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Section 1.MRI Safety

The MRI system is made up of several components all of which have the potential to cause harm to both the patient and the operator.

1.1 Static Magnetic Field

The magnet is always on and as such the static magnetic field will always be present inside the magnet room. There is some evidence of mild sensory effects due to static magnetic fields including

- Vertigo
- Nausea
- Taste Sensations

This seems to be related to the strength of the magnetic field with few symptoms at 1.5T and more appearing approaching a 4T field strength. It should be remembered that exposure of humans to the Static Field is for short periods only and that these effects cease with the exposure. (Westbrook, C. & Kaut, C. 1998)

Flow Affects

Any charged particle moving through a magnetic field will have a current induced via this interaction. As blood, a conductive fluid flows through a magnetic field it induces an electrical biopotential. This electrical biopotential can be seen on the ECG waveform of patients in MRI. It is demonstrated by an increase in the T-wave amplitude on ECG. This change in T-wave amplitude is directly proportional to the strength of the Static Field. It is important that MR radiographers are aware of this as elevated T-waves are also associated with ischemia and myocardial infarction. (Mcrobbie et. al. 2003)

Force Fields

Static Fields also pose hazards through the displacement of ferro-magnetic implants and objects. The static field imparts both a translation (attractive force) and a torque (twisting force). The strong nature of the Static Field also has the potential to disrupt the function of Cardiac Pacemakers. Anyone entering beyond the 5G line should be screened for contraindications to strong Magnetic Fields. Also be aware of what is being taken over the 5G line.

- Ferromagnetic objects may become airborne - 40 km/hr terminal velocity possible at 1.5T
- Test any metal objects with a hand held magnet before allowing them to enter the room.
- Keep the general public behind the 5G line.
- Possible contra-indications include pacemakers, aneurysm clips, intra-vascular coils and stents, heart valves, penile implants, cochlear implants, ocular implants (Fatio eyelid springs), neuro stimulators, bone growth stimulators, drug infusion pumps.


1.2 Gradient Magnetic Fields

Faraday’s Law dictates that changing magnetic fields will generate a voltage and a current in a conductor. The switching of the gradients induces electrical currents in conductive tissues. Voltage generation will occur during the rise and fall times - i.e. while the field is changing. The largest voltages will be generated at the periphery where the gradient amplitudes are highest. The most common result of these voltages is Peripheral Nerve Stimulation (PNS). Other side-effects are listed below; (Mcrobbie et. al. 2003)
Possible effects of induced voltages include:

- magnetophosphenes - electromagnetically induced flashes of light
- seizures
- tissue heating
- peripheral nerve stimulation
- muscle contractions
- cardiac arrhythmias

(Lennon-George, J)

In order to keep these effects to a minimum the following is recommended.

- keep dB/dT less than those required to produce peripheral nerve stimulation.
- max limit for dB/dT of 6 Tesla / sec

Some degree of patient heating has been shown to be a result of Gradient Magnetic effects. However this has been shown to be so low as to be negligible. (Lennon-George, J)

Gradient Noise

The characteristic noise that is heard during an MRI scan is the result of the interaction of Lorentz forces generated by the Gradient Coil when a current is pulsed through them in the presence of a Static Magnetic Field. The noise results as these forces are so strong as to twist the coils on their mounts. In some cases this can be in excess of 100dB particularly when running EPI sequences. Thus all patients need to be provided hearing protection. (McRobbie et. al. 2003)

1.3 Radiofrequency Fields

RF effects are of the greatest concern in terms of MR safety. The main effect is the deposition of energy resulting in tissue heating. Of particular concern are heat sensitive organs such as the eyes and testes. One of the major concerns is where a patient has a metallic implant which results in greater heating.

The amount of energy absorbed will increase with frequency and is therefore greater at higher field strengths. (Woodward, Peggy, 2001 & McRobbie et. al. 2003)

SAR= Specific Absorption Rate - the term used to describe energy dissipation.

SAR is expressed in Watts / kg and will depend on factors including:

- frequency (& therefore the field strength)
- RF pulse - where power deposited is proportional to (flip angle / 90)^2
- TR
- RF coil - transmit / receive or receive only
- volume of tissue within the coil
- conductivity of the tissue (Lennon-George, J)

The SAR is calculated for each sequence on the basis of the sequence parameters and the patient weight.

FDA Recommendations:

- Maximum SAR of 0.4 W/kg (whole body), 3.2 W/kg (head), 8.0 W/kg (in any one gram of tissue).
- RF exposure should be insufficient to produce a core temperature increase of 1 degree C.

- Heating will also vary with the state of the patient’s thermo-regulatory system, the ambient temperature, the humidity and the air flow around the patient.
- The eye and the testes are particularly sensitive organs due to a low capacity for heat dissipation.
- Excessive RF exposure and temperature increase can lead to an increase in blood pressure and heart rate. (Lennon-George, J)
Factors which will reduce SAR if required include:

- reducing the number of slices
- reducing the ETL
- increasing the TR
- GE rather than SE
- Use quadrature coils rather than linear coils for transmission
- Change to low SAR sequence design

(Lennon-George, J & Mcrobbie et. al. 2003)

Patient Burns and the RF Antenna Effect

The use of inappropriate physiological monitoring systems, both the leads and their connectors (ECG dots) can result in heating of these electrodes and a burn to the patient. Inappropriate positioning of these leads as well as surface coil cables used in MR can cause a conductive loop to form and severe heating will result. (Kaut Roth, Carolyn, 2002 & Lennon-George, J)

Recommendations:

- use only equipment tested to be MR compatible
- allow only MR trained staff to use the equipment
- check the integrity of the electrical insulation of all cables
- remove all unnecessary electrically conductive equipment from the bore
- keep electrically conductive equipment from directly contacting the patient
- keep electrically conductive equipment from forming large diameter conductive loops
- position cables to avoid cross points
- position ECG cables to exit down the centre of the bore
- do not allow contact between the patient’s skin and the magnet bore

(Lennon-George, J)

1.4 Pregnancy and MRI

There are no known biological effects of MRI on foetuses.

“MR imaging may be used in pregnant women if other non-ionizing forms of diagnostic imaging are inadequate or if the examination provides important information that would otherwise require exposure to ionizing radiation (e.g., fluoroscopy, CT, etc.). Pregnant patients should be informed that, to date, there has been no indication that the use of clinical MR imaging during pregnancy has produced deleterious effects.” This policy has been adopted by the American College of Radiology and is considered to be the “standard of care” with respect to the use of MRI procedures in pregnant patients. Importantly, this information applies to MR systems operating up to and including 3-Tesla. (MR Imaging Safety and Patient Management issued by the Safety Committee of the Society for Magnetic Resonance Imaging)

Pregnant patients should be reviewed on a case by case basis to assess whether the risks outweigh the potential benefits. (Lennon-George, J)

Pregnant Staff

No increased incidence of spontaneous abortion has been demonstrated among MR radiographers or nurses. The most common policy seems to be that pregnant staff are allowed exposure to the static field (ie. to set patients up) but leave the room during image acquisition to avoid exposure to RF and gradient fields. (Lennon-George, J)

Contrast Media

Contrast is not recommended during pregnancy. It has been shown to cross the placenta. (Lennon-George, J)
1.5 Quenching

Quenching the magnet results in a large amount of helium gas boil off from liquid helium inside the magnet. One litre of liquid helium expands to 760 litres of helium gas. During a quench, the liquid helium is heated causing it to boil off. The quench pipe vents the resultant gas outside. If the vapour pressure exceeds a predetermined value, the vent-bursting disk on the vent pipe ruptures and ventilates the excess vapour. Some may enter the scan room reducing oxygen levels. Helium gas displaces oxygen and is hard to detect until it is too late. Asphyxia is possible due to low oxygen levels. Another concern is the increase in pressure due to helium flooding the scan room. It may become impossible to open an inward opening door due to the pressure differential. In this case kick out panels should be used to reduce the pressure differential and allow opening of the door. In the absence of these panels the scan room window should be broken to allow access to the patient if necessary. (Westbrook, C. & Kaut, C. 1998)
Section 2. MRI Basic Physical Principles

2.1 Hydrogen Atom

MRI is based on the natural magnetisation that is induced in the human body when it is placed inside a strong magnetic field. The hydrogen proton forms the basis of MRI. MRI takes advantage of the fact that the hydrogen atom contains a single proton (+ve charge) and as such exhibits a strong magnetic moment. Hydrogen is the most abundant proton in the human body and it is because of these two factors that we utilise it for MRI imaging. (Fig. 1) (Westbrook, C. & Kaut, C. 1998)

![Fig. 1 The hydrogen nucleus is in reality just a simple bar magnet](image1)

2.2 Alignment

When placed inside a magnetic field the hydrogen protons within the human body will attempt to align with that field. In MRI we refer to the main magnetic field as $B_0$. The hydrogen proton due to the presence of its single positively charged proton can align in only one of two directions. That is parallel to the main magnetic field $B_0$ or antiparallel to the field. A hydrogen proton will align parallel or antiparallel dependent upon the relative energy of the individual proton and also the strength of the main magnetic field. Protons with lower energy align parallel and those with higher energy align antiparallel. These two states are also referred to as Spin up and Spin Down respectively. (Fig. 2) (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

![Fig 2. Two possible orientations for the proton in an externally applied magnetic field](image2)

The stronger the field the fewer protons are able to align antiparallel. This effect of alignment results in a Net Magnetisation Vector. (Westbrook, C. & Kaut, C. 1998)

2.3 Net Magnetisation Vector (NMV)

A good way to think about the NMV is to take the example of a single voxel of tissue. A voxel contains many protons with the net magnetisation vector ($M$) being the vector sum of the individual protons. In the absence of a magnetic field, the spatial orientation of each proton's magnetic moment is random and $M = 0$.

This situation is changed in the presence of a stationary magnetic field ($B_o$) which induces some of the magnetic moments of the protons to align in its direction, partially overcoming thermal randomisation and producing a net magnetisation in the direction of $B_o$. (Fig. 3)
At 1.5 Tesla, the magnitude of this magnetisation corresponds to about $1 \times 10^5$ of the dipoles aligned with the field, with the rest randomly orientated. The fraction is small but the total number of contributing protons is large with about $10^{15}$ protons in a $5\text{mm}^3$ volume.

- The presence of $B_0$ creates an abundance of protons in alignment with this field.
- It is this abundance of protons which forms the NMV


### 2.4 Precession

Each hydrogen nucleus that makes up the NMV is spinning on its own axis as shown in the diagram (Fig. 4). The influence of the static magnetic field $B_0$ creates an additional spin of the NMV around $B_0$. This secondary spin is called precession and causes the magnetic moments to follow a circular path around $B_0$.

