL	NITINOL	NITINOL
Outer jacket distal	TUNGSTEN LOADED POLYMER HYDROPHILIC	TUNGSTEN LOADED POLYMER HYDROPHILIC	NONE	HYDROPHILIC	HYDROPHILIC
Radiopaque Length (cm)	205	182/205	4 PT COIL	5 GOLD COIL	2 GOLD COIL

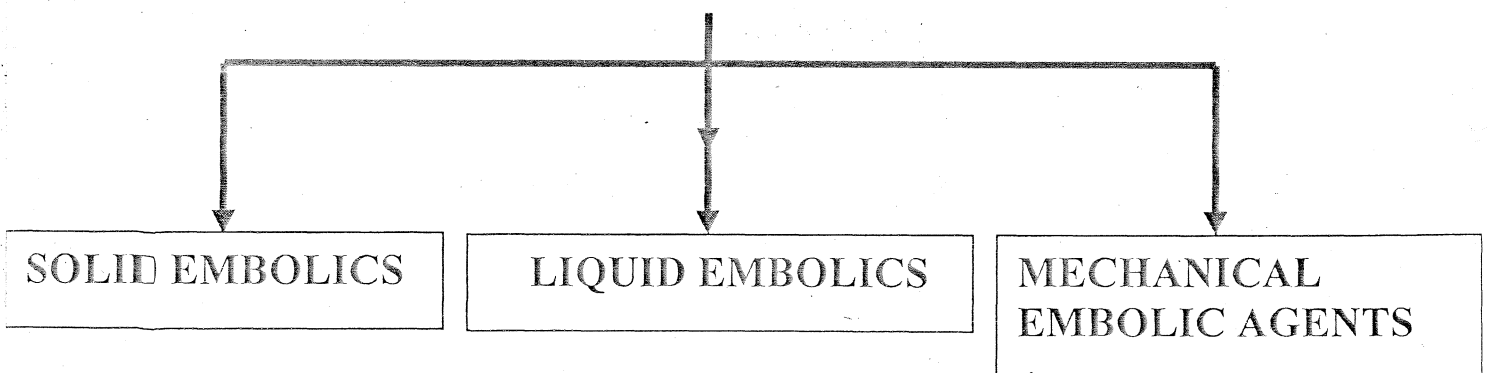
Hydrophilic coating Length (cm)	205	182/205	42	95	200
Shapeable tip	YES 2.0CM	YES 2.0CM	YES 4.0CM	YES 2.5CM	PRESHAPED
Reshapeable tip	YES	YES	NO	NO	NO
Flexible length	39	39	50	35/60	N/a
Tip designs	<u>straight</u>	<u>Straight</u>	<u>straight</u>	<u>straight</u>	<u>45°-70°</u> <u>90/150</u>

EMBOLISATION MATERIALS

The ideal embolic agent procedures should have several characteristics

1. Readily available
2. Easily prepared and injected
3. The agent should have little or no toxicity and cause minimal or no inflammation.
4. Embolic should produce reliable occlusion that will persist for the desired duration at times, permanently.
5. The agent should be radiopaque to illustrate delivery and should not hinder surgical removal of the embolised lesion.

EMBOLISATION MATERIALS



Solid embolic agents

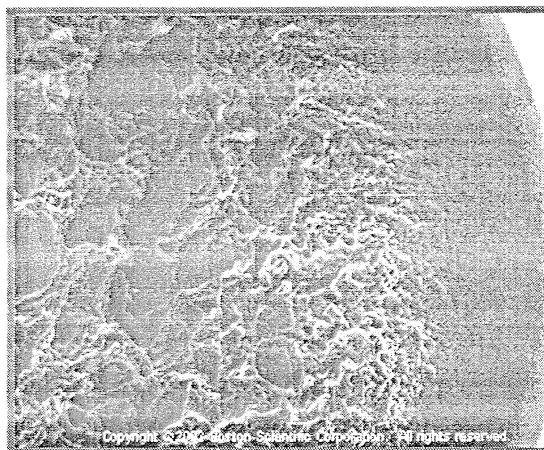
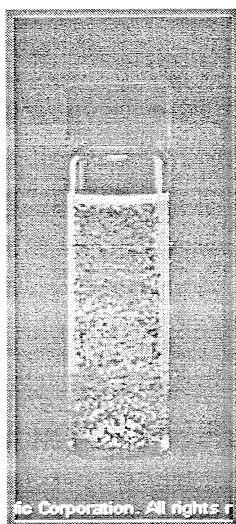
Solids that have been used as a embolic agents include

1. **POLYVINYL ALCOHOL (PVA) PARTICLES**
2. **SILK SUTURES**
3. **EMBOSPHERES**
4. **AVITENE**
5. **GEL FOAM**

All of these agents mechanically obstruct the vessels, which lead to slowing of the flow and eventually clot formation with in the embolised vessel.

Polyvinyl alcohol

These are artificial embolisation devices. These devices are intended to provide vascular occlusion up on selective placement via an angiographic catheter. It is nearly water insoluble polymer made by the reaction of foamed PVA with formaldehyde. The polymer is not linear but composed of cross-linked chains that determines the physical properties such as swelling, suspension in contrast material, and viscosity. They are available in different sizes and different brand names.



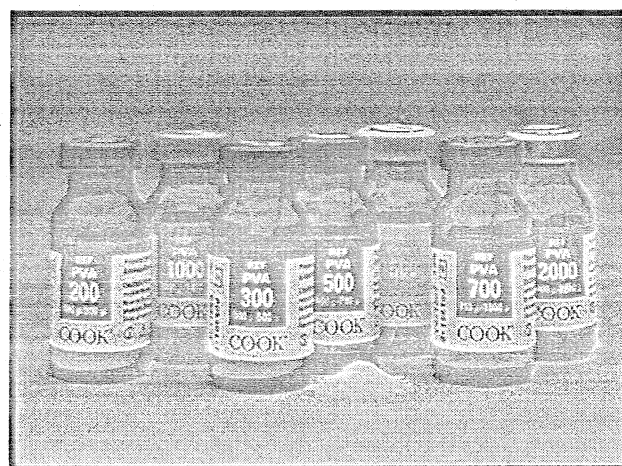
BRAND NAME MANUFACTURES

BIODYNE	COOK
CONTOUR	BOSTON

COOK

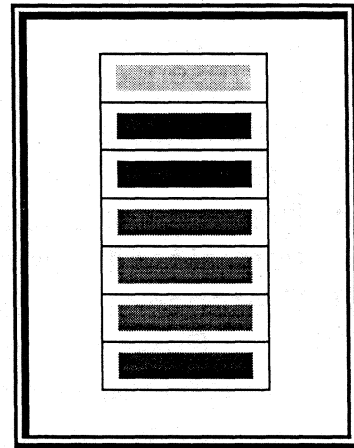
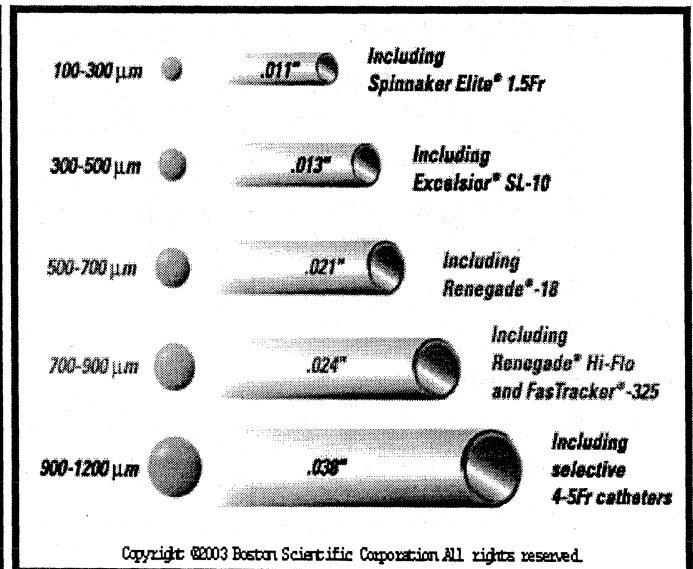
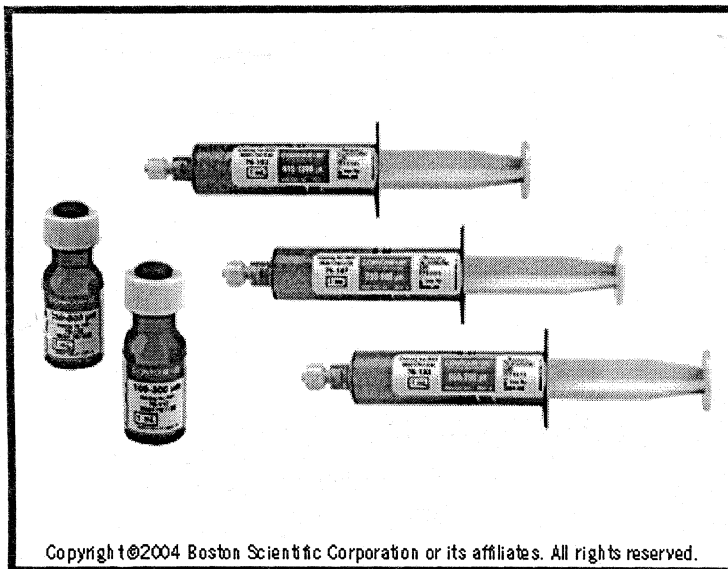
SIZE RANGE

50-100 μ
 100-200 μ
 200-300 μ
 300-500 μ
 500-700 μ
 700-1000 μ
 1000-1500 μ
 1500-2000 μ
 2000-2800 μ



BOSTON

SIZERANGE
45-150 μ
150-250 μ
250-355 μ
355-500 μ
500-710 μ
710-1000 μ
1000-1180 μ

**Contour SE™ Microspheres****INDICATION FOR USE**

Emboli are used for the embolisation of hyper vascular tumours and arteriovenous malformation.

CONTRA INDICATIONS

1. Patient intolerance to temporary occlusion procedures
2. Vascular anatomy or blood flow precludes stable, selective emboli or catheter placement.
3. Presence of likely onset of vasospasm.
4. Presence of likely onset of hemorrhage
5. Presence of arteromatous disease.
6. Presence of feeding arteries smaller than distal branches from which they emerge
7. Presence of patient extra-to-intracranial anastomoses or stimuli
8. Presence of collateral vessel pathways potentially endangering normal territories during embolisation
9. Presence of end arteries leading directly to cranial nerves
10. Presence of arteries supplying the lesion not large enough to accept emboli
11. Vascular resistance peripheral to the feeding arteries precluding passage of emboli in to the lesion.

POTENTIAL COMPLICATIONS

1. Clot formation of the tip of catheter and subsequent dislodgement
2. Ischemia at an undesirable location
3. Capillary bed saturation and tissue damage
4. Ischemic stroke or ischemic infarction
5. Vessel or lesion rupture and hemorrhage
6. Recurrent hemorrhage.
7. Vasospasm
8. Death
9. Undesirable reflux or passage of emboli into the arteries adjacent to the target lesion or through the lesion into the other arteries or arterial beds such as internal carotid artery, pulmonary or coronary circulation.

When the PVA particles used for embolisation of the AVMs, the choice of particle size is based on the flow through the lesion. For PVA embolisation of tumors, the particles are as small as possible to occlude as distally as possible. Epistaxis embolisation usually is performed with 200 μ m- sized PVA particles,

which lessens the risks of mucosal slough that can occur when smaller particles are used.

SILK SUTURES

Some operators have used silk suture in the embolisation of AVMs, The 5-0 or 6-0 silk suture is cut in to appropriate lengths (5mm to 30mm), suspended in contrast material because the silk is not radiopaque, and injected through a micro catheter in to a feeding pedicle of the AVMs, until the flow slows. Further injection may cause reflux of the silk in to more proximal vessels. The silk is easily sectioned during the surgical removal of the AVM.

The two major advantage of the silk are

1. Ready availability
2. Low cost

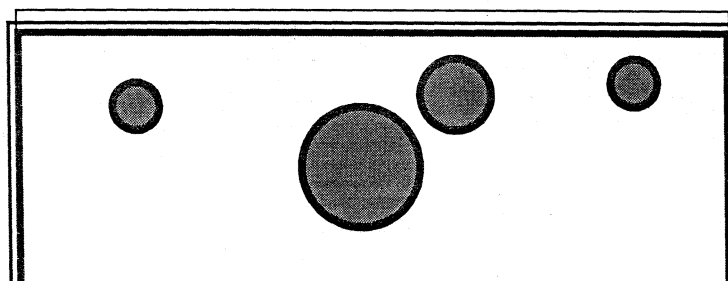
The disadvantages include,

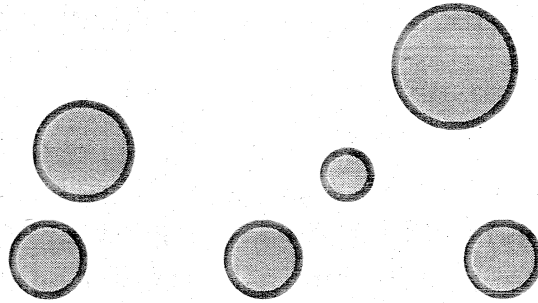
1. Pressure necessary for injection
2. The difficulty in determining the site of deposition. (feeding arteries or draining veins)
3. The acute intravascular and extravascular inflammation that ensues.

EMBOSPHERES

Embospheres is a new particle product that has been specially designed for particle embolisation. It is composed of an acrylic polymer core covered with collagen in gelatin form. The surface is hydrophilic and smooth, which prevents aggregation. These particles are somewhat compressible for delivery through the micro catheter. They obstruct when the vessel lumen size matches their size. Sizes, which are more uniform than with PVA particles, range from 40-1200 μ m. The tissue reaction to embospheres and PVA is similar, both embolics elicit chronic transmural and per vascular inflammation.

EMBOSPHERES





AVITENE

It is a powder composed of purified bovine collagen. The micro fibrils, which are 75-150 μ m in size, function by obstructing the vessels. Avitene is designed to attract platelets to the fibrils and to activate the platelet to release the aggregation factors that result in a thrombus in the interstices of the fibrous mass. It is less effective in thrombocytopenic patients. The powder is absorbable and therefore is a temporary agent if fibrous tissue has not replaced the thrombus. It stimulates a mild chronic cellular inflammatory response. It has been used as a preoperative embolic agent for hyper vascular head and neck lesion.