The speed at which the NMV rotates around $B_0$ is called the precessional frequency. It is measured in megahertz (MHZ). Where one cycle per second is 1Hz. (Westbrook, C. & Kaut, C. 1998)
The Larmor Equation

The value of the precessional frequency is governed by the Larmor equation. The equation states that:

The precessional frequency \( (\omega_0) = B_0 \times y \)

Where by \( B_0 \) is the magnetic field strength of the magnet, and \( y \) is the gyromagnetic ratio.

The gyromagnetic ratio of hydrogen is 42.57 MHz/T at 1.0T field strength. The gyromagnetic ratio is a constant for each type of nucleus however field strengths are variable and as such the precessional frequency for each nucleus will be different dependent on the strength of the magnet.

For example:
- At 1.5T \( \omega_0 \) of hydrogen is 63.86MHz
- At 0.5T \( \omega_0 \) of hydrogen is 21.28MHZ

The precessional frequency is often called the Larmor frequency so be aware these titles are interchangeable. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

2.5 Resonance

Resonance is a phenomenon that occurs when an object is exposed to an oscillating perturbation that has a frequency close to its own natural frequency of oscillation. In MR a hydrogen nucleus will gain energy and resonate if the energy is delivered at exactly its precessional frequency. If the energy is delivered to the nucleus at a different frequency to that of the Larmor frequency of the nucleus resonance will not occur.

The result of resonance is that the NMV moves out of alignment from \( B_0 \). The angle to which the NMV moves out of alignment is known as the flip angle. The magnitude of the flip angle depends upon the amplitude and duration of the RF pulse. Typically the flip angle is 90 degrees.

With a flip angle of 90 degrees the result is that the NMV moves away from \( B_0 \) and is completely transferred into the transverse plane \( B_1 \). The transverse plane is known as \( B_1 \). The NMV now precesses at the Larmor frequency in the transverse plane.

The second result of resonance is that the magnetic moments of the hydrogen protons move into phase with each other. That is all of the protons are in the same position of the precessional path. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

2.6 Signal Creation and Detection

According to Faraday’s law of induction, a changing magnetic field will generate a voltage in a suitably located receiver coil. This is the basis of signal detection in MRI.

For a signal to be detected, the magnetisation induced by \( B_z \) in the direction of \( B_0 \) needs to be converted into a magnetisation which precesses in the X-Y plane. This is done by applying a RF pulse (\( B_1 \)) that causes the NMV to flip and precess in the transverse plane. NMV will precess about \( B_1 \) which can be pulsed for as long as it takes to rotate the magnetisation through 90 degrees into the transverse plane. Individual protons will now be precessing in phase to produce a rotating magnetisation which can be detected with a receiver coil. (Westbrook, C. & Kaut, C. 1998)
2.7 Relaxation of the NMV

When a scanning sequence starts, the magnetisation in the longitudinal plane is flipped into the transverse plane using an RF pulse. This creates a measurable signal in the receiver coil. Over time this signal fades away or decays. After the RF pulse the NMV moves back towards the longitudinal plane via two relaxation processes.

- Spin-Lattice Relaxation, **T1 Relaxation**
- Spin-Spin Relaxation, **T2 Relaxation or T2 decay**

(Westbrook, C. & Kaut, C. 1998)

**T1 relaxation**

T1 relaxation is commonly referred to as spin-lattice relaxation as the interactions that are involved in this relaxation mechanism are between the protons (or spins) and their environment. This type of relaxation deals with the longitudinal component of the magnetisation and T1 is a measure of the time in which it takes the longitudinal component of magnetisation to reach 63% of its initial value. (Fig. 5) (Westbrook, C. & Kaut, C. 1998)

T1 relaxation therefore is the regrowth of the magnetisation along the longitudinal axis. Previously we discussed the concept that it requires energy to move from one alignment to another. Thus as spins realign themselves with the longitudinal axis, they must give up some energy. This energy is given up to the environment in which the spins reside to other nearby electrons or nuclei. (Westbrook, C. & Kaut, C. 1998)

An important equation governs T1 relaxation and Net Magnetisation. It is often the subject of examinations. Basically it states the following.

Net magnetisation is not achieved instantaneously. From the time the field is applied, M grows from zero towards its equilibrium value (M₀) along the Z axis in an exponential fashion such that

\[ M_z = M_0(1 - \exp(-t/T_1)) \]

where T1 is the longitudinal relaxation time. After one T1 interval, 63% of the total magnetisation will have occurred. (Lennon-George, J)

![Fig 5. The T1 relaxation curve](image)
T2 Relaxation or T2 Decay

T2 decay is caused by protons exchanging energy with neighbouring protons. The energy exchange is caused by the interaction between the magnetic fields of adjacent protons. It is known as spin spin relaxation due to the interaction between the spins or protons and results in the decay or loss of transverse magnetisation. This rate of decay is an exponential process and the T2 time of a tissue is the time it takes for 63% of the transverse magnetisation to be lost. Often the T2 time is also described as the time it takes for 37% of transverse magnetisation to remain. (Fig. 6) (Westbrook, C. & Kaut, C. 1998)

This irreversible signal loss due to interactions between neighbouring nuclei can be modelled by an exponential decay with a time constant T2 such that

$$|M_{xy}| = M_0 \exp\left(-\frac{t}{T2}\right)$$

where T2 is the transverse relaxation time. After one T2 time interval, 63% of the transverse magnetisation will be lost. (Lennon-George, J)

This decay is characterised by an exponential with a time constant T2* and encompasses dephasing due to both imperfections in the applied magnetic field as well as due to interactions between neighbouring nuclei. The rapidly diminishing signal detected at the receiver coil is known as a 'Free Induction Decay' (FID). (Westbrook, C. & Kaut, C. 1998)

2.8 Pulse Timing Parameters

Before looking at image contrast and individual pulse sequences it is important to have a understanding of the two pulse timing parameters we use in MRI. TR and TE.

TR: The repetition time: Is the time from the application of one RF pulse to the application of the next RF pulse and is measured in milliseconds (ms). The TR determines the amount of relaxation that is allowed to occur between the end of one RF pulse and the application of the next. Thus TR determines the amount of T1 relaxation that has occurred. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

TE: The echo time: Is the time from the application of the RF pulse to when the signal is read in the receiver coil and is also measured in ms. The TE determines how much T2 decay of transverse magnetisation is allowed to occur before the signal is read. Thus TE controls the amount of T2 relaxation or decay that has occurred. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)
Section 3.  Image Contrast in MRI

3.1 Image Contrast

Images in MRI obtain contrast mainly through the processes of T1 recovery, T2 decay and proton density. T1 recovery and T2 decay have been mentioned above but the PD of a tissue refers to the number of protons per unit volume of a particular tissue.

Different tissues have different properties however to simplify things the vast majority of tissues in the body can be broken down into three general types.

- These are fluids – eg CSF, synovial fluid and oedema.
- Water based tissues – eg muscles, brain and cartilage
- Fat based tissues – fat and bone marrow

(McRobbie et. al. 2003)

3.2 Contrast Mechanisms

T1 Recovery

Different tissues have different T1 times. The T1 time of a tissue refers to how long that tissue takes to recover back to the longitudinal plane after an RF pulse. T1 recovery is a result of spin lattice interactions.

Fat based tissues have short T1 times meaning that they give up the energy absorbed via the RF pulse quickly and thus realign with the longitudinal plane in a short amount of time. This is based on the fact that fat is hydrogen bonded to carbon and molecular mobility is restricted allowing for more efficient spin lattice interactions.

Fluids on the other hand are quite the opposite. Fluids have long T1 times meaning that they take longer to give up the energy absorbed via the RF pulse and thus take longer to realign with the longitudinal plane. Unlike fat based tissue, fluids tend to be hydrogen bonded to oxygen and molecular mobility is higher resulting in spin lattice interactions being less efficient.

In between the two extremes are water based tissues like muscles. These tissues have an intermediate T1 time. (Westbrook, C. & Kaut, C. 1998, McRobbie et. al. 2003 & Woodward, Peggy. 2001)

T2 Decay

Different tissues have different T2 times. The T2 time of a tissue refers to the amount of time it takes for the transverse magnetisation to decay. T2 decay is a result of spin spin interactions meaning the interactions of the small magnetic fields between protons.

Fat based tissues have short T2 times meaning that transverse magnetisation in fat based tissues decays rapidly. This rapid decay is again the result of the bonds between hydrogen and carbon in fat.

Fluids again are the opposite. Fluids have long T2 times meaning that after excitation by an RF pulse fluids hold there transverse magnetisation for longer times. Water based tissues such as muscle, tend to have longer T2 times than fatty based tissue however this is not always the case.

Finally it is important to note that T1 times of tissue are always longer than the T2 times. Also T1 and T2 processes are independent of one another. (Westbrook, C. & Kaut, C. 1998, McRobbie et. al. 2003 & Woodward, Peggy. 2001)
T2* Decay

T2 decay is a result of spin spin interactions between protons. However this is not the only cause of a decaying signal in the transverse plane. Inhomogeneity results in dephasing of the spins and thus contributes to a loss of signal. Inhomogeneities are regions within the field that do not exactly match the overall magnetic field strength.

Thus T2* decay is a faster process than T2 decay due to the combination of factors contributing. Dephasing due to inhomogeneity can be compensated for by using a 180 degree RF pulse. This will be discussed later. (Westbrook, C. & Kaut, C. 1998)

3.3 T1 Contrast

T1 contrast is based on the differences in the T1 times of the tissues in the region of interest.

- Fat based tissues have a short T1 time
- Fluid based tissues have a long T1 time
- Water Based tissues (muscles etc) have an intermediate T1 time.

These differences are exploited using RF pulses and controlling the timing of there application. As the T1 times of fat based tissue are short, after an initial RF pulse fat based tissues relax quickly back into the longitudinal plane where as fluids and water based tissues take a longer period of time to regain there longitudinal magnetisation.

When the next RF pulse is applied the NMV’s of the corresponding tissues are again pushed into the transverse plane. Due to the fact that there is a greater amount of longitudinal magnetisation in fatty based tissues, there will be a corresponding high level of transverse magnetisation after the RF pulse. This equates to a high signal for fatty based tissues.

Thus fat appears bright on T1 images. As fluids take longer to recover there longitudinal magnetisation they will produce less transverse magnetisation after the second RF pulse and subsequently less signal. This equates to fluids appearing dark on T1 images. In between these two extremes are the water based tissues such as muscle. These produce an intermediate signal. (Mcrobbie et. al. 2003)

3.4 T2 Contrast

T2 contrast is based on exploiting the differences in the T2 times of the tissues in the region of interest.

- Fat based tissues have a short T2 time
- Fluid based tissues have a long T2 time
- Water Based tissues (muscles etc) have an intermediate T2 time.

To exploit these differences in T2 times we manipulate and control when we switch on the receiver coil to read the signal. Fluids have a long T2 time meaning that they maintain their transverse magnetisation for a longer period following an RF pulse. On the other hand fat based tissue suffers rapid decay of transverse magnetisation following an RF pulse. Therefore in T2 images fluids will appear bright with a high signal due to good transverse magnetisation and fat based tissue will be dark due to a lack of signal and corresponding poor transverse magnetisation. Again in between these extremes are water based tissues, which return an intermediate signal. (Mcrobbie et. al. 2003)
3.5 PD Contrast

Proton density contrast is a consequence of the relative number of protons per unit volume a tissue contains. To produce contrast differences as a result of PD the transverse component of magnetisation must reflect these differences. Tissues with a high PD will return a high signal, whereas tissues with a low PD will return a low signal. (Westbrook, C. & Kaut, C. 1998)
Section 4. Image Weighting

As we are now aware there are 3 types of contrast in MRI. To demonstrate either T1, PD or T2 contrast specific values of TR and TE are selected for a given pulse sequence. The selection of appropriate TR and TE weights an image so that one contrast mechanism predominates over the other two. (Westbrook, C. & Kaut, C. 1998, Mcrobbie et. al. 2003 & Woodward, Peggy. 2001)

4.1 T1 Weighting

A T1 weighted image is one where the contrast depends on the T1 time differences between the tissues in the region of interest. TR controls how long the NMV’s of the different tissues are allowed to relax back towards the longitudinal plane before being excited by the next RF pulse. To produce contrast the TR must be short enough such that none of the tissues have had time to completely relax back to the longitudinal plane. If the TR is too long then all tissues will have completely relaxed back to the longitudinal plane and upon the next RF pulse there will be no difference in the degree of transverse magnetisation and thus no signal differences to produce contrast.

At the same time a short TE is selected to minimise contrast produced as a result of T2 decay. By selecting a short TE we ensure that minimal differences in signal between tissue types will be present in the transverse plane as a result of the T2 decay process.

- For T1 weighting the TR must be short
- For T1 weighting the TE must be short


Fig 7. Spin Echo images of the brain with a constant TE = 10ms and a variable TR. You can see the effect TR has on controlling T1 image weighting. The signal intensity of the CSF, grey and white matter, and subcutaneous fat is also plotted on the graph adjacent against TR.
4.2 T2 Weighting

A T2 weighted image is one in which contrast is a result of the differences in the T2 times of the tissues being imaged. TE controls to what degree we allow the transverse magnetisation of a tissue to decay before the signal is read. To acquire T2 weighted images the TE must be long enough to give the tissues being imaged time to decay and thus allow us to exploit the differences in their T2 times. If it is too short there will not be enough signal difference between tissues and thus poor contrast.

At the same time a long TR is selected as this minimises contrast produced as a result of T1 relaxation. Using a long TR ensures full longitudinal relaxation minimising signal differences due to T1 relaxation properties between tissues.

- For T2 weighting the TE must be long
- For T2 weighting the TR must be long


4.3 PD weighting

A PD image is one where the difference in the number of protons per unit volume of tissue is the main factor in forming image contrast. Proton density is always present to some extent, however to maximise the degree of PD weighting the effects of T1 and T2 contrast need to be minimised.

This is achieved by using a short TE to minimise contrast due to T2 decay effects and a long TR to minimise contrast as a result of T1 recovery.

- For PD weighting the TE must be short
- For PD weighting the TR must be long

Section 5. Pulse Sequences

5.1 Spin Echo/Fast Spin (Turbo) Echo (Fig 9.)

Spin echo pulse sequences utilize a 90 degree pulse followed by a 180 degree refocusing pulse. The signal that forms after the 180 degree pulse is called a Spin Echo. The advantage of a spin echo is the 180 degree pulse compensates for any magnetic field inhomogeneities and as such eliminates T2* contrast.

**T1 Weighted**
- Short TR to maximise T1 contrast
- Short TE to minimise T2 contrast

**PD Weighted**
- Long TR to minimise T1 contrast
- Short TE to minimise T2 contrast

**T2 Weighted**
- Long TR to minimise T1 contrast
- Long TE to maximise T2 contrast

Modern spin echo sequences tend to be run with multiple 180 degree refocusing pulses. This is called fast spin echo or turbo spin echo. If run as a fast spin-echo sequence, there will be a train of echoes rather than just one.

ie. \(90^\circ\)-180°-echo-180°-echo-180°-echo-180°-echo-180°-echo-180°-echo\) (ETL=6)

The echo train length refers to the number of 180 degree pulses applied.
(Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)

To provide an example we have listed some common examination sequences and there parameters:

- **T1 Axial Brain** TE: Short= 11.4ms TR: Short= 500ms
- **T2 Axial Brain** TE: Long= 105ms TR: Long= 4616ms
- **PD Axial Knee** TE: Short= 23 TR: Long= 2000ms
5.2 Inversion Recovery (Fig 10.)

Inversion recovery is a pulse sequence that begins with a 180 degree inverting pulse. A 90 degree excitation pulse is then applied at a time from the 180 degree inverting pulse known as the TI time.

- **TI**: time from inversion

The contrast of the image depends primarily on the length of the TI. If the 90 degree excitation pulse is applied after the NMV has relaxed back through the transverse plane the contrast in the image depends upon the amount of longitudinal recovery. The resultant image is very heavily T1 weighted, as the 180 degree pulse achieves full saturation and ensures a large contrast difference between differing tissue types as they have a larger dynamic range to relax through compared to a 90 degree pulse.

As with spin echo pulse sequences IR sequences can be run as fast or turbo IR. If run as a Fast IR sequence, there will be a train of echoes rather than just one.

ie. $180^\circ - 90^\circ - 180^\circ$ - echo - $180^\circ$ -echo - $180^\circ$ - echo - $180^\circ$ echo (ETL = 4)

**Tissue Suppression using IR**

At the TI time, a $90^\circ$ pulse flips whatever longitudinal magnetisation is present into the X-Y plane to produce a FID which is then refocussed to form a spin-echo. If the TI is set to correspond to the time when the T1 curve of a particular tissue is crossing the zero line, there will be no longitudinal magnetisation to flip and therefore no signal will arise from that tissue (Fig. 11). This suppression technique is commonly used as:

- ‘STIR’ to suppress the signal from fat with a TI of around 160mSec at 1.5 Tesla.
- ‘FLAIR’ to suppress the signal from CSF with a TI of around 2200mSec at 1.5 Tesla.

(Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)
The time between the $180^\circ$ pulse and the $90^\circ$ pulse is the **TI** (inversion time) and this is the main contrast determinant for inversion recovery scans.

**Note:** STIR (short tau inversion recovery) FLAIR (Fluid Attenuated Inversion Recovery) (Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)

### 5.3 Gradient Echo (Fig 12.)

Gradient echo sequences use low flip angles, this means T1 recovery will take less time and shorter TR intervals are possible. Without the $180$ degree pulse, shorter TE times are also possible. Short TR and TE times allow rapid signal acquisition and so GRE sequences form the basis of many rapid imaging scans such as breath-hold studies and MRA. The signal is refocussed using a rephasing gradient. (Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)

With no $180$ degree pulse, unfortunately GRE sequences are very sensitive to magnetic field inhomogeneities. This may result in unwanted susceptibility artefacts at eg. tissue-air interfaces, but can also be useful due to a greater sensitivity to blood breakdown products. Due to the lack of a $180$ degree refocusing pulse T2 contrast in GRE imaging is referred to as T2* contrast as dephasing due to inhomogeneity will affect the image contrast. Controlling contrast in GRE imaging requires us to manipulate the flip angle, TR and TE. Like spin echo imaging the basic rules for TR and TE still apply. That being short TR and TE for T1 weighting and Long TR and TE for T2 weighting, however the actual values of TR and TE drop considerably in GRE imaging due to the fact that we use a variable flip angle no longer the 90 degrees that is used in SE imaging. The size of the flip angle also has a large bearing on contrast weighting. The smaller the flip angle the faster the recovery of longitudinal magnetisation, this minimises T1 contrast. The larger the flip angle the longer the process of T1 recovery and as such, more emphasis is placed on T1 contrast.
• **T1 Weighting in GRE**
  - Large flip angle 70-110 degrees
  - Short TE 5-10ms
  - Short TR < 50ms

• **T2* Weighting in GRE**
  - Small flip angle 5-20 degrees
  - Long TE 15-25ms
  - Short TR minimum possible depending on number of slices as the small flip angle ensures rapid T1 recovery

• **PD Weighting in GRE**
  - Small flip angle 5-20 degrees
  - Short TE 5-10ms
  - Short TR minimum possible depending on number of slices as the small flip angle ensures rapid T1 recovery

(Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)

### 5.4 Steady State

This is the condition that applies when the TR used is shorter than the T1 and the T2 of the tissue. There will not be enough time for the transverse magnetisation to have decayed away before the next RF pulse is applied.

Each RF pulse produces a FID, each pair of pulses produces a Hahn echo, and stimulated echoes arise from each set of three or more pulses. If the TR is short enough, the tails of the FID’s and the spin echoes merge to produce a continuous signal of varying amplitude. The signal contains both FID and spin-echo components, with the TR being equal to the TAU of the spin-echo.

- RF1 produces FID1
- RF2 produces FID2.
- RF2 will also refocus FID1 to produce a Hahn echo which will occur at the same time as RF3. (A Hahn echo is simply a spin-echo produced with flip angles other than the usual 90 degree and 180 degree combination)

The residual transverse magnetisation can either be incorporated into the sequence or deliberately removed before the next RF pulse. (Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)
5.5 Gradient Echo - Incoherent (eg. SPGR, FLASH) (Fig 13.)

The use of short TR intervals creates the steady state condition, but with these sequences the residual transverse magnetisation is deliberately removed or ‘spoiled’ before the next excitation pulse. Spoiling is done by:

- Gradient spoiling, where a gradient of varying amplitude is turned on just before the next RF pulse. (Siemens’ Flash)
- RF spoiling, where the phase of the RF is changed with each excitation and the receiver locks on only to the phase of the preceding pulse. (GE’s SPGR)

These sequences are used to produce rapid T1 weighted images. eg. breath-hold liver studies. Only the FID is re-phased - the spin-echo is spoiled and not sampled. (Lennon-George, J)

Gradient echo incoherent T1 weighted images utilise sequences with a variety of names depending upon the scanner you are using.