GELFOAM

It is a water insoluble, porous, pliable product that is absorbed by the body in 7-21 days with little tissue reaction. It is prepared from purified pork skin and should not be used in patients with known allergies to porcine collagen. The haemostatic properties are not fully understood but seem to be due to the formation of an artificial clot as matrix forms for platelets that are damaged by contact with the gelfoam. This cause release of thromboplastin, which reacts with thrombin and calcium to produce thrombin, which acts on the fibrinogen in the blood. Gelfoam has the ability to absorb and hold up to 45 times its weight of blood and other fluids. This capacity makes it a very effective vascular occlusive agent. It is available in a sponge

form that can be cut into the desired pledged size, suspended in contrast, and injected through the micro catheter for non permanent occlusion of larger arteries supplying the lesion. Gel foam is not radiopaque, and the contrast in which it is suspended serves as an indirect indication of its course and destination. Although compressible, the gel foam pledges can lodge in the catheter, causing it to become obstructed. The delivery catheter must not be over pressurized or the catheter will rupture and result in a proximal embolisation.

LIQUID EMBOLICS

Liquid agents for neuro interventional use consist of

CYANOACRYLATES (GLUE)

- Histoacryl-(n-butyl 2-cyano acrylates) is commonly used
- Need skill full & care full handling.
- Capable of reaching distal small vessel.
- Exposure of glue to the ionic solution causes polymerization.
- Polymerization can be slowed by addition of iophendylate or glacial acetic acid.
- Tantalum, bismuth or lipidol gives better radiopacity to the glue.
- Speed of the polymerization can control by addition of lipidol.

Complications

1. Stroke due to occlusion of undesired branches.
2. Obstruction of venous out flow.
3. Obstruction of catheter.
4. Polymerization leads to a degree of angioneclerosis.

HISTOACRYL CONCENTRATION CHART

NO	CONCENTRATION	HISTOACRYL	LIPIDOL
1	15%	0.5ml	2.8ml
2	17%	0.5ml	2.4ml
3	20%	0.5ml	2ml
4	22%	0.5ml	1.7ml
5	25%	0.5ml	1.5ml
6	33%	0.5ml	1ml
7	40%	1ml	1.5ml
8	50%	0.5ml	0.5ml
9	60%	1.5ml	1ml
10	66%	1ml	0.5ml
11	75%	1.5ml	0.5ml
12	80%	2ml	0.5ml

DEHYDRATED ALCOHOL

It is a liquid agent used in the same way as cyanoacrylates for the treatment of AVM's and some tumors. In the past the alcohol was opacified by dissolving metrizamide powder in it, and the solution was injected under fluoroscopic control. Because metrizamide powder is no longer available, operators opacify the alcohol with a small amount of concentrated nonionic contrast material.

Alcohol injures tissue by denaturing proteins of the cell wall, particularly the endothelial cells, and causing precipitation of the protoplasm. This leads to the thrombus formation and a coagulative necrosis. Alcohol injection is very painful, general anesthesia is usually required. The maximum volume of alcohol used in a treatment session is 1cc/kg body weight and this is usually well tolerated. The alcohol may cause a significant rise in pulmonary vascular resistance and pulmonary arterial pressures.

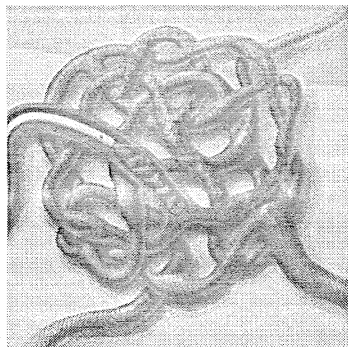
Besides brain AVM's and head, neck, and spine tumors, facial venous malformations can be treated with successfully with these agent Care must be taken to confine the alcohol to the venous mal formation, because tissue necrosis and superficial skin necrosis can be significant complications.

ETHIBOLIC

This is a 60% alcohol solution of zein (corn protein), sodium diatrizoate (ionic contrast), oleum papaveris, and propylene glycol

(for sterility). This biodegradable solution produces thrombosis, necrosis, and a fibrotic reaction with a short -lived inflammatory response that may produce pain and fever, possibly requiring medical management. Ethibolic is safer and produce less ischemia and necrosis of the skin than alcohol does. Injection is less painful than alcohol injection is; it does require sedation and analgesia but not general anesthesia.

ONYX



This liquid is a proprietary ethylene alcohol copolymer suspended in DMSO and opacified with tantalum powder. It stays in liquid form until it contacts blood or other aqueous solutions. The onyx then begins to precipitate, quickly changing from a liquid to a solid from the out side to the inside. Its major advantage is that it adheres to itself but not to the delivery catheter, so that slow injections with slight reflux along the micro catheter tip can be used without fear of adherence of the cast to the micro catheter. However, if significant reflux occurs, catheter retrieval may be impossible.

HYDROGEL

- Available in either liquid or particle form.
- It forms soft shapeless mass when exposed to water and not produce inflammatory response.
- Similar to EVAL (ethylene vinyl alcohol co-polymer)
- Delivered through micro catheter.

GDC COILS
MATRIX 3D-COILS
COOK DETACHABLE COILS
MTI COILS

MECHANICAL EMBOLIC AGENTS

RETRIEVABLE COILS

NON-RETRIEVABLE
COILS.

GDC COILS (electrically detachable)

- They are platinum micro coils.
- It is soldered to one end of an insulated stainless steel

TRUFILL PUSHABLE COILS
FIBERED PLATINUM COILS
LIQUID COILS

- guide wire, a short segment of the soldered end is exposed
- After coil positioning, low voltage current employs electrolysis to detach coils.
 - Advantage is withdrawal of the coil before final placement.
 - Available in two systems
 1. GDC-18 SYSTEM
 2. GDC-10 SYSTEM

They are available in various size and shape.

- STANDARD COILS
- 2D-COILS
- 3D-COILS
- SOFT COILS
- ULTRA SOFT COILS
- STRECH RESISTANT COILS
- THROMBOGENIC GDC COILS

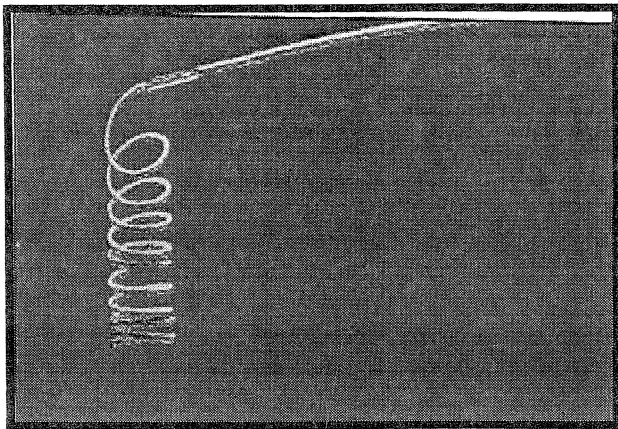
GDC-10, 18 & DELIVERY WIRE

The GDC coils are manufactured from a platinum alloy, which permits them to conform to the often-irregular shape of saccular aneurysm. The delivery wire has been designed to provide two important benefits of the GDC technology:

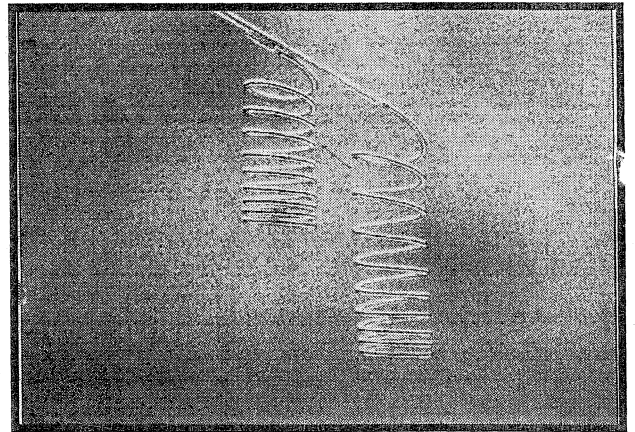
- 1. CONTROLLED DEPLOYMENT.**
- 2. ELECTROLYTIC DETACHMENT.**

The GDC system is a fast, accurate and effective endovascular approach to treating high surgical risk intracranial aneurysm. At the heart of the GDC system is a soft, platinum coil that is attached to a stainless steel delivery wire. The softness of platinum allows the coil to conform to the often-irregular shape of intracranial aneurysm. And because the coils are available in a variety of helical diameters and lengths, the interventionalist can deploy and release the optimal combination of coils necessary to occlude the aneurysm sac.

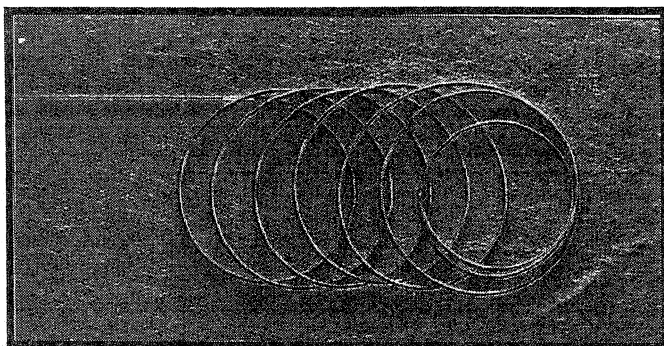
**GDC-18&10 SYSTEM STANDARD COILS
18&10 SOFT COILS**



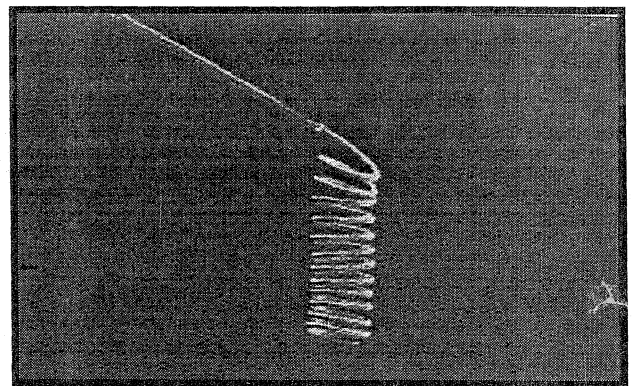
GDC

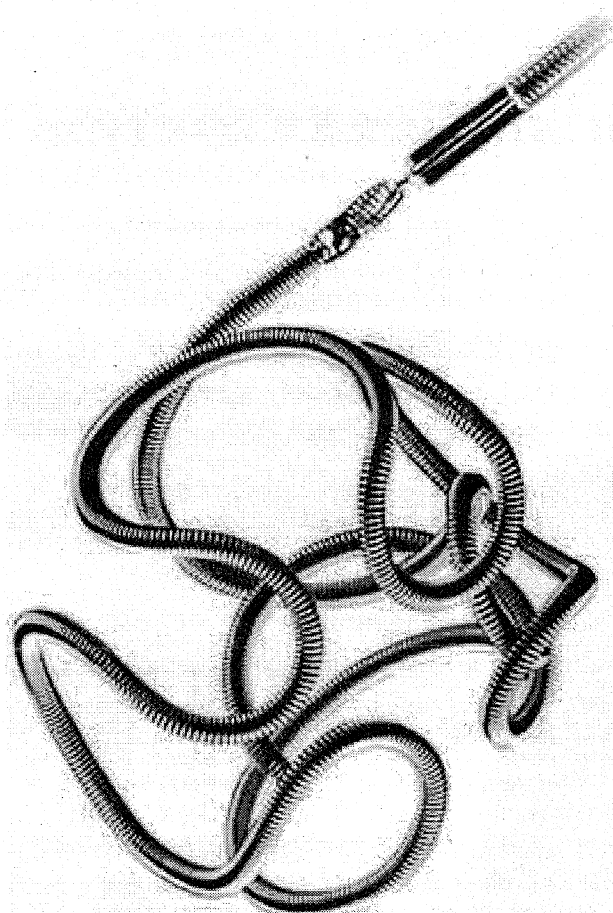


GDC 18&10 2-DIAMMETER



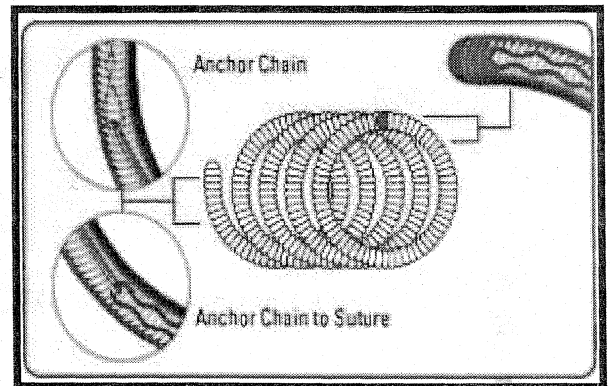
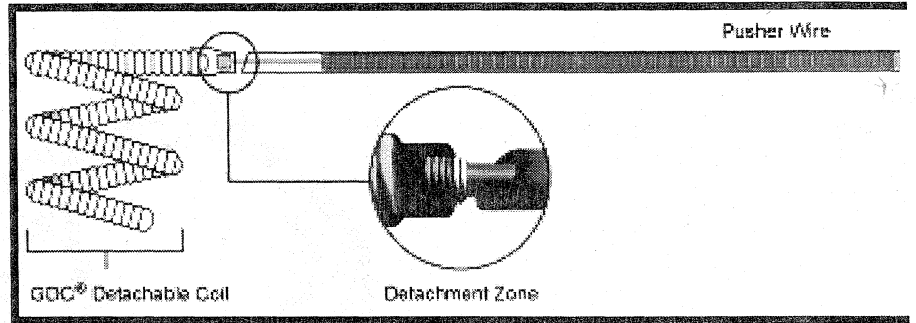
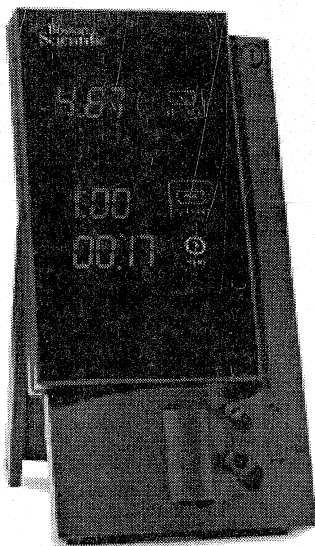
GDC SOFT SR



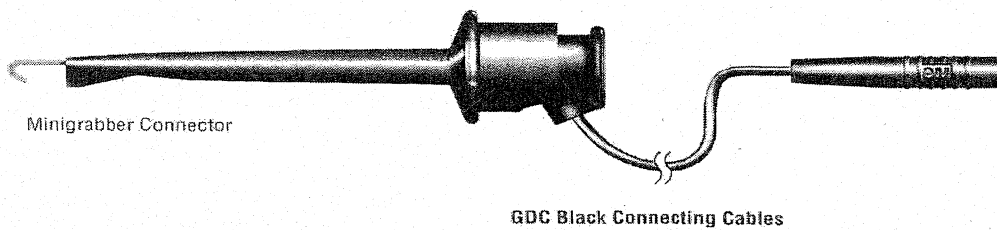
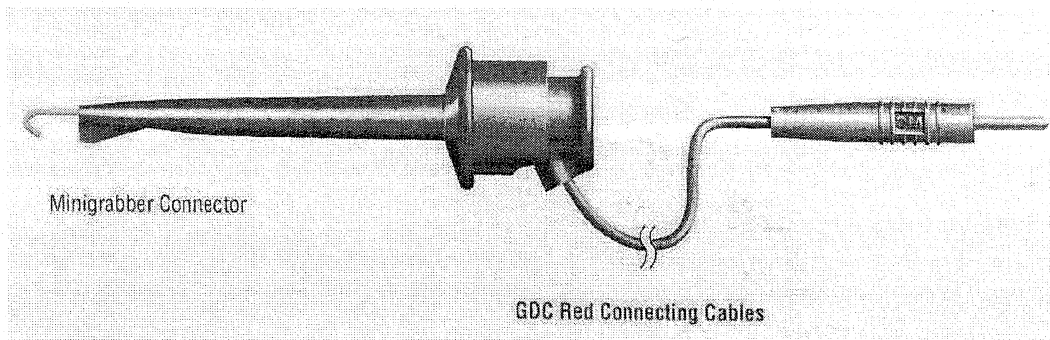