- FLASH: FAST LOW ANGLE SHOT (Siemens)
- SPGR: SPOILED GRADIENT RECALLED ECHO (GE)
- T1-FFE: T1 FAST FIELD ECHO (Phillips)
- RF FAST: RF SPOILED FOURIER ACQUIRED STEADY STATE TECHNIQUE (Picker/Marconi) (McRobbie et. al. 2003)

For GRE incoherent T1 weighted scans the following parameters are suggested

- Flip angle 30-45 degree
- Short TE: minimum possible: often two are used for in and out of phase (minimum TE ensures minimal T2*)
- Short TR: 20-50ms

Note: combination of TR and Flip angle maintains steady state. (McRobbie et. al. 2003)

5.6 Gradient Echo - Coherent (eg. GRASS, FISP, FFE) (Fig 14.)
Short TR values ensure that the steady state is achieved with these sequences. The transverse magnetisation left over at the time of the next excitation pulse is kept coherent using a re-winder gradient after readout. This compensates for the dephasing which occurs due to the phase encoding process.

Coherent gradient echo pulse sequences produce images that are T2* weighted. They can be used to determine if a vessel is patent or if a region contains fluid. (Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)

Gradient echo coherent T2 weighted images utilise sequences with a variety of names depending upon the scanner you are using.

- **FISP**: *Fast Imaging with Steady Precession* (Siemens)
- **GRE**: *Gradient Rewound Echo* (GE)
- **T2-FFE**: *T2 FAST FIELD ECHO* (Phillips)
- **CE FAST**: *CE FOURIER ACQUIRED STEADY STATE TECHNIQUE* (Picker/Marconi) (Mcrobbie et. al. 2003)

For GRE coherent T2* weighted scans the following parameters are suggested

- Flip angle 30-45 degree
- Long TE: 15ms+ (this maximises T2* contrast)
- Short TR : 20-50ms

Note: combination of TR and Flip angle maintains steady state. (Mcrobbie et. al. 2003)

5.7 Echo Planar Pulse Sequences

With echo planar imaging, a single echo train is used to collect data from all lines of K-space during on TR period. Use of this technique subsequently considerably shortens the acquisition time. There are two types of EPI sequences; SE and GRE sequences. All the lines of K-space can be acquired in a single TR (in single shot EPI) or in two or more TR’s. (in multishot EPI) The phase and frequency encoding gradients are turned on and off very rapidly, a technique which allows the rapid filling of K-space. EPI imaging is the sequence of choice for diffusion weighted imaging, for which a EPI SE sequence is typically used. EPI imaging is more vulnerable to magnetic susceptibility effects and provides greater tissue contrast than does imaging with standard GRE sequences. EPI imaging sequences are widely used to assess cerebral perfusion. (Radiographics 2006)
Diffusion Imaging

Diffusion weighting enables the operator to distinguish between rapid diffusion of protons (unrestricted diffusion) and slow diffusion of protons (restricted diffusion). For DWI either an EP or a fast GRE sequence is used and two equal gradient pulses are applied (one on each side of the 180 degree RF pulse in EP sequences). If no net movement of spinning protons occurs between application of the two gradient pulses, the first gradient dephases the spins and the second gradient rephases them. Thus high signal intensity is seen. If however there is net movement of protons these protons will be affected differently by each gradient and will thus return lower signal intensity.

The main use for this technique is in the detection of stroke. These diffusion weighted images are usually combined with an ADC map. ADC being: Apparent Diffusion Coefficient. For the calculation of the ADC map, two sets of images are obtained.

- One set is obtained without the application of a diffusion gradient which results in a set of images with an appearance similar to that of a T2 weighted set of images.
- The other set is obtained with a diffusion gradient.

Without going into the complicated maths involved the ADC calculation is based on the negative logarithm of the ratio of the two sets of images.

The main point of note is being able to recognise the appearance of stroke on Diffusion and ADC maps. Firstly areas of restricted diffusion appear dark on ADC maps and bright on diffusion weighted images. This is due to the fact that stroke results in a region of restricted diffusion within the brain. On the other hand areas of unrestricted diffusion appear bright on ADC maps and dark on diffusion weighted images.
Sometimes the appearance of high signal intensity on DWI also may be due to T2 effects or so called T2 shine through. The absence of these effects on ADC maps allows areas of restricted diffusion from recent stroke to appear dark and areas of unrestricted diffusion in remote tissues or older stroke areas to appear relatively bright.

Thus diffusion and ADC maps together allow us to determine the age of a stroke event. (Fig. 16) Areas affected recently by an acute stroke are characterised by restricted diffusion. (Radiographics 2006)

**Stroke:**
- Bright areas on diffusion weighted images
- Dark areas on ADC maps

Those affected recently by a subacute stroke appear somewhat bright on DWI and may appear moderately so on ADC maps. Areas affected by old stroke are depicted as dark areas on DWI and bright regions on ADC maps. (Radiographics 2006)

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**Fig 16.** Combination of FLAIR DWI and ADC maps for the comparison of stroke. (a.) Axial T2 FLAIR image shows three areas affected by strokes. (arrows). (b.) Axial diffusion weighted EPI image. (c.) Axial ADC map. These two images allow determination of the age of the strokes. The affected areas of the right frontal operculum and left frontal lobe (single arrow) show no evidence of restricted diffusion. This finding indicates that the stroke is old. In contrast the affected area of the left parietal lobe (double arrows) appears bright in (b.) and dark in (c.) giving evidence of restricted diffusion indicative of recent stroke. The area of the second most recent stroke, that in the left frontal lobe (single arrow) appears somewhat bright in both (b.) and (c.); and the area of the oldest stroke, that in the right frontal operculum (single arrow) appears dark in both (b.) and (c.).
Section 6. Spatial Encoding

6.1 Gradient Coils

Gradients are responsible for performing the following three tasks in encoding:

- **Slice Selection**- locating a slice within the scan plane selected
- **Spatially Encoding (locating) signal along the long axis of the anatomy** this is called frequency encoding.
- **Spatially Encoding (locating) signal along the short axis of anatomy** this is called phase encoding.

The signal received is a conglomerate echo from all of the protons excited by the RF pulse. To produce an image, the signal components must be encoded in such a way that they can be traced back to the voxels from which they arose. (Westbrook, C. & Kaut, C. 1998)

6.2 Slice Selection Gradient

For 2D MRI, a slice-select gradient is applied over the top of the main magnetic field and perpendicular to the desired plane:

- **X axis** for a sagittal slice
- **Y axis** for a coronal slice
- **Z axis** for an axial slice

![Fig 17. The Y and Z gradients as slice selectors. The Z axis gradient produces axial slices. The Y axis gradient produces coronal slices.](image)

The precessional frequency will vary along this axis due to the application of the gradient and an RF pulse can be applied with a bandwidth of frequencies so that only protons in the desired slice are excited. (Westbrook, C. & Kaut, C. 1998)
The slice thickness can be decreased by:

- using a steeper gradient
- using a narrower bandwidth RF pulse.

The signal from the single slice must now be resolved along both axes. This is done with frequency and phase encoding. (Westbrook, C. & Kaut, C. 1998)

### 6.3 Frequency Encoding Gradient

The frequency encoding gradient is switched on as the echo is received and will produce a frequency shift dependent on position along this axis. This will allow the signal to be resolved in one direction.

The strength of the frequency encoding gradient required will depend on:

- the field of view
- the receiver bandwidth - i.e. the range of frequencies used to span the FOV (Note: this is different from the transmitter bandwidth used for selective slice excitation). (Westbrook, C. & Kaut, C. 1998)

### 6.4 Phase Encoding Gradient

The phase encoding gradient is pulsed on briefly after the excitation pulse to produce a phase shift between adjacent rows of pixels. This is repeated as many times as there are pixels in the phase encoding direction. With each step, the strength of the gradient is varied slightly from a maximum positive value though zero to a maximum negative value. This will produce a different phase shift between pixel rows for each step. To maximise spatial resolution, the steepest positive and negative steps will need to create a 180 degree phase shift between adjacent columns since this will produce the maximum possible phase separation.
The gradient strength required for these maximum steps will depend on:
- the field of view
- the number of pixels in the phase encoding direction.

(Westbrook, C. & Kaut, C. 1998)

### 6.5 Sampling

The echo is sampled and digitised in the presence of a frequency encoding (or read-out) gradient. This is known as analog to digital conversion. The results are stored in an array known as k-space. (see 7.0 K-Space)

- Converting the analog waveform into a digital signal requires us to sample the waveform at multiple different points. In order to represent this waveform accurately we must sample the waveform at a high enough rate. This is known as the **sampling rate**. The sampling rate used in MRI is known as the Nyquist frequency and relates to The Nyquist Theorem. The Nyquist Theorem is important to MRI and it dictates that any signal must be sampled at least twice per cycle to represent it accurately. (Westbrook, C. & Kaut, C. 1998)
- The **sampling time** is the duration of the gradient while 256 or 512 frequencies are sampled. (eg. 8 msec)
- The **sampling rate** is the rate at which these samples are taken.
- The **receiver bandwidth** is the range of frequencies sampled during read-out (eg 16 kHz)

Reducing the bandwidth will:
- increase sampling time
- increase SNR
- increase echo spacing for FSE scans and increase blurring
- increase chemical shift and susceptibility artefact.

(Lennon-George, J)

### 6.6 Fourier Transformation

Fourier analysis allows any time varying signal (such as a spin-echo) to be represented as a spectrum of the frequencies present.

![Fig 19. Fourier Transformation converts RF intensity vs. time into Signal amplitude vs. frequency thus allowing us to reconstruct images.](image-url)
• The mathematical procedure relating the functions of amplitude against time and amplitude against frequency is known as Fourier Transformation - if one is known, the other can be readily calculated. This process is ideally suited to MR image reconstruction which uses frequency shifts to spatially localise the signal. (Westbrook, C. & Kaut, C. 1998)
Section 7. K-Space

7.1 What is K-Space

K-Space is simply an array of numbers whose Fourier Transformation is the MR image. Simply K-Space is best thought of as temporary image space, which is used to digitize MR signals during acquisition. (Westbrook, C. & Kaut, C. 1998)

In 2DFT imaging, each horizontal row of K-space corresponds to the echo data obtained by digitising the signal from one phase encoding step. The values at the left of each row are collected early in each echo those at the right are obtained late. The centre of the row will correspond to the centre of each echo and will therefore contain the largest values. (Westbrook, C. & Kaut, C. 1998)

Each row of k-space data is obtained by varying the amplitude of the phase encoding gradient. The rows at the top and bottom of the k-space grid are obtained with steep gradient slopes. This will cause smaller signal amplitudes due to gradient-induced dephasing, however this will maximise the phase separation of adjacent points. These lines therefore contribute most towards the spatial resolution of the image. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

The central rows are obtained using shallow gradient slopes so the echo amplitudes will be larger due to little gradient induced dephasing. These lines will contribute most towards the contrast of the image. It should be remembered that individual cells in k-space do not correspond to individual pixels in the image - each pixel is represented in every k-space cell. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

K-Space can be filled in a variety of ways depending upon whether the operator is using a sequential technique, 2D volumetric or 3D volumetric technique.