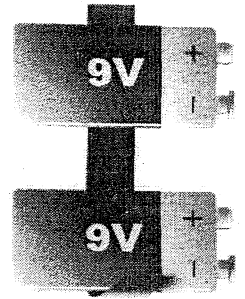
GDC 18 3D SHAPE

GDC-DELIVERY SYSTEM



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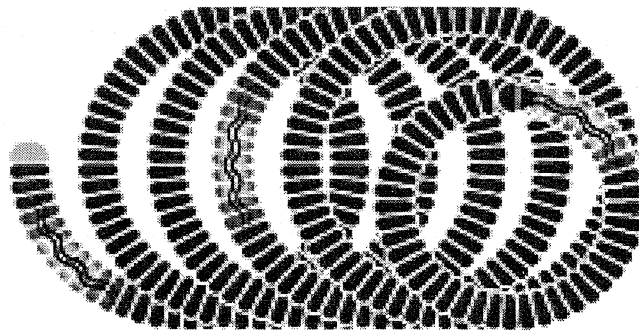
MATRIX COIL

Matrix detachable coils are platinum coils covered with an absorbable polymer and attached to a stainless steel delivery wire. Matrix detachable coils are designed for use with Boston scientific target 2 tip infusion catheters and a GDC power supply.

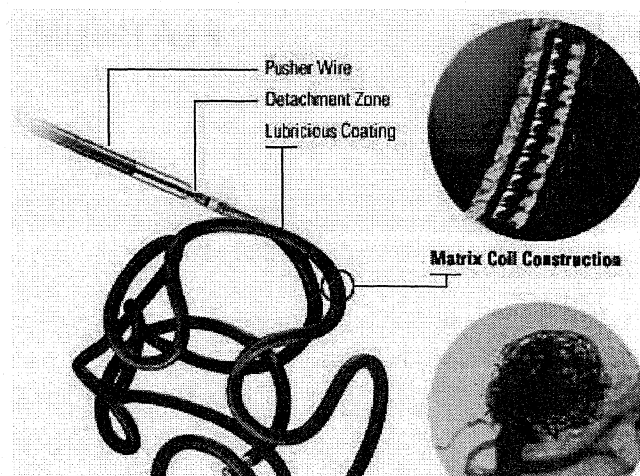
Matrix detachable coils:-

- Increased the amount of mature connective tissue present with aneurysm at 14 days.
- Increased neck tissue thickness of aneurysm at 14 and 90 days.
- Reduced cross sectional area of the aneurysm at 90 and 180 days as measured from angiograms and histological sections.
- Did not impact average parent artery diameter.

MATRIX 2D-COILS



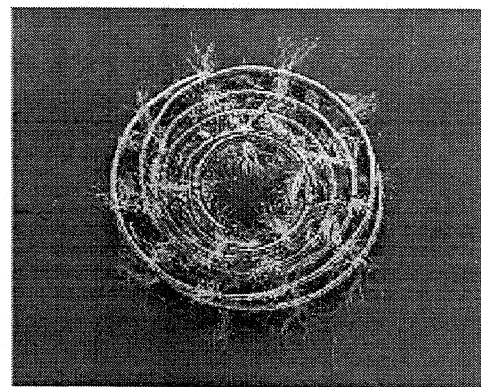
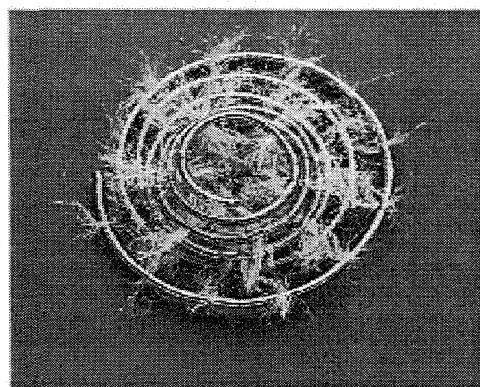
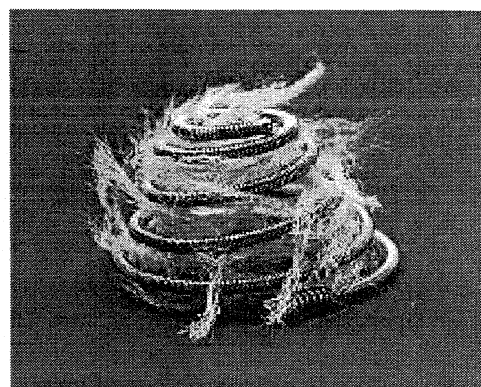
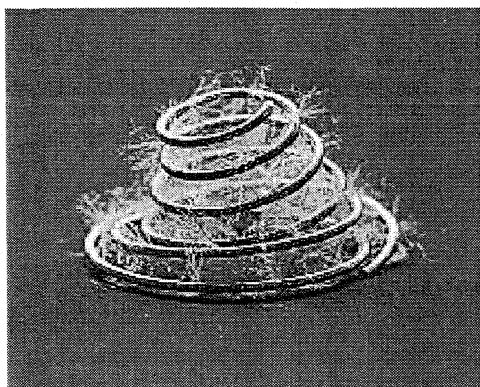
MATRIX 3D-COILS



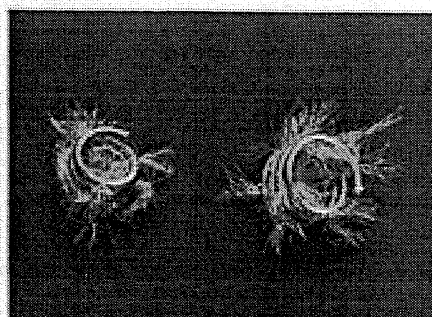
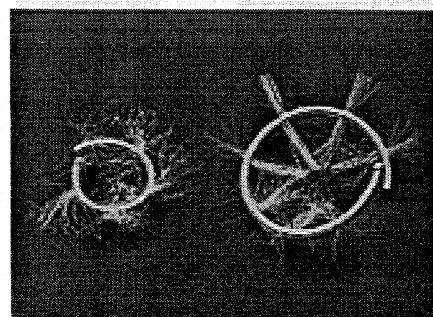
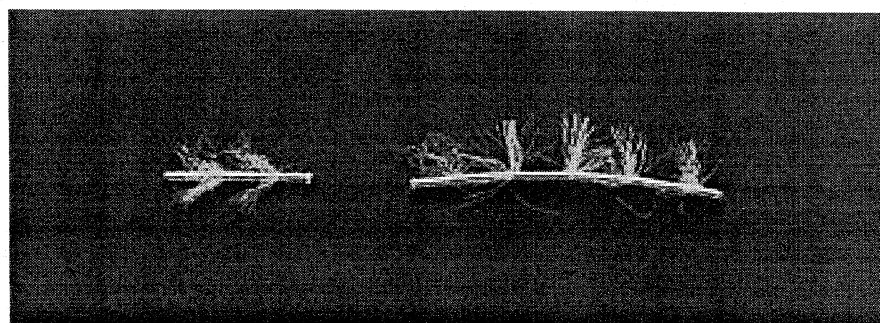


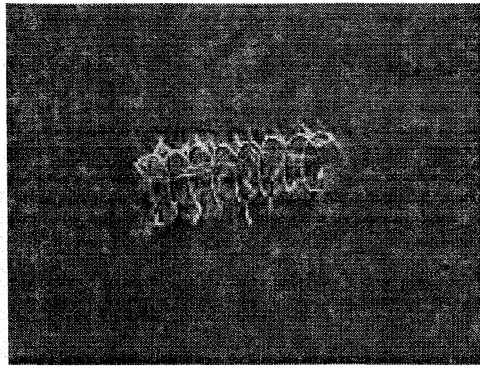
TORNADO®
EMBOLISATION COILS
PLATINUM .018 INCH

EMBOLISATION COILS
PLATINUM .035 INCH

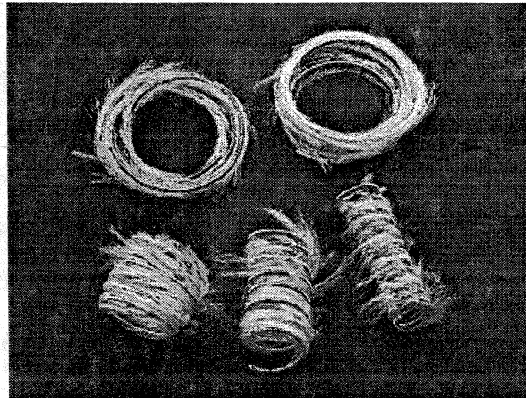


MICRONESTER® PLATINUM EMBOLIZATION COILS PLATINUM
HILAL EMBOLISATION COILS PLATINUM
PLATINUM

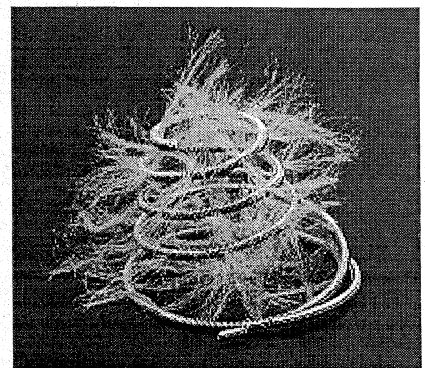
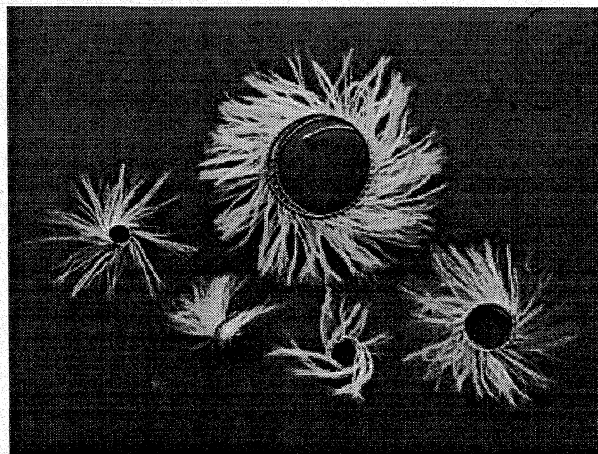




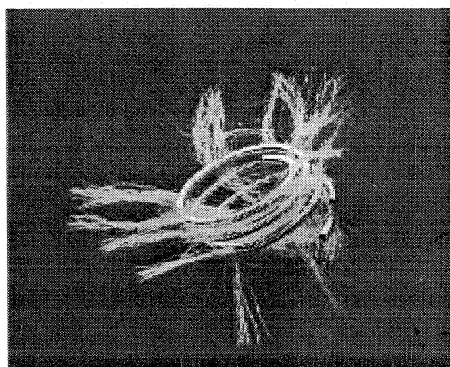
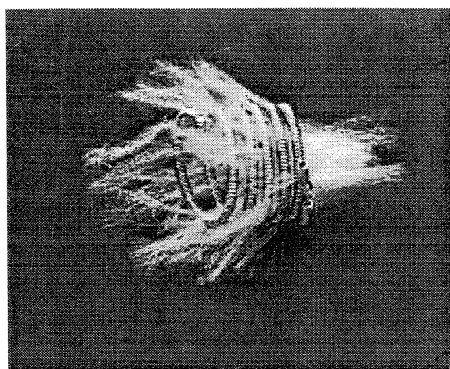
**NESTER® EMBOLIZATION COILS
PLATINUM**



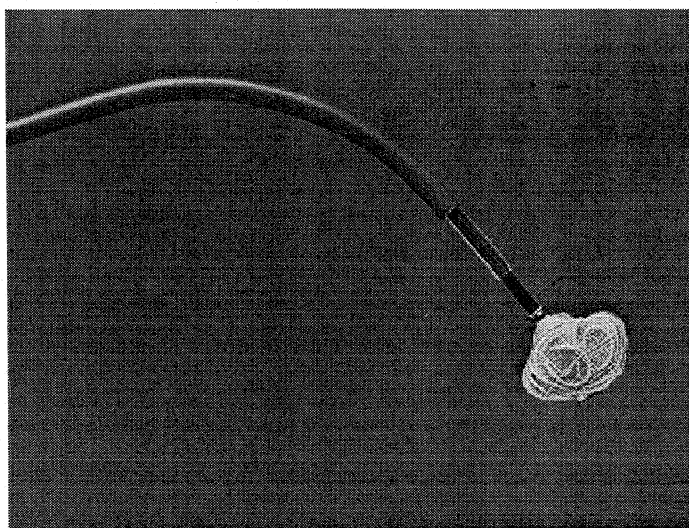
**DETACHABLE TORNADO®
EMBOLIZATION COILS - PLATINUM
EMBOLIZATION COILS - STAINLESS STEEL**



DETACHABLE EMBOLIZATION COILS



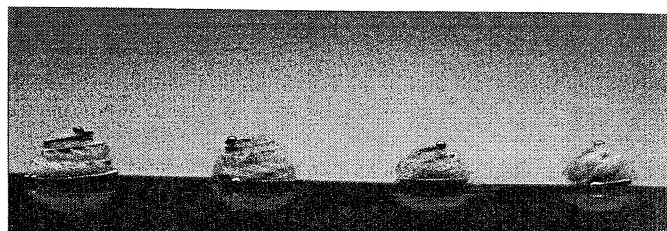
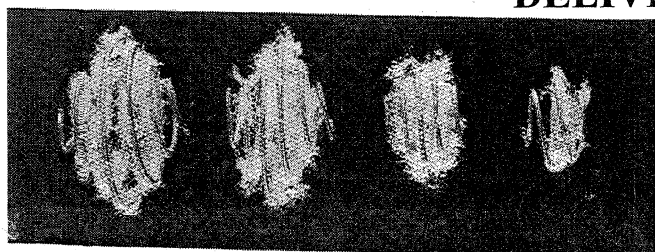
GIANTURCO-GRIFKA VASCULAR OCCLUSION DEVICE