2D volumetric acquisitions fill one line of K-space for slice 1 and then fill the same line of K-space for slice 2 etc. When one line of K-space has been filled for all slices the next line of K-space is filled and so on until all the K-space data has been acquired for all slices. Once the sequence is finished all of the slices are reconstructed and displayed at the same time. This is the most common method of K-space filling. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

3D volumetric techniques acquire data from an entire volume of tissue. The excitation pulse is not slice selective rather the whole prescribed imaging volume is excited. As such K-space is three dimensional. At the end of the acquisition the volume slab is divided into slices using an extra gradient called the slice select gradient. A 3D FT is then used to reconstruct the images. An example of this technique is a 3D T2* C-spine. (Mcrobbie et. al. 2003)

We don’t really use many sequential techniques however one example is breath hold MRCP sequences. In this technique we acquire a single slice sequentially. Sequential techniques acquire all the data from slice 1 and then move onto slice 2 etc. The slices are then displayed as acquired.
7.2 Manipulating K-Space

K-Space can be manipulated to suit the needs of the MRI examination under progress. Many of these techniques are made possible due to the mirrored halves of K-Space. That being the top and bottom halves as well as the right and left halves of K-Space mirror each other. This can be advantageous to the operator and patient as manipulation of k-space can decrease scan times.

7.3 Partial / Fractional Echo (Fig 21.)

- Relies on echo symmetry from left to right sides of K-Space
- Only the right hand side of the echo is read by the frequency encoding gradient - the remaining portion is calculated. (ie sample the right half of k-space and calculate the left)
- Allows the peak of the echo to occur closer to the RF excitation pulse so shorter TE’s are possible for MRA, T1 spin-echo and fast GRE. (Lennon-George, J)
7.4 Partial Fourier Imaging (Fig. 22)

- Also relies on symmetry - this time between the positive and negative phase encodings or halves of K-Space
- Sample the top half of K-space and calculate the bottom half.
- Actually need to sample slightly over half for phase correction.
- Reduces the acquisition time by nearly one half.
- Reduces the SNR to about 70% of a full acquisition \((1/\sqrt{2})\) (Lennon-George, J)

![Fig 21. K-Space in Partial Echo, a little over half of k-space is acquired with the remainder calculated due to the mirrored nature of k-space.](image)

![Fig 22. K-Space in Partial Fourier Imaging. Little over half of k-space is acquired with the remainder calculated due to the mirrored nature of k-space.](image)

7.5 Rectangular FOV

- Can be used for rectangular anatomical areas (e.g. axial pelvis, sagittal spine)
- The field of view and the image matrix in the phase encoding direction are both reduced - e.g. to ½ or ¾ - this will cut down the imaging time accordingly.
- The gradient increment between each phase encoding step is increased to maintain spatial resolution - i.e. for ½ FOV the increment will be doubled so that the steepest phase encoding steps are still performed.
- SNR will be reduced because of the reduction in the number of phase encodings. (Lennon-George, J)

![Fig 23. K-Space appearance in a Rectangular FOV orientation. By shrinking the phase encoding FOV we reduce the number of phase encoding and thus the image acquisition time](image)
Section 8. SNR and Spatial Resolution

This is the ratio of the amplitude of the signal to the average amplitude of the noise. The signal can be increased or decreased relative to the noise which is constant for each patient. Many factors affect SNR.

8.1 Spatial Resolution

Spatial resolution is determined by the size of the voxel which depends on:

- Slice thickness
- FOV
- Matrix

Slice thickness will determine the ability to resolve small structures in the slice select direction - increasing the slice thickness will decrease resolution and increase partial volume averaging.

The in-plane resolution is determined by the pixel size which is the FOV divided by the matrix. Since both of these parameters may be asymmetrical, resolution in the frequency and phase directions may well be different which will make the pixels rectangular rather than square.

eg:

- FOV = 20cm x 20cm  Matrix = 256 x 224
  Resolution (freq) = 200 / 256 = 0.78 mm. Resolution (phase) = 200 / 224 = 0.89 mm.

- FOV = 22cm x 16cm  Matrix = 512 x 192
  Resolution (freq) = 220 / 512 = 0.43 mm. Resolution (phase) = 160 / 192 = 0.83 mm.
  (Lennon-George, J)

8.2 Coil Types

- A quadrature (or circular polarised) coil offers a theoretical $\sqrt{2}$ SNR improvement over a linear coil of the same size.
- For surface coils, SNR will increase as the size of the coil decreases.
  (Lennon-George, J)

8.3 Proton Density

As the signal in MRI is dependent upon the presence of protons tissues with a high proton density will reproduce a higher signal than those with relatively few protons. A good example of this is the lack of signal received from lung tissue. (Lennon-George, J)
8.4 TR, TE and Flip angle

**TR:** SNR will increase as TR increases until full remagnetisation is reached. (longer TR allows greater longitudinal recovery up until this process is complete.)

**TE:** SNR will decrease as the TE is increased. (long TE times allow large amounts of signal decay prior to the signal being read.)

**Flip \( \alpha \):** SNR will be maximum at the ‘Ernst Angle’, where \( \cos \alpha = \exp(-TR/T1) \).

- For sequences where TR >> T1, this will be 90° - i.e. SNR will increase with the flip angle up to a maximum at 90° as more and more magnetisation is flipped into the X-Y plane.
- For GRE sequences using short TR’s, the Ernst angle will be less than 90°.
  
  Eg. brain tissue with a T1 of 800 mSec:
  
  \[ TR = 3000, \text{ Ernst } \alpha = 89°. \]
  
  \[ TR = 100, \text{ Ernst } \alpha = 28°. \] (Lennon-George, J)

8.5 Voxel Volume

This has a direct bearing on the number of protons contributing to the signal and there is a linear increase in SNR with voxel volume.

- Doubling the slice thickness will double the voxel volume and double the SNR.
- Doubling the FOV in both directions will quadruple the voxel volume and give a four fold increase in SNR.
- Increasing the matrix from 256 x 256 to 512 x 512 (without changing the FOV) will reduce voxel size and SNR by a factor of 4. (This will be partially offset by the \( \sqrt{2} \) increase in SNR from doubling the number of phase encodings) (Lennon-George, J)

8.6 Number of Phase Encodings

- SNR is proportional to the square root of the number of phase encodings.
- Time saving techniques such as partial Fourier imaging will have a SNR penalty as a result of reducing the number of phase encodings.
- If only half of the phase encodings are performed, SNR will be reduced by a factor of \( \sqrt{1/2} = 0.71 \). (Lennon-George, J)

8.7 NEX

This is the number of times that each phase encoding step is repeated. Signal is constant and will add linearly with NEX whereas noise is random and will only increase by a \( \sqrt{2} \) factor.

- Doubling the NEX will change the SNR by a factor of \( \sqrt{2} \) (i.e. 1.41).
- This will also double the scan time.
- Halving the NEX will change the SNR by a factor of \( \sqrt{1/2} \) (i.e.0.71).
  (Lennon-George, J)

8.8 3DFT

- For 3D scans, each extra slice or partition requires a further set of phase encodings which will increase the SNR.
- SNR will therefore increase with the square root of the number of slices - doubling the number of slices will double the acquisition time and provide a \( \sqrt{2} \) increase in SNR. (Lennon-George, J)
8.9 Reducing the Bandwidth

Since noise is considered to be evenly spread throughout all frequencies, reducing the bandwidth will reduce the amount of noise picked up and therefore improve the SNR. Halving the bandwidth will give a theoretical SNR increase of $\sqrt{2}$.

If the bandwidth is reduced from 16 kHz to 8 kHz, then only half as many cycles will occur during the sampling time. To obtain the same number of frequency samples, the sampling time will need to be doubled. This will increase the minimum TE and may reduce the number of slices possible for a given TR.

For FSE scans, increasing the sampling time will increase the echo spacing. This will increase the amount of FSE blurring since a wider range of actual TE values will be sampled.

![Fig 24. Reducing bandwidth will decrease the amount of noise sampled and will require an increase in the sampling time](image)

Reducing the bandwidth will also increase the amount of chemical shift between protons in water and lipid environments.

- At 1.5 Tesla, the 3.5ppm difference in precessional frequency will be $3.5 \times 63 = 220$Hz. With a 32 kHz bandwidth and a 256 matrix, the frequency range across each pixel will be $32,000 / 256 = 125$ kHz per pixel. Misregistration due to chemical shift will therefore be $220$Hz / $125$ KHz/pixel = 1.8 pixels.
- If the bandwidth is reduced to 8kHz, the frequency range across each pixel will be $31$Hz and the misregistration will increase to $220 / 31 = 7.1$ pixels.

(Lennon-George, J)

Reducing the bandwidth will also increase the amount of chemical shift between protons in water and lipid environments.

- At 1.5 Tesla, the 3.5ppm difference in precessional frequency will be $3.5 \times 63 = 220$Hz. With a 32kHz bandwidth and a 256 matrix, the frequency range across each pixel will be $32,000 / 256 = 125$ kHz per pixel. Misregistration due to chemical shift will therefore be $220$Hz / $125$ KHz/pixel = 1.8 pixels.
- If the bandwidth is reduced to 8kHz, the frequency range across each pixel will be $31$Hz and the misregistration will increase to $220 / 31 = 7.1$ pixels.

(Lennon-George, J & McRobbie et. al. 2003)

Thus care must be taken when altering the bandwidth of a sequence as chemical shift artefact can be caused.
<table>
<thead>
<tr>
<th>Parameter</th>
<th>Benefit</th>
<th>Limitation</th>
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<td>increased number of slices</td>
<td>decreased T1 weighting</td>
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Table 1. SNR and Resolution; Parameters and their trade offs.
Section 9. MR Scan Time

9.1 Scan Time

Modification of parameters above which can influence the quality of signal may be costly to scan time as all imaging parameters are related. Thus there has to be a trade of in certain cases between the amount of signal we acquire and the overall scan time. It is possible to set up a sequence with large returns of SNR but an incredibly high scan time which places unrealistic expectations on the patient.

In all MRI systems scan time is dependent on:

- TR
- Number of phase encodings
- Number of NSA’s
- Number of Slices in 3D imaging
- ETL in FSE and the number of shots in echo planar imaging. (EPI)

When setting up sequences every effort should be made to keep the scan time to a minimum while maximising image quality.