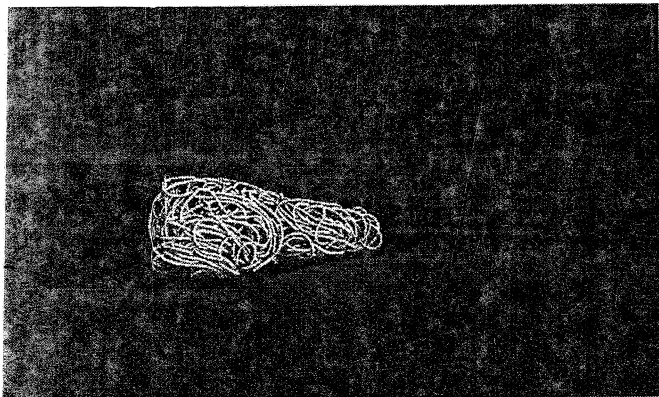
VORTX DIAMOND SHAPE COILS

VORTX 35 COILS

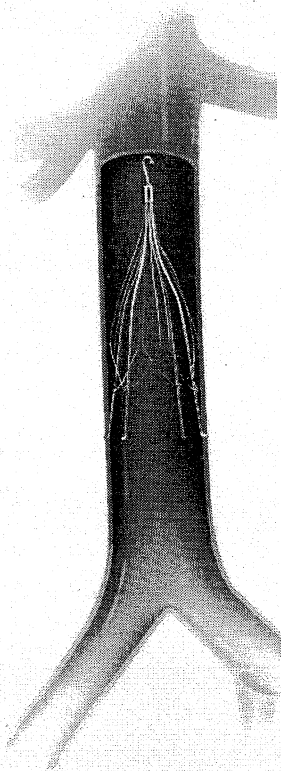
**FLIPPER™ DETACHABLE EMBOLIZATION COIL
DELIVERY SYSTEM**



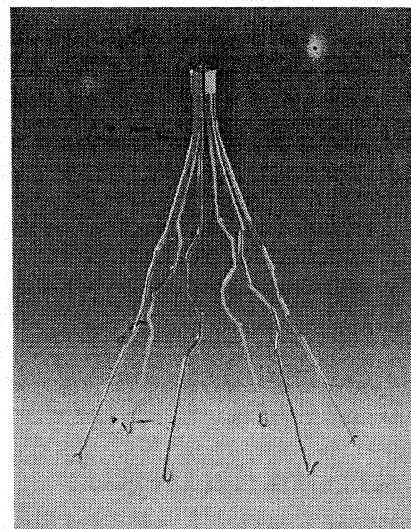
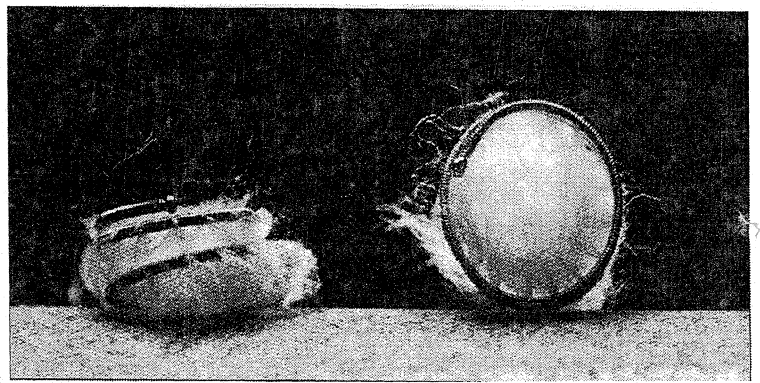
BERNSTEIN LIQUID COILS



IVC FILTER



**.035 FIBERED PLATINUM
MICRO COILS**



The design of a Gunther tulip venacava filter is a complex construction consisting of 4 main struts 0.45 mm in diameter. Each struts carries an elongated wire loop extending along threefourths of its length. A cross sectional view of Gunther tulip filter demonstrates twice the number of filter wire found as compared to the leading conical filter design

FEMORAL AND JUGULAR SYSTEM:

Used for the prevention of recurrent pulmonary embolism via placement in the venacava in the following situations.

1. Pulmonary thrombo embolism when anticoagulation therapy is contraindicated.
 2. Failure of anticoagulation therapy in thrombo embolic disease.
 3. Emergency treatment following massive pulmonary embolism where anticipated benefits of conventional therapy are reduced etc...
-

UROKINASE

Is a plasminogen activator and is used as a thrombolytic agent. Urokinase is an enzyme obtained from the human urine or from the tissue cultures of human kidney cells. It converts tissue plasminogen to enzyme plasmin. Plasmin degrades fibrin clots, fibrinogen and other plasma proteins.

INDICATIONS

- Coronary artery thrombosis associated with an evolving transmural myocardial infarct
- Acute pulmonary embolism

- Deep vein thrombosis
- Peripheral vascular thrombo-embolism

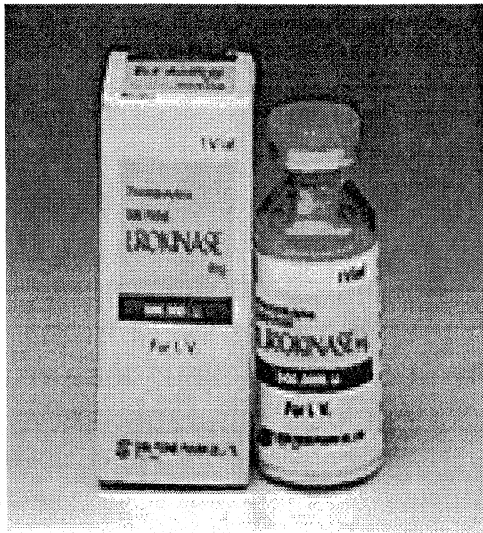
Urokinase therapy should be started 6 hours after the onset of symptoms.

Loading dose: 4,400 i.u/ kg of urokinase re-constituted in 15 ml of sterile physiological saline or 5% dextrose and give i.v. over 10 minutes.

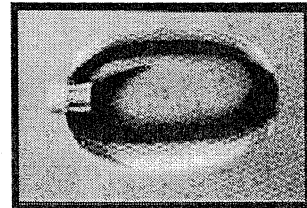
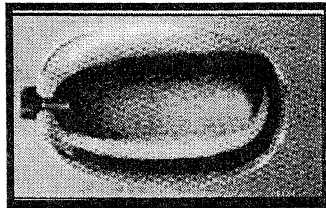
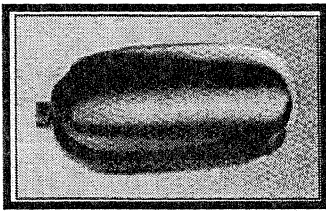
Maintenance dose:

On the first day, the loading dose is followed by 4,400 i.u/kg of urokinase dissolved in 200 ml of sterile physiological saline is given over a period of 8-12 hours. On the second day, the therapy is continued with a daily dose of 4,400 i.u/kg. The total duration of the treatment usually lasts for 6-13 days. For a longer duration, higher doses can be given depending on the clinical urgency, hematological status and the findings of the angiography of the patient.

-+



DETACHABLE BALLOONS



Detachable balloons are designed to occlude the entire vessel with one application. Balloons can sometimes deflate and migrate causing a stroke; they can cause embolus formation and release, vessel rupture and hemorrhage, or infarction. They may also cause Neurological or functioning deficit, vasospasm, ischemia at an undesirable location and death.

Contraindications to balloon occlusion are the patient intolerance to temporary occlusion procedures. Vascular anatomy or blood flow that precludes stable catheter placement. The presence or likely onset of vasospasm. Friable vascular constriction that may cause rupture of a blood vessel and clips, bone fragments, calcifications or any irregularities which may cause damage to the balloon.

catheter over a guide wire. The lumen can also be used for infusion of contrast medium.

The catheter shaft tapers beneath the balloon segment to achieve the lowest possible deflated profile.

Radiopaque markers are placed under the balloon segment of the catheter to provide visual reference points for balloon positioning within the vessel.

INDICATIONS FOR USE

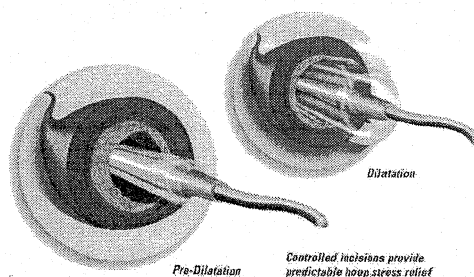
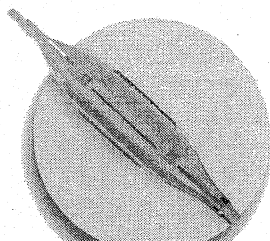
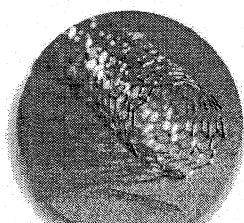
Balloon dilation catheters are recommended for percutaneous transluminal angioplasty of the iliac, femoral and renal arteries and for the treatment of obstructive lesion of native or synthetic arteriovenous dialysis fistulae.

POTENTIAL COMPLICATIONS

The complications that may result from a balloon dilation procedure include:

- Vessel perforation
- Vessel spasm
- Hemorrhage
- Hematoma
- Hypotension
- Pain and tenderness
- Arrhythmias
- Systemic embolisation
- Endocarditis
- Vascular thrombosis
- Drug reactions
- Allergic reaction to contrast media
- Pyrogenic reactions
- Arteriovenous fistulae
- Thromboembolic episodes
- Vessel dissection
- Infection
- Death

CUTTING BALLOON



The Rotational angioplasty system is a catheter based angioplasty device utilizing a diamond – coated elliptical burr at the tip of flexible drive shaft. Tracking co-axially over a guide wire and rotating at upto 190,000 RPM, the burr ablates plaque into fine particles that are disposed by the body's reticuloendothelial system.

Indications for use

Percutaneous rotational angioplasty is a sole therapy or adjunctive balloon angioplasty, is indicated in patients with coronary artery disease who are acceptable candidates for coronary artery bypass graft surgery and who meets one of the following selection criteria*single vessel atherosclerotic coronary artery disease with a stenosis that can be passed with a guide wire*multiple vessel coronary artery disease that in the physician's judgement does not pose undue risk to the patient. *Certain who have had prior percutaneous transluminal angioplasty (PTCA), and who have a restenosis of the native vessel *native vessel atherosclerotic coronary artery disease that less than 25 mm in length.

Contraindications

1. Occlusions through which a guide wire will not pass.
2. Last remaining vessel with compromised ventricular function
3. Saphenous vein grafts
4. Angiographic evidence of thrombus prior to treatment with rotational angioplasty. Such patients may be treated with thrombolytics (e.g. urokinase when thrombus has been resolved for two or four weeks, lesion may be treated with the rotational angioplasty.
5. Angiographic evidence of significant dissection at the treatment site. The patient may be treated conservatively for approximately four weeks to permit the dissection to heal before treating the lesion with rotational angioplasty.

STENTS. Desirable stent characteristics 1. Biocompatibility.

2. Metallurgic properties

- * **LOW THROMBOGENICITY**
- * **CORROSION RESISTANCE**

3. High radiopacity

4. Physical properties

- *High expansion ratio
- *Predictable expanded size.
- *Stable after delivery

Resistant to external deformation once deployed.

(Palmaz > Wall stent > Strecker.)

- *Available in a range of lengths and diameters.
- *Delivered on a low profile catheter.
- *Precise delivery.
- *Retrievable in case of errant placement.
- *Amenable to non-invasive imaging follow up,
Including MRI.

5. Inexpensive.

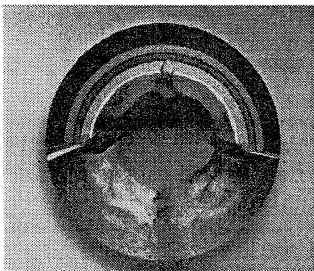
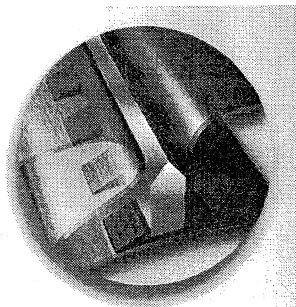
Basic types – Self expanding / Balloon expanded
Covered / Uncovered.

SELF-EXPANDING STENT

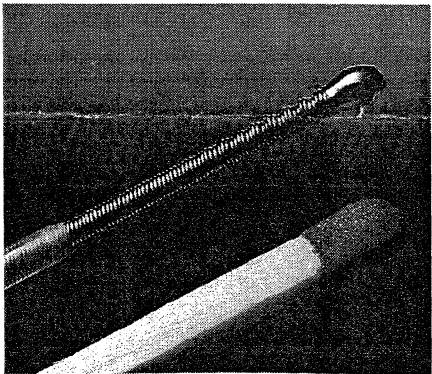
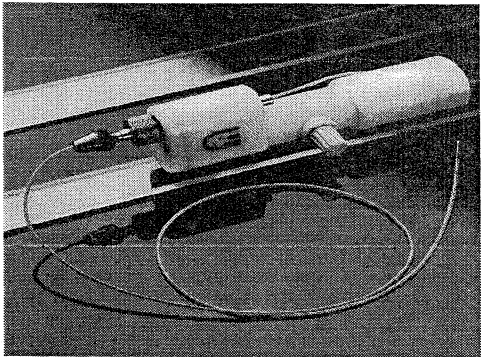
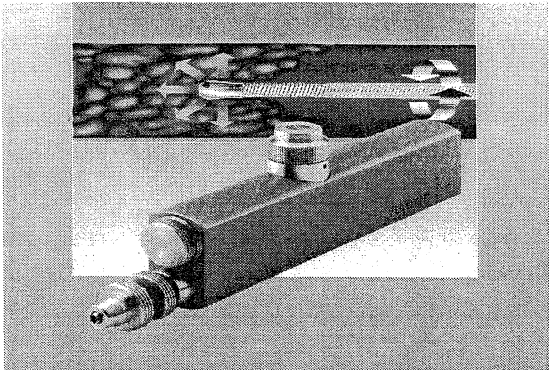
Works by

1. Spring action triggered by unloading the
Device from delivery catheter.
2. Thermal shape memory - Stent assumes
Configuration when warmed to body temperature

SELF-EXPANDING STENTS



Rotational Angioplasty (Rotablator)



1. Wall stent (Schneider)

Advantages 1. Flexible

2. Available in varying lengths

Drawbacks 1. Poor radiopacity.

2. Maximal shortening (30-40%)

Other uses

TIPS, esophageal dilatation,

Tracheobronchial, Enteral end prosthesis

SELF-EXPANDING STENTS

2. Nitinol spring coil stent. Made of nickel-titanium alloy; thermally triggered shape.

3. Gianturco Z stent (Cook)

Stainless steel; zigzag wire

4. Memotherm stent (Bard):

Nitinol stent with diamond pattern.

BALLOON EXPANDABLE STENTS

1. Palmaz stent (Johnson & Johnson).

Made of stainless steel.

Advantages: Ease and accuracy of delivery.

More radiopaque than Wall Stent.

Excellent compression resistance.

Disadvantages:

Lack of flexibility.

Needs large sheath for introduction

BALLOON EXPANDABLE STENTS

2. Strecker stent.

High radiopacity.

Non-ferromagnetic; so MR compatible

3. Cordis stent

Made of Tantalum; flexible in undeployed state

4. Gianturco-Roubin (Book binder) Flex stent. Stainless steel stent.

5. Wiktor stent

Sinusoidal wire; made of Tantalum

COVERED STENTS : Stents with a sleeve (cover) of thin Expandable material.

Uses:

*To exclude aneurysms, both true and false.

*To exclude traumatic vessel disruption

*In esophagus / trachea to cover fistulae

COVERED STENTS

1. Cragg endopro system: (Mintec) Cragg nitinol stent covered with 0.1 mm thin

Dacron.

2. Wall Graft Covering sleeve made of polyethylene terephthalate.

3. Bifurcated stent system also available for AAA Repair.

COVERED STENTS

Contraindications:

1. Lesion within 0.5 cm of bifurcation or essential branch.

2. Patients in whom ant platelets, anticoagulants or thrombolytic drugs are contraindicated.

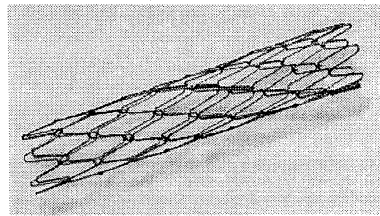
3. Connective tissue disorder.

4. Mycotic vascular lesion.

5. Where lesion crosses a joint, Wall Graft cannot

be used.

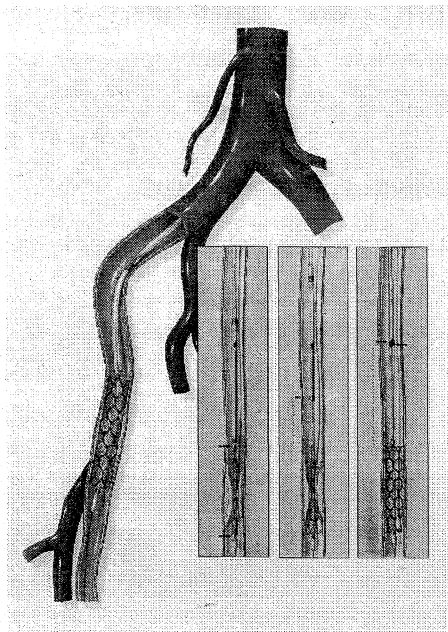
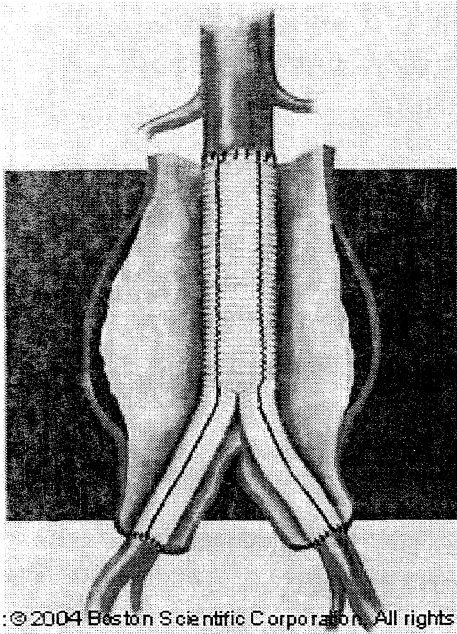
Vascular Stents:



Stents are metallic endo prosthetic materials, which help to maintain the patency obtained through angioplasty. They can also be percutaneously deployed in the precise location, under imaging guidance.

These procedures are extensively used in diseases of blood vessels of neck, limbs, heart etc.

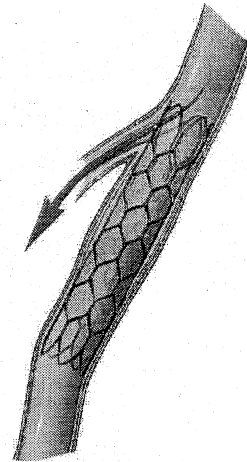




Arterial Vessels

Otherwise referred to as 'Therapeutic Embolisation', using particulate materials delivered through micro catheters can be used in settings of acute bleeding from torn vessels, vascular malformations, Aneurysms, Vascular tumors. Embolic materials in use include alcohol, metallic coils, gel foam, etc.

Vascular Therapy



This includes 'thrombolytic therapy' - meaning lysis of clotted blood inside important vessels and 'Selective Chemotherapy' - designed to administer ideal therapeutic concentrations to a tumor.

II. Non Vascular Interventions

1. Biopsy Procedures

Samples can be taken even from inaccessible sites of human body using special needle systems, under imaging guidance for diagnostic purposes.

2. Percutaneous Drainage Procedures

Drainage of abscesses and collections can be done by percutaneous needle placement, which would otherwise need open surgery.

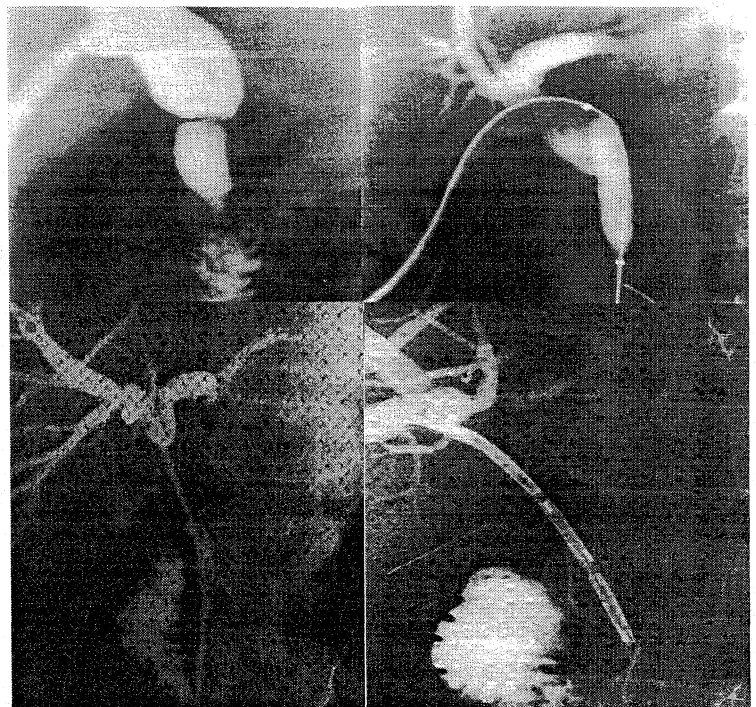
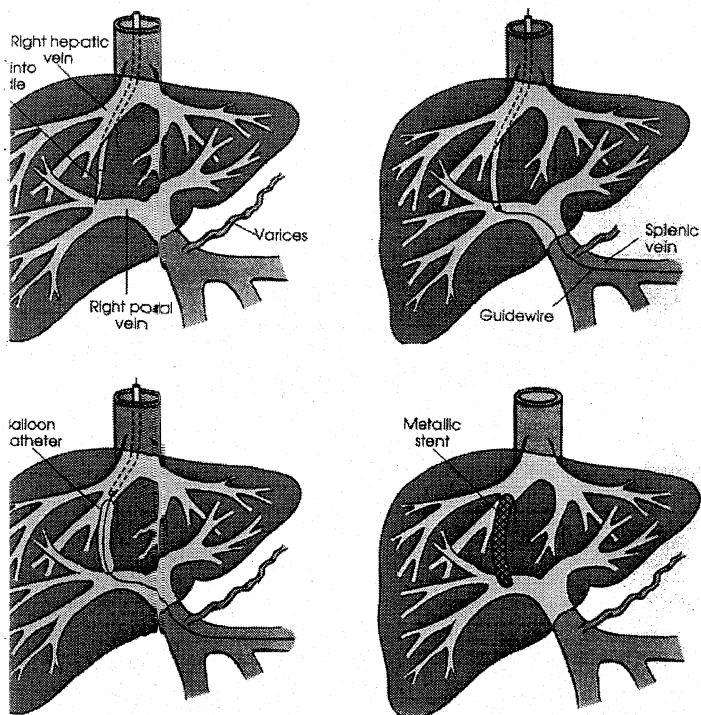
3. Percutaneous decompression

Obstructions of stomach, Intestine, Renal System, etc. can be relieved through guided procedures

4. Removal of Foreign Bodies

Metallic and non-metallic foreign bodies can be located and removed under X-Ray screening

1. Hepato Biliary interventions



The Interventional procedures done in the liver and biliary system is coming up as a subspecialty in itself.

In addition to routine biopsy techniques and abscess drainage the main emphasis is given to percutaneous removal of gallstones, biliary decompression, drainage and even endobiliary stenting in cases of obstruction to the biliary tree. Patients with obstructive jaundice are treated like this, who otherwise stands a high risk for surgery. Hepatic tumours can also be treated by 'Chemoembolisation'.

6. Genito Urinary System

Obstruction to urinary tract can also be tackled successfully by imaging guided procedures. Stones in the Kidneys, ureters and bladder also can be removed.

Now Interventional Radiology offers treatment for impotence, Pelvic pain syndromes, etc.

7. The Nervous System

In addition to the vascular interventions described earlier, imaging guided procedures like Stereotactic brain biopsy, radiotherapy and surgery are extensively being practiced.

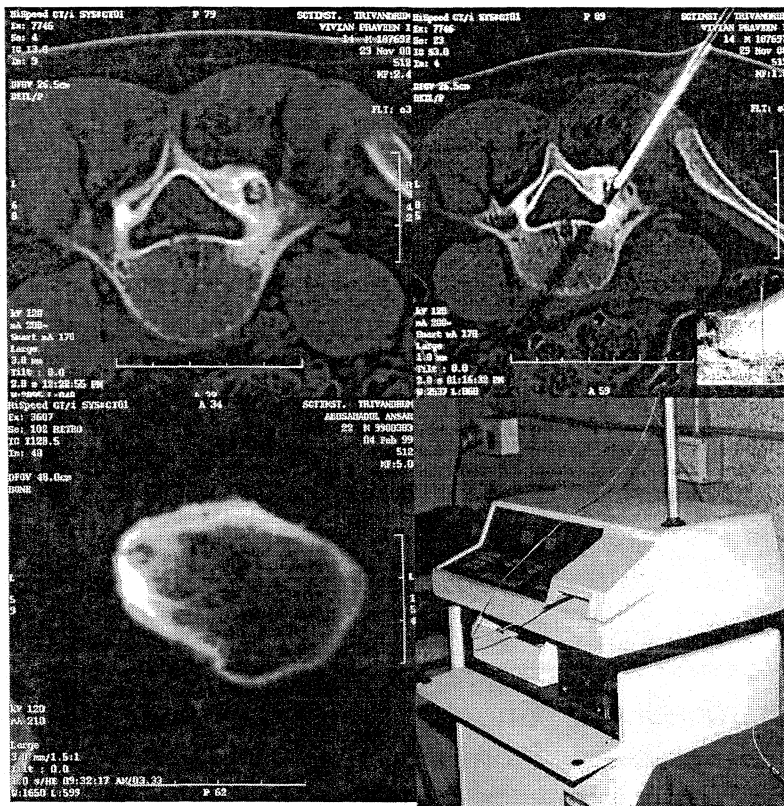
8. The Heart

The angioplasty and stenting procedures are particularly useful in coronary artery diseases. This has come up as a routine procedure and even replaced some of the indications of coronary bypass grafting.

9. Skeletal System

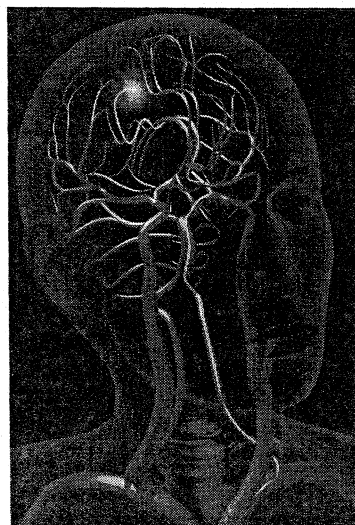
The spinal disc prolapse can be treated by percutaneous laser decompression. Other musculoskeletal applications include biopsies, bone reinforcement etc.

The list of Interventional procedures describe are innumerable are every day newer ones are added to it.



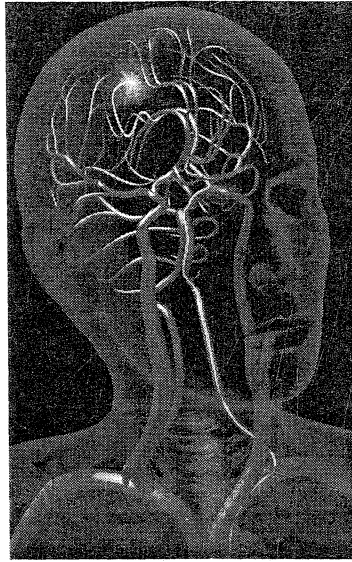
Complications:

Like any other field of invasive medicine interventional radiology is also not devoid of potential risk and side effects. The procedures are to be undertaken with extreme care and even the slightest error can lead on to devastating consequences. But with increasing experience and ever improving technology this relatively newer branch of medicine is assuming a pivotal role in patient management.



Introduction To Neuro Interventional Procedures

Introduction To Neuro Interventional Procedures



One of the newest and most promising areas of medical imaging is in the field of Neuro interventional radiology. It offers the patient an alternative to surgery and the long healing process that surgery entails. Viewing the human circulatory system as interconnected roadways, the radiologist has devised methods of entering and traveling within the system to reach the most inner recesses of the body. The utilization of these arterial passageways is at the heart of interventional radiology. Radiologist's that specialize in the brain and spinal cord can guide tiny flexible micro-catheters, metal coils, and other mini devices into minuscule blood vessels of the brain and neck.

Their goal is to correct the life threatening abnormality of the patient, which is either beyond the surgeons reach or would bleed too much if surgery were to be performed. Patient's with bizarre symptoms that arise because some parts of their brain are either starved for or overwhelmed by blood are mildly sedated or given a general anesthetic while the radiologist delicately intervene to correct the abnormality, whether it be removing various obstructions to normal blood flow or to close off abnormal openings (fistulas) or outpouchings of blood vessel walls (aneurysm) which may cause hemorrhage if it was to rupture. Because the work takes place inside

people's brains and uses devices guided from outside the body, the risk of failure or causing a second problem while trying to correct the first one is high. But if all goes well, people get their lives handed back to them.