In 2D imaging or single slice acquisitions

Scan Time= TR × NPE × NSA

In 3D imaging

Scan Time= TR × NPE × NSA × N_{SL}

In FSE Imaging

Scan Time= TR × NPE × NSA ÷ ETL

In Echo Planar Imaging

Scan Time= TR × NSA × N_{SHOTS}

(Lennon-George, J & Mcrobbie et. al. 2003)
Section 10. Pulse Sequence Optional Extras.

10.1 Pre Saturation Pulses/Bands

Pre-saturation is a technique used to nullify the signal from a volume of tissue. A selective RF pulse saturates a defined volume of tissue so that when the 90° pulse is applied immediately afterwards, there will be no longitudinal magnetisation present to generate signal. There are two common uses:

- 1. To remove signal from tissue outside the area of interest, which may otherwise give rise to motion artefact.

For example, a pre-saturation slab may be placed:
- over the aorta for lumbar spine studies helping reduce pulsation artefact from the moving aorta.
- over the anterior neck in cervical spine studies to reduce swallowing artefact.
- over the anterior abdominal wall to reduce breathing artefact.
- above and below a stack of axial spine slices to reduce the effect of blood and CSF flow into the imaging volume.

(Lennon-George, J)

- 2. In MRA studies to limit the study to either arterial or venous flow. For example:
  - a superior slab will stop the jugular veins appearing on a 2DTOF neck carotid study.
  - an inferior slab will stop the cerebral arteries from appearing on a head MRV study.

(Lennon-George, J)

Presaturation pulses can also be used to help eliminate phase wrap artefact due to the presence of anatomy in the phase direction which extends outside the field of view. More on this in artefacts. (Westbrook, C. & Kaut, C. 1998)

10.2 Fat Suppression FATSAT

Fat Sat utilises the fact that the precessional frequencies of Hydrogen in fat and Hydrogen in water differ by about 3.5ppm. This equals about 220Hz at 1.5 Tesla. This means that it is possible to apply a narrow bandwidth 90° pulse at the resonant frequency of fat which tips only the fat magnetisation into the X-Y plane having no effect on the longitudinal magnetisation of other tissue types. The transverse magnetisation is then destroyed with a spoiler gradient. This is done non-selectively. Meaning with no slice selection gradient, this ensures that the saturation technique applies to fat throughout the entire FOV. The normal pulse sequence follows immediately afterwards, at which time there will be no longitudinal fat magnetisation.

Signal will arise only from hydrogen in other tissues as no longitudinal magnetisation is present in fat tissue which could be converted into a signal in the transverse plane. (Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)
10.3 Magnetisation Transfer

Protons which have long enough T2 values to be imaged with MRI are known as ‘free’ protons. Other protons mainly those bound to large protein molecules have such short T2 times that their signal will have decayed away before it can be sampled. These are known as ‘bound’ protons.

While bound protons themselves do not contribute to the signal, they can affect the relaxation times of neighbouring free protons by way of a transfer of magnetisation between the two. This increases the process of T2 decay.

In tissues with high protein content, such as brain and liver, cross-relaxation between the free and bound protons makes a significant contribution to the overall relaxation process.

A saturation pulse can be applied to the bound protons before the excitation pulse. This will inhibit their ability to assist in the relaxation of the free protons and thereby reduce the signal intensity from these tissues.

Current Uses of MTC:

- **MR Angiography**
  Suppression of background brain tissue with an MT pulse will increase the contrast between brain and flowing blood, allowing an improvement in the demonstration of small vessels.

- **Early Detection of Demyelination**
  Demyelination results in loss of macro-molecular structure, so cross-relaxation makes a smaller contribution to relaxation and signal intensity.

- **Increased Contrast with Gadolinium Enhancement**
  Macro-molecular cross relaxation does not contribute to gadolinium enhancement. Background tissue suppression will therefore render Gd enhanced lesion more conspicuous.

  (Lennon-George, J)

10.4 FSE optimisation

- Used for fast scans such as FSE, FMPIR, EPI.
- Undesirable phase shifts may occur due to eddy currents and RF misalignment.
- These are most noticeable on FSE scans which combine a number of echoes.
- Phase errors between acquired echoes may lead to ghosting or signal loss.
• FSE Optimisation compensates for these phase errors by obtaining additional data prior to image acquisition and then using RF phase shifts and gradient amplitude calibration.  
  (Lennon-George, J)

10.5 Flow Compensation (Gradient Moment Nulling)

Flow compensation is an option the MR user can select to compensate for the effects of flow in the final MR images. This technique compensates for the phase shifts acquired by nuclei as they flow in the presence of a magnetic gradient.

As a proton moves along a gradient, it will experience varying magnetic field strengths which will cause varying rates of precession. This will produce phase shifts with respect to the stationary tissue within the imaging voxel leading to an overall reduction in signal intensity.

Flow compensation uses additional gradient pulses with carefully calculated strengths and durations inserted into the pulse sequence. These are applied before the signal is received to compensate in advance for motion induced dephasing. At the time that the echo is sampled, the net phase shift for stationary and moving tissue will be zero.

• First order GMN compensates for flow with a constant velocity.
• Second order GMR uses more elaborate gradient pulses and also compensates for accelerated flow.
• The operator can choose to apply GMR (flow compensation) to the readout gradient or to the slice select gradient depending on the direction of flow.
• The extra gradient pulses take a finite time to apply and will increase the minimum TE and allow fewer slices for a given TR.  
  (Lennon-George, J)
Section 11. Artefacts

11.1 Phase Wrap

Phase wrap artefact occurs whenever the anatomy continues beyond the borders of the Field of View. It results in the tissues outside the FOV being reproduced at the opposite edge of the scan (wrapped in) in the phase encode direction. The wrapped in image can overlay the real anatomy of interest and significantly degrade the scan quality.

Fig 26. (a) Tissue outside the FOV in the phase encode direction wraps into the final image. (b) With phase oversampling the reconstructed image is larger than the required FOV, and the computer just throws away the unwanted regions.

This phenomenon is most common in the phase encode direction but it also occurs in the slice direction in 3D imaging as the slice encode direction is also phase encoded in this method of imaging. This causes the end slices of the volume to wrap into one another.

One method of reducing phase wrap is to saturate the signals outside the FOV using Sat Bands. This method however is not always successful as sat bands don’t always fully suppress the signal from this tissue.

The most reliable method is to use phase oversampling. This technique increases the FOV in the phase encoding direction and also increases the number of phase encode steps so that the pixel size remains the same. All scanners now offer this phase oversampling option and the user can select a percentage of oversampling to best cover all the anatomy in the FOV.

Implementing phase oversampling will increase the scan time, but as we are acquiring more data signal to noise ratio will also increase.

(Mcrobbie et. al. 2003)
11.2 Gibbs and Truncation Artefacts

Although these two artefacts are two separate entities Gibbs and Truncation artefacts are both sampling artefacts that have similar appearances. This artefact appears as parallel lines of high and low signal intensity adjacent to high contrast anatomic boundaries such as fat-tissue, bone-tissue or CSF-cord. It appears because Fourier Transforms are used to process the signal data.

In theory, any signal can be represented as a sum of multiple sine or cosine waves, each with its own amplitude and frequency. This process is able to accurately reproduce anatomy where there is a gradual change in signal intensity. With abrupt changes, however, Fourier Transformation is less accurate and can introduce erroneous information. This is all based on the amount of times we sample the signal waveform. The higher the number of samples the better we are able to accurately represent the changes.

The artefact is caused by having the acquisition matrix too small. This means that the voxels are too large to accurately represent the high contrast boundary. It is most often a problem in the phase encode direction which is typically smaller than the frequency encode direction to minimise scan time.

As a general rule the phase encode matrix should never be less than half the frequency encode matrix.

![Fig 27.](image)

(a) A low phase encode matrix can cause Gibbs artefact, alternating light and dark bands near a high contrast interface. (b) Increasing the phase encode matrix avoids the artefact

11.3 Motion Artefact (Ghosting / Phase Mismapping)

Motion artefact is one of the most annoying artefacts as we are not able to control the patient. It appears in the phase encoding direction and is not a problem in the frequency direction since all 256 or 512 samples are acquired in a few milliseconds.

In the phase encoding direction, data is collected throughout the sequence and so structures subject to patient and physiological motion will be sampled at different locations. This will produce phase mismapping and if the motion is regular, as in the cases of pulsatile blood flow or respiration, discrete ghosts will appear in the phase encoding direction. The displacement of the ghosts will depend on the period of the motion, the more rapid the motion the more widely spaced the ghosts will be. The intensity of the ghosts will increase with the amplitude of the motion.

Respiratory and Physiological artefacts are often tolerated in the final image or can be dealt with via gating methods or changing the phase and frequency encoding directions to suit the plane of motion. Patient movement however is always a problem, some patients can be particularly uncooperative and it’s up to the MR radiographer to do there best to make them comfortable and encourage them to keep still. New Radial Based K-space filling techniques are also helping overcome this problem. These are known as Blade sequences on Siemens and Propeller on GE. (Westbrook, C. & Kaut, C. 1998, Mcrobbie et. al. 2003 & Lennon-George, J)

Fig 28. Patient motion during an axial T2 Brain

Fig 29. Motion Artefacts due to (a) respiration, (b) cardiac motion, (c) peristalsis.
11.4 Susceptibility and Metal Artefacts

Susceptibility and metal artefacts are closely related. They have the same appearance on the final image except that susceptibility artefacts are more subtle than metal ones. Metal artefacts are characterised by an area of zero signal often with a very high intensity rim on one or two edges. Neighbouring regions demonstrate significant geometric distortion. Susceptibility artefacts may have reduced rather than zero intensity and may not suffer from geometric distortion.

Susceptibility refers to the degree to which a tissue becomes magnetised when placed in a magnetic field. This will affect the local field strength and therefore the precessional frequencies. When two tissues with different susceptibilities are next to each other, at the interface there will be distortion of the magnetic field causing de-phasing and signal loss. Most metals have very high susceptibilities compared to human tissue resulting in large inhomogeneities around the object. Susceptibility and Metal Artefacts always appear worse on GRE imaging as there is not a 180 degree refocussing pulse to compensate for dephasing.

More subtle forms of this artefact can sometimes be seen at natural interfaces such as air and tissue in the paranasal sinuses. Magnetic susceptibility artefact is commonly used to detect cerebral haemorrhage due to the phase distortion and signal loss caused by blood break-down products and thus susceptibility can also be used to an advantage.