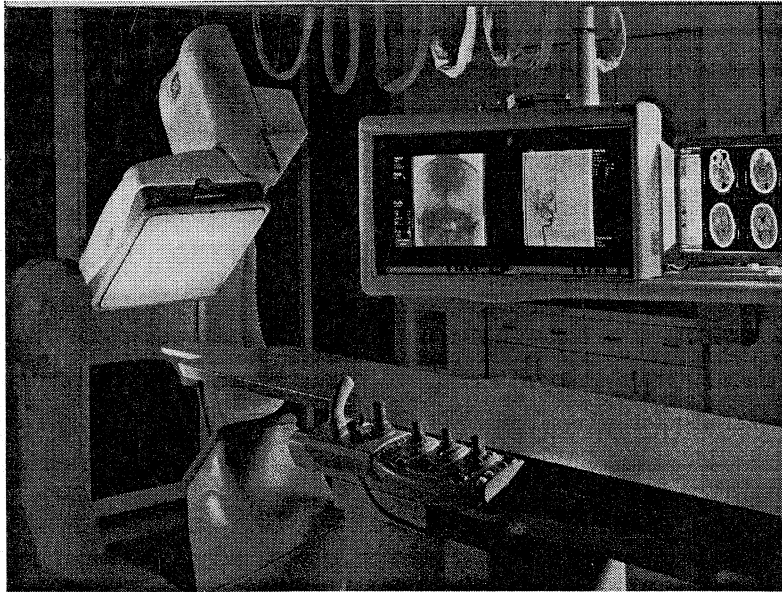
NEWER TRENDS

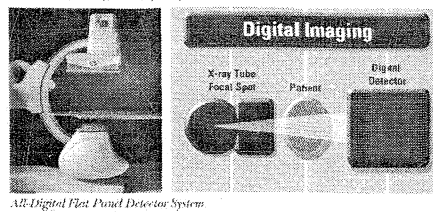
IN DSA

FLAT PANNEL DSA

DYNA CT

3D-ROTATIONAL ANGIOGRAPHY





NUCLEAR MEDICINE

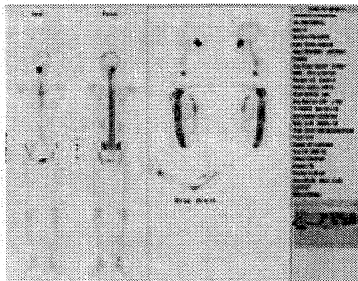
Nuclear medicine is a branch of medicine and medical imaging that uses unsealed radioactive substances in diagnosis and therapy. These substances consist of radionuclides, or pharmaceuticals that have been labeled with radionuclides (radiopharmaceuticals). In diagnosis, radioactive substances are administered to patients and the radiation emitted is measured. The majority of these diagnostic tests involve the formation of an image using a gamma camera. Imaging may also be referred to as radionuclide imaging or nuclear scintigraphy. Other diagnostic tests use probes to acquire measurements from parts of the body, or counters for the measurement of samples taken from the patient. In therapy, radionuclides are administered to treat disease or provide palliative pain relief. For example, administration of Iodine-131 is often used for the treatment of thyrotoxicosis and thyroid cancer.

Nuclear medicine imaging tests differ from most other imaging modalities in that the tests primarily show the physiological function of the system being investigated as opposed to the anatomy. In some centres, the nuclear medicine images can be superimposed on images from modalities such as CT or MRI to highlight which part of the body the radiopharmaceutical is concentrated in. This practice is often referred to as image fusion

CONTENTS

- 1 Diagnostic Testing
- 2 Types of studies
- 3 Imaging equipment
- 4 Analysis
- 5 Radiation dose

DIAGNOSTIC TESTING



☐ Bone scintigraphy of a young woman.

Diagnostic tests in nuclear medicine exploit the way that the body handles substances differently when there is disease or pathology present. The radionuclide introduced into the body is often chemically bound to a complex that acts characteristically within the body; this is commonly known as a tracer. In the presence of disease, a tracer will often be distributed around the body and/or processed differently. For example, the ligand methylene-diphosphonate (MDP) can be preferentially taken up by bone. By chemically attaching technetium-99m to MDP, radioactivity can be transported and attached to bone

for imaging. Any increased physiological function, such as due to a fracture in the bone, will usually mean increased concentration of the tracer. This often results in the appearance of a 'hot-spot' which is a focal increase in radio-accumulation, or a general increase in radio-accumulation throughout the physiological system. Some disease processes result in the exclusion of a tracer, resulting in the appearance of a 'cold-spot'. Many tracer complexes have been developed in order to image or treat many different organs, glands, and physiological processes. The types of tests can be split into two broad groups: in-vivo and in-vitro:

- *In-vivo* tests are measurements directly involving the patient. By far the most common are gamma camera imaging investigations, though non-imaging probes are also used to measure the levels of radioactivity within a patient.
- *In-vitro* tests are measurements of samples taken from the patient (e.g. blood, urine, breath).

TYPES OF STUDIES

A typical nuclear medicine study involves administration of a radionuclide into the body by injection in liquid or aggregate form, inhalation in gaseous form or, rarely, injection of a radionuclide that has undergone micro-encapsulation. Some specialist studies require the labeling of a patient's own cells with a radionuclide (leukocyte scintigraphy and red cell scintigraphy). Most diagnostic radionuclides emit gamma rays, while the cell-damaging properties of beta particles are used in therapeutic applications. Refined radionuclides for use in nuclear medicine are derived from fission or fusion processes in nuclear reactors or cyclotrons, or take advantage of natural decay processes in dedicated generators, i.e. Molybdenum/Technetium or Strontium/Rubidium.

The most commonly used liquid radionuclides are:

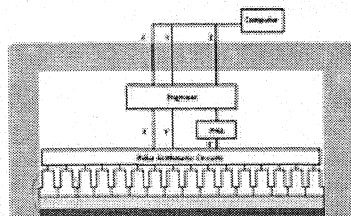
- Technetium-99m
- Iodine-123 and 131
- Thallium-201
- Gallium-67
- Fluorine-18

- Indium-111

The most commonly used gaseous/aerosol radionuclides are:

- Xenon-133

IMAGING EQUIPMENT



Diagrammatic cross section of gamma camera detector

The radiation emitted from the radionuclide inside the body is usually detected using a gamma camera. Traditionally, gamma-cameras have consisted of a gamma-ray detector, such as a single large sodium iodide $NaI(Tl)$ scintillation crystal, coupled with an imaging sub-system such as an array of photomultiplier tubes and associated electronics. Solid-state gamma-ray detectors are available[1], but are not yet commonplace. Gamma-cameras employ lead collimators to increase the image resolution by limiting the detection of unwanted gamma-rays.

Gamma-camera performance is usually a balance of spatial resolution against sensitivity. A typical gamma-camera will have a resolution of 4 to 6 mm and will be able to capture several hundred thousand gamma-ray 'events' per second. The gamma-camera detects the X and Y position of each gamma-ray event, using these coordinates to place a pixel in an image matrix to build a recognisable

image. The units of a raw nuclear medicine image is 'counts' or 'kilocounts', referring to the number of gamma-ray events detected. In nuclear medicine, the value of an image pixel is the integral of gamma-ray events in that pixel position over time. That is, the pixel appears brighter as more counts are detected in that position. In non-tomographic images, the pixel can also be thought of as the line integral of radionuclide distribution of a perpendicular line extending from the pixel position through the body of the patient.

Since each nuclear medicine radionuclide has a unique gamma-ray emission energy spectrum, and since the energy of a gamma-ray is detected in a gamma-camera by the brightness of the scintillation associated with an event, gamma-cameras employ energy 'windows' to gate or limit the imaging process to gamma-ray events of particular energies. An energy window is usually tailored to the peak of the energy spectrum of a particular radionuclide, and to ignore other gamma-rays that would otherwise contribute noise to the image. This allows noise caused by Compton scattering to be gated out.

ANALYSIS

The end result of the nuclear medicine imaging process is a "dataset" comprising one or more images. In multi-image datasets the array of images may represent a time sequence (ie. cine or movie) often called a "dynamic" dataset, a cardiac gated time sequence, or a spatial sequence where the gamma-camera is moved relative to the patient. SPECT (single photon emission computed tomography) is the process by which images acquired from a rotating gamma-camera are reconstructed to produce an image of a "slice" through the patient at a particular position. A collection of parallel slices form a slice-stack, a three-dimensional representation of the distribution of radionuclide in the patient.

The nuclear medicine computer may require millions of lines of source code to provide quantitative analysis packages for each of the specific imaging techniques available in nuclear medicine.

RADIATION DOSE

A patient undergoing a nuclear medicine procedure will receive a radiation dose. Under present international guidelines it is assumed that any radiation dose, however small, presents a risk. The radiation doses delivered to a patient in a nuclear medicine investigation present a very small risk of inducing cancer. In this respect it is similar to the risk from X-ray investigations except that the dose is delivered internally rather than externally.

The radiation dose from a nuclear medicine investigation is expressed as an effective dose with units of sieverts (usually given in millisieverts, mSv).

The effective dose resulting from an investigation is influenced by the amount of radioactivity administered in megabecquerels (MBq), the physical properties of the radiopharmaceutical used, its distribution in the body and its rate of clearance from the body.

Effective doses can range from 6 μ Sv (0.006 mSv) for a 3 MBq chromium-51 EDTA measurement of glomerular filtration rate to 37 mSv for a 150 MBq thallium-201 non-specific tumour imaging procedure. The common bone scan with 600 MBq of technetium-99m-MDP has an effective dose of 3 mSv

BASIC PRICIPLES OF ULTRASOUND

PHYSICAL PROPERTIES

Definition of Ultrasound (US):

Sound with frequency greater than 20,000 cycles per second or 20kHz. Audible sound sensed by the human ear are in the range of 20Hz to 20kHz.

Advantages:

Ultrasound can be directed as a beam.

Ultrasound obeys the laws of reflection and refraction.

Ultrasound is reflected by objects of small size.

Disadvantages:

Ultrasound propagates poorly through a gaseous medium.
The amount of ultrasound reflected depends on the acoustic mismatch.

The Four Acoustic Variables:

Pressure - the amount of force over a given area.
Distance - particle displacement with the wave
Temperature -
Density

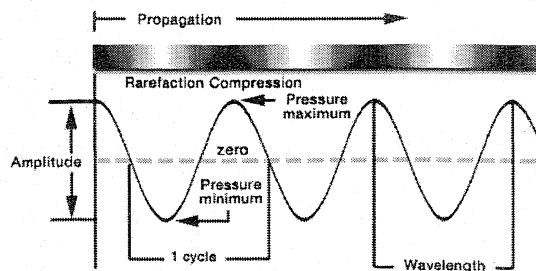
Reflection and Propagation:

Effect of propagation through gaseous zones - poor propagation, inadequate imaging.

Effect of propagation through dense zones - nearly all of the US is reflected. Structures below dense zones are poorly imaged.

Examples of dense materials - bone, calcium, metal.

Material	Speed of Propagation
bone	4080 m/s
blood	1570 m/s
tissue	1540 m/s
fat	1450 m/s
air	330 m/s



Definitions:

Cycle - the combination of one rarefaction and one compression equals one cycle.

Amplitude - the maximum displacement of a particle or pressure wave.

Intensity - the amount of force or energy of sound.

Decibel (dB) - a numerical expression of the relative loudness of sound.

Wavelength - the distance between the onset of peak compression or cycle to the next.

Velocity - the velocity is the speed at which sound waves travel through a particular medium. Velocity is equal to the frequency x wavelength.

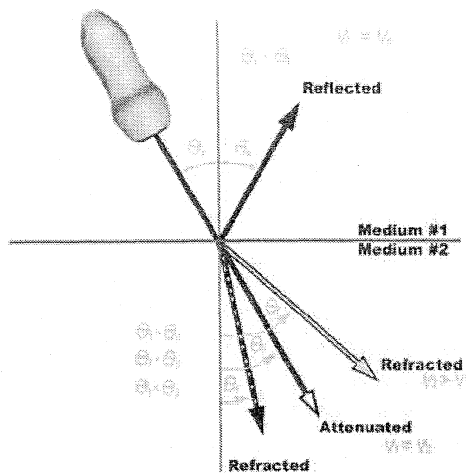
$$V = f \times \lambda$$

The velocity of US through human soft tissue is 1540 meters per second.

Frequency - the number of cycles per unit of time. Frequency and wavelength are inversely related. The higher the frequency the smaller the wavelength.

Acoustic Impedance - simply put, acoustic impedance is dependent on the density of the material in which sound is propagated through. The greater the impedance the more dense the material.

Reflection - the portion of a sound that is returned from the boundary of a medium. (echo) The angle of incidence influences the reflected and refracted waves.



Refraction - the change of sound direction on passing from one medium to another.

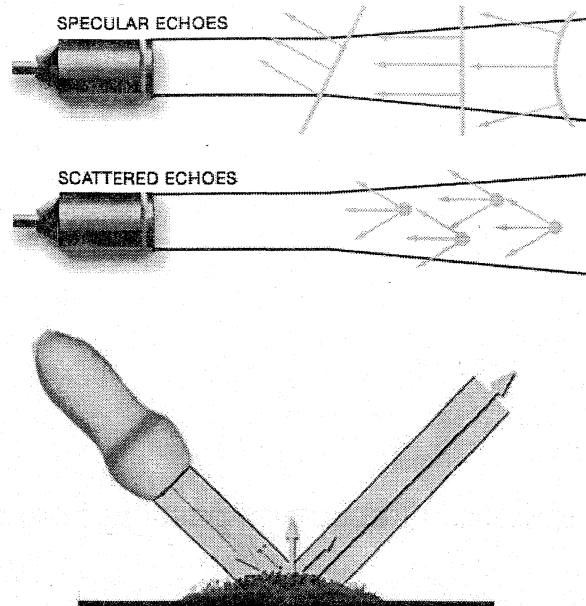
Acoustic Mismatch - the boundary between two different media where reflection and refraction occurs.

Attenuation - the decrease in amplitude and intensity as a sound wave travels through a medium.

Types of Echoes:

Specular - echoes originating from relatively large, regularly shaped objects with smooth surfaces. These echoes are relatively intense and angle dependent. (i.e. IVS, valves)

Scattered - echoes originating from relatively small, weakly reflective, irregularly shaped objects are less angle dependant and less intense. (ie. blood cells)



Scattering: Reflection and Refraction are affected by the material being imaged.

Frequencies:

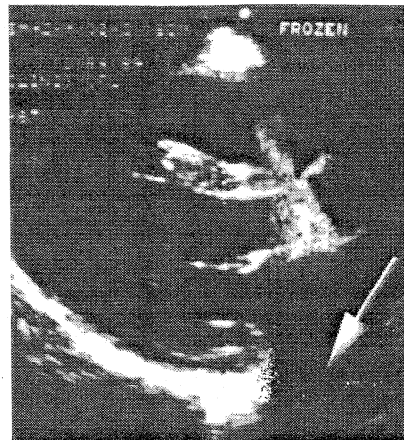
Frequencies for adult imaging - 2.0MHz to 3.0MHz.

Frequencies for pediatric imaging - 5.0MHz to 7.5MHz to 12MHz.

Effect of higher frequencies on penetration - the higher the frequency the less penetration, the lower the frequency the greater the penetration.

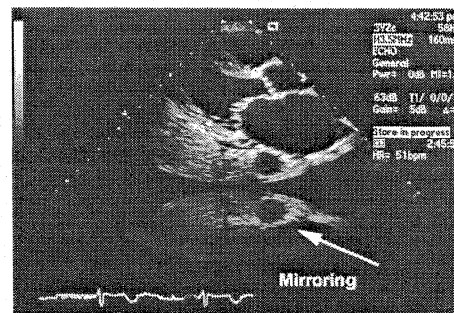
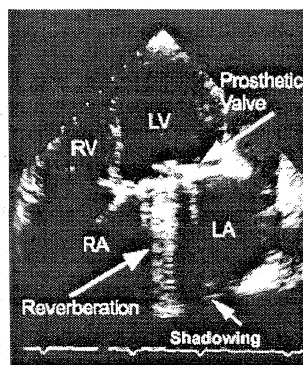
Artifacts:

Acoustic Shadowing - the loss of information below an object because the greater portion of the sound energy was reflected back by the object. This occurs in objects like prosthetic valves.



Enhancement - the increase in reflection amplitude from objects that lie behind a weakly attenuating structure. Enhancement may occur in structures below a cyst.

Reverberation - the unsuitable reflections generated when the sound wave strikes a highly reflective object creating artifacts that degrade the image. The peak of the sector scan window is usually filled with reverberations due to the initial transmission of sound energy reflecting off of the chest wall and being reflected off the transducer face in a repetitious fashion. Reverberations may occur in more internal structures like the diaphragm or from dense objects such as a mechanical valve prosthesis. **Mirroring** may occur as sound energy is reflected off dense structures and displayed on the screen as a double image.



Side-Lobe - produced from the side lobes of the ultrasound beam. This artifact appears as false structures in the scan plane.

DOPPLER PRINCIPLES

Christian Johann Doppler described the effect of motion of sound sources and its effect on the frequency of the sound to the observer. In medical applications we find that the frequency of the reflected signal is modified by the velocity and direction of blood flow. If blood cells are moving towards the transducer, they increase the frequency of the returning signal. As cells move away from the transducer, the frequency of the returning signal decreases.

The mathematical formula is:

$$\Delta F/F = 2V/C \cos \Phi \text{ or rearranged: } F_d = F_r - F_o$$

The frequency difference is equal to the reflected frequency minus the originating frequency. If the resulting frequency is higher then there is a positive Doppler shift and the object is moving toward the transducer and if the resulting frequency is lower, there is a negative Doppler shift and it is moving away from the transducer. The angle theta, cos D component is the angle of incidence of the beam upon the object. For the most accurate determination of flow, the beam should be parallel to the flow of blood where the angle theta is zero. If the angle of incidence is greater, the results are less reliable. It is generally accepted that results from the Doppler shift where the angle theta is greater than 20 degrees is not used for calculation.

Doppler Instrumentation:

Doppler techniques are dependent on the transducers used. The transducer operating in continuous wave mode utilizes one half of the element(s) and are continuously sending sound energy while the other half is continuously receiving the reflected signals.

If the transducer is being used in a pulsed wave mode, the whole transducer is used to send and then receive the returning signals.

Comparing the two modes of Doppler techniques describes the advantages and disadvantages.

	Advantages	Disadvantages
Continuous	Accurately measures	Lacks range resolution

Wave	high velocity flows	
	Ability to measure	Aliasing of velocities
	velocities at a specific	above the Nyquist limit
Pulsed wave	location (range	(inability to measure
	resolution)	high velocities
		accurately)

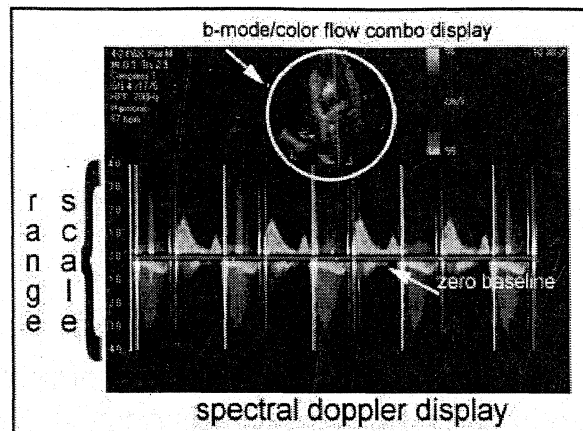
Pulsed wave techniques have proven to be very valuable in blood flow studies. The technique allows the accurate measurement of blood flow at a specific area in the heart and detection of both velocity and direction. Measurement is performed by timing the reception of the returning signals giving a view of flows at specific depths. The region where flow velocities are measured is called the sample volume. Errors in the accuracy of the information arise if the velocities exceed a certain speed. The highest velocity accurately measured is called the Nyquist limit.

Nyquist Limit - defined as $\frac{1}{2}$ the Pulse Repetition Frequency (the number of pulses per second.) If the velocity of flow exceeds the Nyquist limit, the direction and velocity are inaccurately displayed and, in fact, appear to change direction. Color flow Doppler capitalizes on this effect allowing us to detect flow disturbances from laminar to turbulent flow.

High PRF - a Doppler technique that attempts to overcome the effects of the Nyquist limit. This technique may be seen as a compromise of pulsed wave and continuous wave properties and involves the use of multiple sample volumes thereby increasing the accuracy of velocity measurements at the cost of range ambiguity.

Doppler Displays:

The display of Doppler velocity data is the Doppler frequency shifts versus time. Included in the display are the Doppler settings such as frequency, calibration, range, and timing markers.

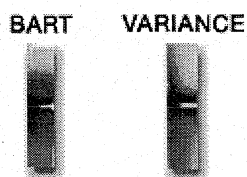


Doppler Controls:

Controls used during the Doppler examination are dependent on manufacturers specifications and the modes available. Controls for the cursor, sample volume length and depth, angle correction, gain, filters, and spectral averaging are typically included.

Color Flow Imaging:

Sampling methods - CFI is based on pulsed Doppler technology where multiple sample volumes among multiple planes are detected and displayed utilizing color mapping for direction and velocity flow data. Common mapping formats are BART (Blue Away, Red Towards) or RABT, and enhanced or variance flow maps where saturations and intensities indicate higher velocities and turbulence or acceleration, respectively. Some maps utilize a third color, green, to indicate accelerating velocities and turbulence.



Artifacts - aliasing of the data displayed in pulsed wave technology is utilized as a benefit in determining transitions from laminar to turbulent flow. Other artifacts associated with CFI and spectral Doppler are artifacts due to gain set too high, "ghosting" from improperly set wall filters (low frequency), mirroring, crying/talking artifacts, and signal loss from data sharing.

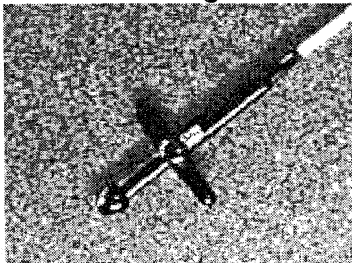
Limitations of CFI - CFI is a "qualitative" examination and has not yet been "quantified", that is, results cannot be measured to give discrete numbers for diagnosis. Qualitative assessment gives comment on the overall view or quality of the results as in flow conditions and jet direction, velocity, and pattern. Quantitative results are those measured and given discrete numbers used in calculations. There are some current semi-quantitative results given as ratios of jet length by jet width to determine the degree of regurgitation given as mild, moderate or severe.

Piezoelectric Properties:

a "crystalline" material such as quartz that changes shape when an electric current is applied creating sound waves and when struck by sound waves creates electrical currents.

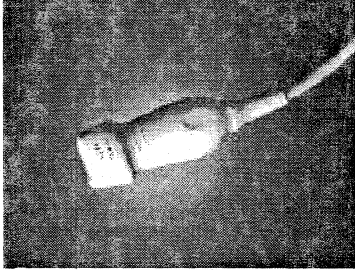
Types:

Nonimaging - transducers that are used in detection of blood flow such as a continuous wave transducer. The CW transducer, also called a Pedoff transducer, is two separate elements. One element is always transmitting while the other element is always receiving.



Imaging - transducers that are used to image cardiac structures but also have the capability of using various Doppler techniques to detect blood flow. This transducer, sometimes called a duplex transducer, is made up of multiple elements spending part of the time transmitting and part of the time receiving sound energy. Continuous wave, pulse wave, high PRF, ColorFlow, m-mode, and 2D are the various modes

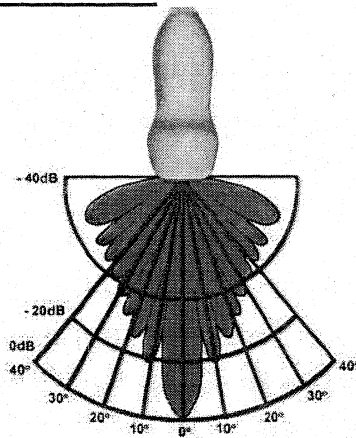
that this type of transducer can perform.



Materials:

Elements - commercial echocardiographs use ceramics such as barium titanate or lead zirconate titanate.

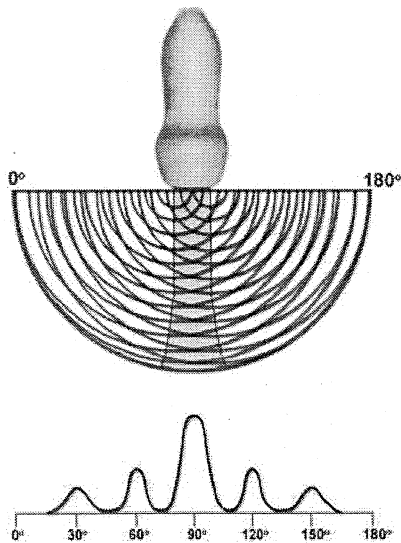
4. BEAM PROPERTIES:



Sound wave profile - the wavefront generated by the transducer is a 3 dimensional beam that contains useful and unuseable components. Longitudinal Waves - the wave in which the particle motion is parallel to the direction of the wave travel. A series of longitudinal waves make up the ultrasound beam.

Sheer Waves - waves that travel perpendicular to the direction of the sound beam. These occur mainly in the bone and are not considered in imaging soft tissue.

Side-Lobes - artifact that is generated from extraneous side beams.



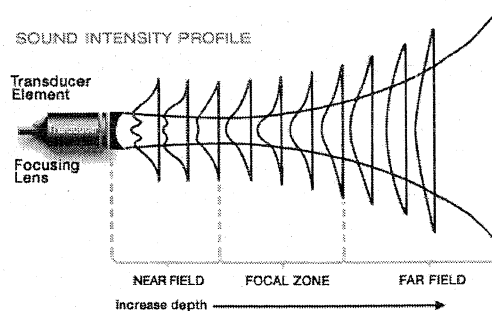
sound wave intensity variations and side lobe generation

Zones

Near - the region of a sound beam in which the beam diameter decreases as the distance from the transducer increases. This zone is called the Fresnel (Fra-nel, the s is silent) zone.

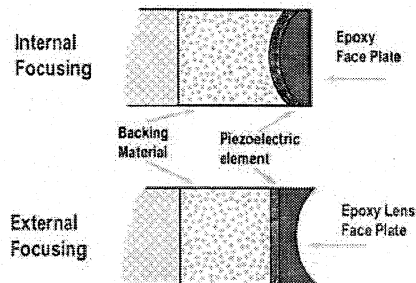
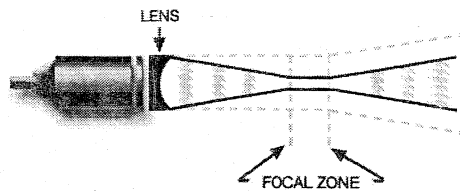
Focal - the region where the beam diameter is most concentrated giving the greatest degree of focus.

Far - the region where the beam diameter increases as the distance from the transducer increases. This zone is called the Fraunhofer zone



Focusing:

Mechanical - performed by placing an acoustic lens on the surface of the transducer or using a transducer with a concave face.



External Focusing - a lens is used on the transducer face to bend the soundwaves to a focal point.

Internal Focusing - the piezoelectric element is formed in a curved element that focus the beam.

Electronic - using a process called phased array, multiple elements are fired sequentially to focus the beam .

Beam Focusing

Array transducers have the ability to be dynamically focused by stimulating each element as shown in the diagram below. The individual wave fronts add up to a tight beam of ultrasound energy.

Beam Steering:

Array transducers have the ability to be steered as well as focused. Like focusing, the beam is directed by sequentially stimulating each element as shown. This feature creates the sector scan by rapidly steering the beam from left to right to give the two dimensional cross sectional image.

Resolution:

Lateral resolution - the ability to resolve objects side by side. Lateral resolution is proportionally affected by the frequency, the higher the frequency the greater the lateral resolution. Higher frequency transducers are used in fetal and pediatric echocardiography

because the lateral resolution displays the smaller structures in those patients and there is less need for depth penetration. Lower frequencies are used for adults where structures are larger and the need for greater depth penetration is important.

Axial Resolution - Axial resolution is the ability to resolve objects that lie one above the other. Axial resolution is inversely proportional to the frequency of the transducer depending on the size of the patient. The higher the frequency the lower the axial resolution is in large patients. This state results from the rapid absorption of the ultrasound energy with lower penetration. Lower frequencies are utilized to increase depth of penetration.

Depth of Penetration - Higher frequencies are attenuated by tissue more than lower frequencies. This means that the higher the frequency the greater the resolution but the lower the depth of penetration. User lower frequencies for adults and higher frequencies for children. The advent of harmonic imaging allows the use of a lower frequency pulse to be picked up and sampled at a higher frequency (the second harmonic) where the low frequency allows greater penetration and high frequency provides better resolution.

	2.0mHz	5.0mHz
Axial	Decrease	Increase
Lateral	Decrease	Increase
Penetration	Increase	Decrease

Basic Components of the Imaging System:

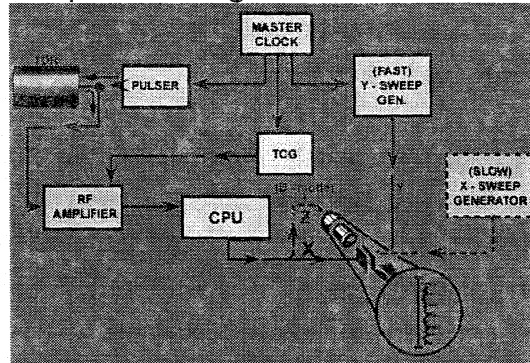
Transducer - the probe housing the elements, backing material, electrodes, matching layer and protective face that both sends and receives the sound waves.

Transmitter - the component that creates the impulses sent to the transducer to generate sound energy. Also called the pulser.