The artefact can be reduced by using:
- spin-echo rather than gradient echo
- shorter TE’s
- wider bandwidths
- smaller FOV’s
- 

(McRobbie et. al. 2003 & Lennon-George, J)

11.5 Chemical Shift Artefact

Chemical Shift artefact occurs because of slight differences in the precessional frequencies of protons in different chemical environments. In fat molecules, an electron cloud shields the Hydrogen from the applied magnetic field and reduces its effect. This causes a lower precessional frequency. This is due to the fact that fat consists of carbon molecules bonded to hydrogen.

In water there is no such shielding due to the bonds between hydrogen and oxygen. Thus protons experience the full effect of the applied magnetic field and precess at a faster rate.
The difference in precessional frequencies increases with field strength and is in the order of 3.5 parts per million. At 1.5 Tesla this difference is:

\[ 64 \text{ MHz} \times \frac{3.5}{1,000,000} = 224 \text{ Hz}. \]

Since frequency is used to spatially encode an image, this difference will cause spatial errors to occur during the image reconstruction process. On the image, this artefact will appear in the frequency direction as a bright band of signal at the low frequency side and a void at the high frequency side of a fat-water interface.

A receiver bandwidth of 32 kHz and a 256 image matrix will produce a frequency range across each pixel of

\[ 32,000 \div 256 = 125 \text{ Hz}. \]

The chemical shift in this case will be: \[ 224 \div 125 = 1.8 \text{ pixels}. \]

At a 20 cm FOV this represents a shift of:

\[ 1.8 \times 200 \div 256 = 1.4 \text{ mm} \]

and will increase with increasing FOV.

Chemical shift can be reduced by using:

- a wider bandwidth
- a smaller FOV
- chemical fat saturation or STIR techniques

(Westbrook, C. & Kaut, C. 1998 & Lennon-George, J)

11.6 Phase Cancellation Effect (In and Out of Phase Imaging)

The Phase Cancellation Effect can occur in voxels which contain both water and fat when using a gradient echo sequence. Without the 180 degree refocusing pulse, at the time of echo sampling the signals from fat and water may be out of phase to some degree because of their different precessional frequencies.

The phase relationship between the signal from fat and water varies with time. At certain intervals they will be completely in phase and the signals will add, at other times they will be 180 degrees out of phase causing signal cancellation.

At 1.5 Tesla, the signals will be in phase at multiples of 4.2 milliseconds - ie. at 4.2 ms, 8.4 ms etc. Maximum cancellation will occur in between these times at 2.1 ms, 6.3 ms etc. Choosing a TE at these times will produce maximum or minimum signal for voxels containing both fat and water.
Common sites for chemical shift cancellation include cellular bone marrow and interfaces between tissue and adipose. The effect can be used to advantage in detecting microscopic fat to characterise lesions such as adrenal adenomas. They will show a definite loss of signal intensity on the out of phase image (TE = 2.1 ms) when compared to the in phase image. (TE 4.2 ms).

(Macrobbie et. al. 2003 & Lennon-George, J)

### 11.7 Cross Talk

This refers to the interference which occurs between adjacent slices and is due to the imperfect nature of RF pulses. Ideally, each RF pulse would have a rectangular profile supplying maximum amplitude at the desired frequencies and zero amplitude above or below. Unfortunately, this is not possible and RF pulse profiles have a curved component. Whenever a slice is excited there will always be some excitation of tissue on either side. This can produce saturation effects with an associated loss of signal intensity.

Remedies include:
- inter-slice gaps
- interleaving within the same TR
- interleaving with separate acquisition for odd and even slices (which doubles the time)

(Lennon-George, J)

### 11.8 Zipper Artefacts

The Zipper artefact is caused by RF breakthrough and is probably the most common equipment artefact. This appears as a zipper like band of light and dark pixels passing through the image. It is often two to three pixels wide and extending across an image in the phase encoding direction.

It is usually caused by extraneous RF getting into the examination through some disruption to the RF shielding - eg. a partially open door or a conductive cable passed through a wave-guide. This RF is picked up by the receive coils and mapped into the image. Another common cause is RF emission is from equipment within the examination room such as a pulse oximeter. (Lennon-George, J)
11.9 Reconstruction Artefacts

These appear as a cross hatch or herringbone pattern usually over the entire image and in some cases multiple images. They are also referred to as herringbone or corduroy artefacts. They are caused by data handling or reconstruction errors and may affect only one image of a series. The cross hatch appearance is due to noise spikes in the raw data of K-Space.

Once the Fourier Transform is run they appear in the image as cross hatching. Reconstructing the raw data again may solve the problem although it will often be necessary to repeat the sequence. (Lennon-George, J)

Fig 35. Herring Bone artefact often caused by a noise spike in K-space.

11.10 Artefacts Due to Equipment Failure

MRI systems are complex and breakdowns in various components do happen. The resultant images can be particularly strange. Their appearance however can often help in identifying the problem. A skewing of the entire image can be an indication of a malfunctioning gradient coil or amplifier. Sometimes although rarely this is also due to the loss of a shim coil. Reduced intensity or shading through an entire image is often the result of a failure in the transmit or receive RF chain. (Westbrook, C. & Kaut, C. 1998)

Fig 36. (a) Loss of a gradient or shim coil causes skewing of the entire image. (b) Shading caused by an RF coil fault.
**Section 12. Flow Effects and MR Angiography**

Magnetic Resonance angiography (MRA) uses the inherent motion sensitivity of MRI to visualise blood flow from within vessels. There are two main types of flow imaging methods that can be used and they rely on the difference of moving spins to produce a useful image.

**TOF – Time of Flight:** relies on flow dependent changes in longitudinal magnetisation.

**PC – Phase Contrast:** relies on flow dependent changes in transverse magnetisation.

### 12.1 TOF Effect

The moving spins in MRI such as flowing blood have a significant effect on the final signal. These effects can be image degrading such as artefacts. However they can also be useful to produce MR angiograms. Blood flow can be one of the most annoying artefacts in MR due to its variably nature and also the pulsation of vessels as blood flows.

The first effect of blood flow is its movement through slices. The TOF effect arises because some or all of the blood within an imaged slice is replaced during the repetition time or between the application of the 90 degree and 180 degree pulses TE/2.

In spin echo imaging three types of signal behaviour can be identified due to the moving nature of blood. Firstly if blood is “stationary” the spins will experience both the 90 degree and 180 degree pulse generating an echo. Now whilst blood is never truly stationary unless your patient is “dead” the signal returned from it is dependent on the blood velocity and the slice thickness. Thus if we have a thick slice, slow flowing blood may appear to be “stationary” and thus give us a signal.

When the blood velocity reaches the point where the blood within a slice has completely left in the time between the 90 degree and 180 degree pulses then no signal is returned. This is known as high velocity signal loss or wash out.

The third type of signal behaviour occurs at an intermediate stage compared to the above two examples. This occurs when blood is moving at a speed such that only some of the blood leaves between the two excitation pulses. The portion that remains is diluted by fresh blood flowing into the slice. Only the blood that remains in the slice receiving both pulses will produce a signal. This results in a reduction of signal due to the wash out effect. (Westbrook, C. & Kaut, C. 1998 & Mcr Robbie et. al. 2003)

![Fig 37. TOF effect in Spin Echo imaging. If blood is stationary (a) it receives both the 90 and 180 degree pulses and produces a spin echo like any other tissue. When \( v \geq 2z/TE \) (c) all the blood that received the 90 degree pulse has moved out of the slice during TE/2 the time between the 90 and 180 degree pulses, so the vessel has a flow void. At intermediate velocities, e.g. \( v = z/TE \) (b) there will be a reduced signal compared to (a) but greater than (c) due to partial wash-out of blood during TE/2. Note: \( v \) is velocity and \( z \) is slice thickness.](image-url)
Gradient echo images are different due to the lack of a 180 degree refocussing pulse. GRE sequences only ever demonstrate flow related enhancement and never show wash out signal losses. Depending on the velocity of the blood differing degrees of flow related enhancement will be observed. In the case of slow flow a GE sequence will submit this blood to multiple RF pulses and thus the blood will quickly become saturated and no longer contribute to signal. In the case of fast flow, fresh blood will be moving into an imaging slice and will return high signal as it experiences an RF pulse but is not already saturated as this is the first such pulse it has experienced. (Westbrook, C. & Kaut, C. 1998 & Mcrobbie et. al. 2003)

12.2 Phase Shift Effects.

The second mechanism responsible for affecting the signal intensity of flow is the spin phase phenomenon. We should remember that one of the effects of placing protons in a static magnetic field and submitting them to gradients is that the spins of these protons will precess in proportion to the strength and duration of the gradient that is applied to them. The slice select and frequency encoding gradients are balanced so that no net phase shift exists after slice selection and at the echo time. The phase encoding gradient needs to be changed to acquire the phase shifts for spatial encoding. Unfortunately balancing the gradients only works for static protons. Those that move even in the presence of balanced gradients will acquire a phase shift which is directly related to the velocity of the spins. The result is the characteristic flow artefact seen in the phase encoding direction.

The major problem with blood flowing through a vessel lies in the fact that individual spins are often moving at a variety of velocities. As blood vessels are typically only a few voxels across each voxel can contain a variety of spins at differing velocities and thus each of these spins will have a different phase shift with the net effect being that if this intervoxel dephasing is sufficient complete signal cancellation can occur.

This effect is often seen in “laminar flow” where maximum velocity is at the middle of the vessel and stationary blood at the edges. This occurs as adjacent layers of fluid glide past each other without mixing. The result of this is a misrepresentation of the apparent width of the blood vessel. It appears that the vessels diameter is actually less than it truly is.

Other varieties of flow are seen below

Flow Types

- **Laminar flow** – typical pattern of flow which varies from slowest at the vessel wall (due to friction) to fastest at the centre of the lumen.
- **Turbulent flow** – random fluctuation of velocities.
- **Vortex flow** – at and immediately after a stricture there will be high velocity flow centrally and spiralling flow near the walls.
- **Stagnant flow** – flow is so slow that it behaves like stationary tissue in terms of signal characteristics.  
  (Lennon-George, J)

12.3 Avoiding Flow Artefacts

Avoiding these effects involves dealing with the above two phenomenon. The TOF effect (also known as the in flow effect.) is most effectively dealt with by eliminating the signal from spins before they enter the imaging slice. This can be achieved using presaturation pulses adjacent to the imaging slice. This ensures that blood entering the imaging slices is already saturated and thus will not return any signal. The disadvantage is that as presaturation pulses are another RF pulse the TR needs to be increased to allow for their use.

Reduction of intervoxel phase dephasing is achieved using a technique known as Gradient Moment Nulling or flow compensation. This has been mentioned previously. The drawback to this technique is it results in an increase in the minimum TE.