Receiver - the component that receives the current generated in the transducer from the returning sound energy.

Amplifier - the component that amplifies the returning signals and prepares them to be displayed on the CRT.

CPU - the central processing unit that



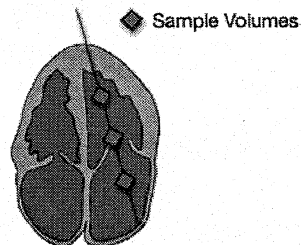
Pulsing:

Pulse Repetition Frequency (PRF) - the number of pulses per second.

Duty Factor - the amount of time that the transducer is transmitting sound energy. Most transducers are acting as the receiver 99 % of the time. the duty factor is 1.

Medical Application PRFs -

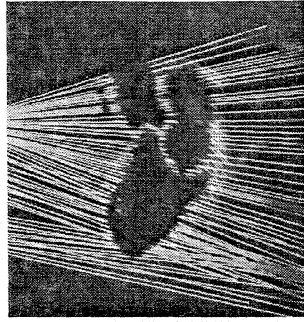
High PRF - the range for most commercial echocardiographs is between 200 and 5000 pulses per second. The varies with the type of mode in operation.



Modes:

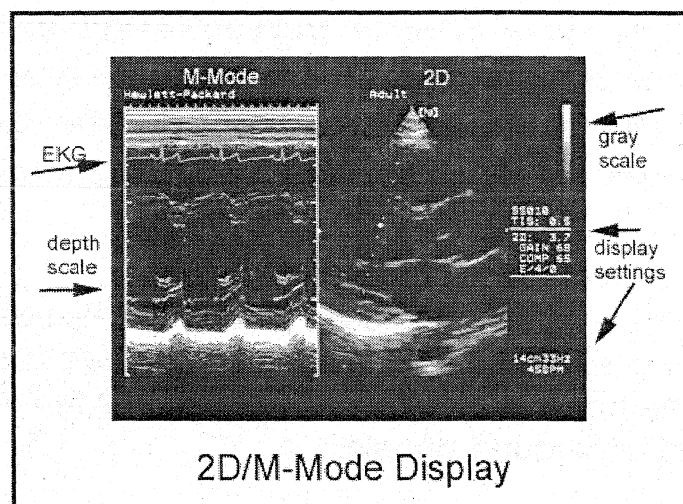
A Mode - amplitude mode. Where the signals are displayed as spikes that are dependent on the amplitude of the returning sound energy.

B Mode - brightness mode. Where the signals are displayed as various points whose brightness depends on the amplitude of the returning sound energy.

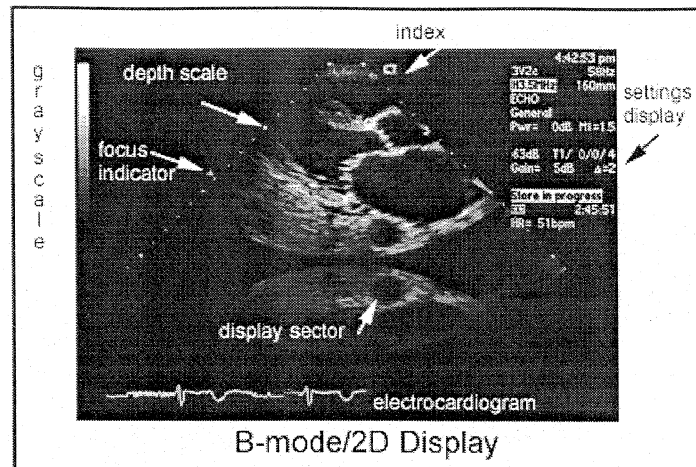


Old format B-mode type scan

M Mode - motion mode. The application of B-mode and a strip chart recorder allows visualization of the structures as a function of depth and time.



2D Mode - 2 dimensional mode name is usually reserved to indicate b-mode imaging of the heart. The spatially oriented B-mode where structures are seen as a function of depth and width. The beam is rapidly swept back and forth to create a cross section of the imaged structures.



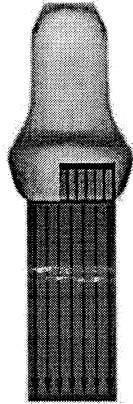
Scan Types:

Mechanical - transducers that use a combination of single element oscillation back and forth, rotating multiple elements, or a single element and set of acoustic mirrors to generate the sweeping beam for 2D imaging. This transducer type is sometimes called the "wobbler" because of the vibration created as the mirrors rotate or oscillate inside the housing.

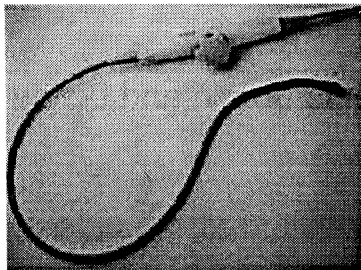
Electronic or Phased

Sector - creates a sector or pie shaped scan plane. This is most useful for cardiac work where the beam is directed between the ribs to image the heart. The beam is electronically swept in a fan shape to transmit and receive ultrasound signals.

Linear - creates a linear or rectangular shaped scan plane. This is most useful in abdominal, OB/GYN, and small parts work where the organs are not blocked by bones or ribs. A wide footprint allows the operator to image larger organs. This method is simplified because there are no obstructions to the signal by ribs and other bones.



Transesophageal -



The piezoelectric device has been placed at the end of an endoscope so that the transducer may be passed into the esophagus to image the heart from inside the body.

Advantages of Electronic over Mechanical - dynamic focusing, smaller housing.

Disadvantages of Electronic over Mechanical - more expensive.

SIGNAL PROCESSING

Terms:

Decibel (dB) - a unit describing the ratio of logarithmic power amplification of the signal.

Dynamic range - the range between the minimum low intensity and maximum high intensity signals that a system is capable of displaying.

Reject - The control that allows the operator to ignore the weak

echoes that may clutter the display and obscure the higher amplitude signals. It filters out all signals below a fixed amplitude to improve the recording

TGC - Time Gain Compensation. The control that allows the operator to amplify the returning signal from deeper structures in the body. The sector is divided into depths and altering the TGC control can dampen or amplify the returning signals

Damping - a method of reducing the pulse duration, or ringing of the transducer, and thereby increasing lateral resolution.

Compress - a control on some systems that affects the gray scale and overall gain.

Axial Resolution - the resolution of objects above and below each other.

Lateral Resolution - the resolution of objects that are side by side.

Side-Lobe - produced from the side lobes of the ultrasound beam. This artifact appears as false structures in the scan plane.

Reverberation - produced from the multiple reflections from an object as the sound energy bounces back and forth between the object and the transducers face or dense structure.

Shadowing - the loss of data below a dense object because the majority of the sound energy has been reflected. This occurs typically in prosthetic valves.

Biologic Effects and Safety:

Two bioeffects of ultrasound are heat and cavitation. As sound energy is transmitted through tissue, some energy is reflected and some energy is absorbed as heat. It has not been determined that medical ultrasound causes any deleterious effects. Cavitation is the formation, growth, and dynamic behavior of gas bubbles neither which have been determined harmful at the level of the medical usage. Mechanical effects of ultrasound include shear forces and

acoustic streaming.

AIUM Statements:

The American Institute of Ultrasound in Medicine states that as of 1982, no independently confirmed significant biologic effects had been observed in mammalian tissue below (medical usage) 100mW/cm².

Measures of Intensity for Ultrasound:

Spatial Peak, Temporal Peak Intensity (SPTP) - the peak intensity at the point in space where the intensity is highest and that occurs when the ultrasonic emitting device is "on". It is highest of the measured intensities.

Spatial Average, Temporal Average Intensity (SATA) - the average power output of the device over the pulse repetition period divided by a reference, usually that of the transducer face. This measurement of intensity is the most quickly determined and most frequently quoted by manufacturers. It also yields the lowest more commonly used measures of intensity.

Spatial Average, Temporal Peak Intensity (SATP) - the peak intensity over a selected area, such as the transducer face, that occurs when the ultrasonic emitting device is "on."

Spatial Peak, Temporal Average Intensity (SPTA) - the maximal spatial intensity when the sound beam is "on" average over the pulse repetition period.

Spatial Peak Pulse Average Intensity (SPPA) - the pulse-averaged intensity measured at the point in space where the value is maximal.

Spatial Average Pulse Average (SAPA) - the pulse average intensity averaged over the beam cross sectional area.

PICTURE ARCHIVING AND COMMUNICATION SYSTEM

In medical imaging, **picture archiving and communication systems** (PACS) are computers or networks dedicated to the storage, retrieval, distribution and presentation of images

TYPES OF PACS

Full PACS handle images from various modalities, such as ultrasonography, magnetic resonance imaging, positron emission tomography, computed tomography, endoscopy, mammography and radiography (plain X-rays).

Small-scale systems that handle images from a single modality (usually connected to a single acquisition device) are sometimes called *mini-PACS*

USES

PACS replaces hard-copy based means of managing medical images, such as film archives. It expands on the possibilities of such conventional systems by providing capabilities of off-site viewing and reporting (distance education, tele-diagnosis). Additionally, it enables practitioners at various physical locations to peruse the same information simultaneously, (teleradiology). With the decreasing price of digital storage, PACS systems provide a growing cost and space advantage over film archives.

PACS is offered by virtually all the major medical imaging equipment manufacturers. There are also several PACS open-source projects. The most difficult area for PACS systems is interpreting the DICOM image format which is slightly variably implemented between different radiology equipment vendors. The ability to point the tags in the DICOM format coming from vendors equipment to useable titles in a PACS is a feature common to most vendors and software offerings

ARCHITECTURE

Typically a PACS network consists of a central server which stores a database containing the images. This server is connected to one or more clients via a LAN or a WAN which provide or utilize the images. Web-based PACS is becoming more and more common: these

systems utilize the Internet as their means of communication, usually via VPN or SSL. The software (thin or smart client) is loaded via ActiveX, Java, or .NET Framework. Definitions vary, but most claim that for a system to be truly web based, each individual image should have its own URL. Client workstations can use local peripherals for scanning image films into the system, printing image films from the system and interactive display of digital images. PACS workstations offer means of manipulating the images (crop, rotate, zoom, brightness, contrast and others).

Modern radiology equipment, modalities, feed patient images directly to the PACS in digital form. For backwards compatibility, most hospital imaging departments and radiology practices employ a film digitizer.

The medical images are stored in an independent format. The most common format for image storage is DICOM (Digital Imaging and Communications in Medicine), a NEMA standard

INTEGRATION

A full PACS system should provide a single point of access for images and their associated data (i.e. it should support multiple modalities). It should also interface with existing hospital information systems: hospital information system (HIS) and radiology information system (RIS).

Interfacing between multiple systems provides a more consistent and more reliable dataset:

- Less risk of entering an incorrect patient ID for a study – modalities that support DICOM worklists can retrieve identifying patient information (patient name, patient number, accession number) for upcoming cases and present that to the technologist, preventing data entry errors during acquisition. Once the acquisition is complete, the PACS can compare the embedded image data with a list of scheduled studies from RIS, and can flag a warning if the image data does not match a scheduled study.

- Data saved in the PACS can be tagged with unique patient identifiers (such as a social security number or NHS number) obtained from HIS. Providing a robust method of merging datasets from multiple hospitals, even where the different centers use different ID systems internally.

An interface can also improve workflow patterns:

- When a study has been reported by a radiologist the PACS can mark it as read. This avoids needless double-reading. The report can be attached to the images and be viewable via a single interface.
- Improved use of online storage and nearline storage in the image archive. The PACS can obtain lists of appointments and admissions in advance, allowing images to be pre-fetched from nearline storage (for example, tape libraries or DVD jukeboxes) onto online disk storage (RAID array)

Recognition of the importance of integration has led a number of suppliers to develop fully integrated RIS/PACS systems. These may offer a number of advanced features:

- Dictation of reports can be integrated into a single system. The recording is automatically sent to a transcriptionist's workstation for typing, but it can also be made available for access by physicians, avoiding typing delays for urgent results, or retained in case of typing error.
- Provides a single tool for quality control and audit purposes. Rejected images can be tagged, allowing later analysis (as may be required under radiation protection legislation). Workloads and turn-around time can be reported automatically for management purposes.

COMPUTED RADIOGRAPHY

The last few decades have seen numerous technological advancements in the field of radiology with the introduction of newer technologies such as Sonography, Color Doppler, CT, MRI and DSA. In fact this has prompted the change of the nomenclature of this speciality from radiology to medical imaging. Alas the modality that invented this specialty, conventional radiology or more fondly known as plain films has been bereft of any significant technological advancement. The other modalities have significantly benefited because of their reliance on computer technologies. As the hard ware and soft ware revolution has progressed globally in all fields, this has naturally benefited these modalities but not conventional radiology, which has practically no reliance on computer technologies. The key to advancements in conventional radiology is in digitizing the images so they can be manipulated in an electronic format and thus enhanced.

There are two means available to obtain Digital X-Rays

1. Digital Radiography
2. Computed Radiography

Digital Radiography: The X-Ray machines are digital machines with flat panel detectors. There are no available portable digital X-Ray machines. So portable X-rays would still remain conventional.

Computed Radiography: CR uses standard X-Ray machines. There is no need to change existing X-Ray machines as is required in Digital Radiology. There is only a change in the recording device i.e. the cassette. In CR rather than a film, the image is exposed on a digital plate. The Digital image is then transferred to a reader, where the image is displayed on a monitor. This image being digital can be modified to adjust the exposure. Once the quality is approved the image may be printed on film in a laser camera. The images may also be stored in an electronic format on CD or sent to a remote location in the hospital via a local area network or even emailed to any location

in the city, country or world. The images may also be transferred to a workstation similar to a CT or MRI workstation. If the department has existing CT and MRI workstations these can be used without the need to obtain a new workstation. There are numerous advantages of a workstation, the images can be modified, cropped, magnified and labeled

There are numerous advantages of obtaining Digital images for conventional Radiology

1. Marked improvement in image quality

Chest, Spine and bone images are of exceptional quality, with better trabecular details.

2. No need for retakes.

Conventional X-Rays are still a manual procedure; the technician sets the exposure depending on the size of the patient. This could occasionally be faulty resulting in an over or under exposure. When this occurs it will need to be corrected by repeating the X-Ray after readjusting the exposure factors. This results in additional radiation to the patient as well as loss to the institution, as the first film will need to be discarded.

With Digital X-Rays the exposure factors do not need to be so accurate, variations in exposure can be adjusted on the monitor after the image has been obtained. Thus a perfect image can then be printed.

Spoil film rates range from 10-15%, similar to most top class institution obtaining X-Rays using conventional means. Conversion to digital will result in a saving of 10-15% on the film bill.

3. No loss of Films

Occasionally in very busy departments films may get misplaced. These films need to be repeated, again resulting in additional radiation exposure and loss of revenue.

4. Multiple images from one exposure

In post bypass patients the physician would like soft exposure X-Rays