12.4 Time of Flight MR Angiography

Time of Flight MR angiography uses gradient echo sequences with short TR’s and TE’s to saturate out the signal from stationary tissue while moving spins flowing into an imaging slice yield high signal. Post processing in the form of MIP’s then accentuates these differences and creates a diagnostic image.

2D TOF

Involves the sequential acquisition of many thin (1-2mm) slices. These slices are then MIP’ed to create an image of the vessels. The advantages of this method are good background tissue suppression and the technique is also sensitive to slow flow due to the thin nature of the slices. The disadvantages are the long TE’s required when selecting thin slices leading to a loss of signal from turbulent or complex flow. (Mcrobbie et. al. 2003)
Advantages of 2DTOF as opposed to 3D include:

- it is more sensitive to slow flow
- it provides superior blood / background contrast
- it is possible to image longer lengths of vessel since the slice by slice approach means that this technique does not suffer from the eventual saturation of flowing nuclei which can be a problem with the 3D approach. (Lennon-George, J)

Applications include:

- neck carotid arteries
- cerebral venous system (MRV)
- peripheral vessels such as femoral arteries and run offs (Lennon-George, J)

3D TOF MRA

Also relying on the flow related enhancement effect, 3D TOF studies excite a whole volume of tissue which is then divided into thin contiguous slices with an extra set of phase encoding steps. There is a limit to the thickness of the slab, since eventually the fresh nuclei entering the volume will become saturated and no longer return high signal relative to the stationary background tissue. As with 2D TOF, a pre-saturation slab can be used to limit the study to flow in one direction only. (Mcrobbie et. al. 2003)

A typical single slab Circle of Willis study may have parameters along the lines of:

TR = 40mSec, TE = 2mSec (minimum possible), flip angle = 20°, 60 slices 0.7mm thick. (Lennon-George, J)
Advantages of 3DTOF as opposed to 2DTOF include:

- greater SNR due to the excitation of a thick slab rather than a thin slice.
- This will allow better resolution through thinner slices, larger matrices and smaller FOV’s.
- less intra voxel dephasing will occur due to the reduced voxel size.
- selecting a slab rather than a thin slice is less demanding on the gradient system – this will allow shorter TE times which will also reduce intra voxel dephasing. (Lennon-George, J)

Applications:
3D TOF MRA is ideally suited to anatomy such as the Circle of Willis where high detail is required over a limited area. (Lennon-George, J)

12.5 Phase Contrast Angiography

Phase Contrast MRA relies on detecting changes in the phase of blood’s transverse magnetisation as it moves along a magnetic field gradient. PCA makes use of bipolar gradients in opposite polarity and direction thus becoming sensitive to flow whilst suppressing background tissue that is stationary. PC MRA then subtracts the two acquisitions so that the signals from stationary spins are subtracted out leaving only the signals from flowing spins.

Sensitisation to flow is obtained along the direction of the applied bipolar gradient. If the bipolar gradient pulses are applied in the Z axis direction then phase shifts are induced in flow that occurs in this direction. This sensitises the PC MRA to flow in the head to foot direction. Since flow can occur in all three directions bipolar gradients are applied in all three. These are known as flow encoding axes. However the more flow encoded axes the greater the imaging time. As with TOF MRA images can be acquired in 2D or 3D.

(McRobbie et. al. 2003)

Velocity Encoding

PC-MRA can also be sensitised to flow velocity. Velocity encoding or VENC compensates for projected flow velocity within vessels by controlling the strength of the bipolar gradients. Depending on the velocity of the blood flow within a vessel the VENC is set to match. Only vessels with blood flow at this VENC or lower will be depicted in the final image. (McRobbie et. al. 2003)
Advantages of PC-MRA

- Sensitivity to a variety of vascular velocities
- Sensitivity to flow within the FOV
- Reduced intra voxel dephasing
- Increased background suppression (Mcrobbie et. al. 2003)

Disadvantages of PC-MRA

- Long imaging times with 3D
- More sensitive to turbulence (Mcrobbie et. al. 2003)

12.6 MR Angiography Optional Extras

There are a wide variety of options which the user can select when setting up a MR angiographic sequence. The vast majority of which are listed below.

12.7 Pre-saturation

As previously discussed, pre saturation can be used to limit a study to either arterial or venous flow by suppressing the signal from nuclei flowing into the slab from one direction. (Lennon-George, J)

12.8 Flow Compensation

As discussed under ‘Pulse Sequence Options’ flow compensation uses additional gradient lobes to compensate for phase shifts acquired by moving nuclei. This will increase the signal from flowing blood and improve the CNR of blood to background tissue.

The signal enhancement will be off-set to some degree since longer echo times are required, however the fact that flow compensation is widely used for both 2D and 3D TOF studies suggests that it must provide a net gain. (Lennon-George, J)

12.9 Magnetisation Transfer

As discussed under ‘Pulse Sequence Options’, a magnetisation transfer saturation pulse will suppress the signal from tissues such as brain which have a high protein content. This will increase the contrast between blood and background tissue and assist in the visualisation of smaller vessels.
The MT pulse requires a significant increase in the TR (e.g., from 21mSec – 36mSec) which will increase the sequence time proportionally. MT will not suppress lipid signal and so in some situations it may not justify the increase in scan time.

(Lennon-George, J)

12.10 MOTSA (Multiple Overlapping Thin Slab Acquisition)

This technique exploits the resolution and SNR advantages of the 3D approach while getting around its major limitation – namely the fact that only short sections can be imaged due to the saturation of nuclei moving through the slab.

Multiple slabs are imaged individually with some overlap of slices since the image quality of the end slices is sub-optimal. The slabs are then combined to produce high resolution MIP images of the whole section.

![Fig 44. Demonstrating progressive saturation of laminar flow in a 3D volume. In this example of a carotid artery flow entering the artery will slowly become saturated as it moves through the thick volume of tissue being excited for this study. This effect is most noticeable at the distal end of the volume and at the edges of the vessel due to the slower flowing blood present here. By using MOTSA we can overcome this problem.](image)

The penalty paid is in terms of time – if three slabs are required to cover the anatomy, the imaging time will be three times that for a single slab. A four slab cerebral study will cover much of the cerebral vasculature, but may take 7-10 minutes to acquire.

(McRobbie et. al. 2003 & Lennon-George, J)

![Fig 45. MOTSA technique used to reduce progressive saturation in 3D TOF. Note the Venetian blind artefact at the slab boundaries](image)
12.11 Ramped Flip Angle Excitation

This technique varies the flip angle of the excitation pulse across the slab from a minimum at the entry slice (for the desired direction of flow) to a maximum at the exit slice.

This is designed to decrease the saturation effect on blood flowing in the desired direction while increasing the saturation for nuclei flowing the opposite way. It should improve the visualisation of slow flow and assist in the suppression of signal from unwanted vessels. An example is the TONE MRA we often run for better 3D MRA in the brain. (Lennon-George, J)

12.12 MIP (Maximum Intensity Projection)

This technique is routinely used to reformat MRA images. The stack of 2D or 3D slices is treated as a volume of data which can be projected in any orientation as a 2D map of the highest intensity pixels. This separates out the vessels and allows them to be displayed at the orientation most appropriate to their anatomical position. It is also possible to combine a series of images at incremental projection angles into a cine loop to give a 3 dimensional perception of the vascular anatomy. (Lennon-George, J)
Section 13. Contrast Agents in MRI

In spite of the generally excellent soft tissue contrast in MR images there are still circumstances where the relaxation properties of normal and pathological tissue are similar. Indications include tumours, infection, infarction, inflammation, disc v scarring in post operative lumbar spines, MR angiography and MR arthrography. (Lennon-George, J)

13.1 Molecular Make-up

Gadolinium, the paramagnetic ion used in MR contrast agents, has 7 unpaired electrons which gives rise to a large magnetic moment and makes it the most efficient proton relaxation enhancer.

In its free ionic form, Gadolinium is not suitable for use as a contrast agent because:
- it is toxic
- it forms insoluble salts which are taken up by the liver

It needs to be bound to a carrier molecule called a ligand to produce a combined molecule called a chelate. eg:
- Gd plus DTPA = Gd-DTPA or gadopenetate (Magnevist)
- Gd plus DTPA-BMA = Gd-DTPA-BMA or gadodiamide - a non-ionic derivative (Omniscan)

(Lennon-George, J)

13.2 T1 Relaxation Effects

T1 relaxation is facilitated by fluctuating magnetic fields such as those produced by the tumbling motion of neighbouring molecules (dipole-dipole interactions). Molecules the size of fat tumble and therefore produce field fluctuations at a frequency near the Larmor frequency which will provide the most efficient T1 relaxation.

Normally, water molecules tumble too rapidly for efficient relaxation. If a tumbling molecule with a large magnetic moment (i.e. Gadolinium) is placed in the presence of water protons, local field fluctuations will be created with a substantial component at or near the Larmor frequency - so T1 relaxation will be enhanced.

Both T1 and T2 relaxation times are shortened although the effect is much more pronounced for T1. (Lennon-George, J)

13.3 Enhancement

- Areas of hyper-vascularity - eg scar tissue
- Vessels - contrast enhanced MRA
- Blood brain barrier :
  - Extra-axial areas outside the BBB such as the falx, pituitary and choroid plexus will enhance.
  - With an intact BBB, intra-axial areas will not enhance. Disruption of the BBB will allow accumulation of gadolinium in lesions such as neoplasms, infarcts and abscesses.

(Lennon-George, J)

13.4 Biological Considerations and NSF

<table>
<thead>
<tr>
<th>Distribution</th>
<th>Distribution half life (mean +/- SD) :</th>
</tr>
</thead>
<tbody>
<tr>
<td>Magnevist</td>
<td>.2 +/- .13 hours</td>
</tr>
<tr>
<td>Omniscan</td>
<td>3.7 +/- 2.7 minutes</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Clearance</th>
<th>Renal and plasma clearance rates the same</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.7 ml/min/kg</td>
<td>Glomerular filtration</td>
</tr>
</tbody>
</table>

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Excretion
- 80% via kidneys in 3 hours
- 95% via kidneys in 24 hours
- 98% via faeces and urine in one week

Toxicity
- Low toxicity
  - Min lethal dose for rats and mice > 200 x patient imaging dose

Dose
- Standard dose is 0.1 mmol/kg (.2ml/kg)
- Maximum dose = triple dose = 0.3mmol/kg

Contraindications
- Previous reaction
  - Haemolytic anaemia
  - Sickle-cell anaemia
  - Impaired renal function
  - Abnormal liver function
  - Pregnancy - only use if potential benefits outweigh risks
    - will cross the placenta
  - Nursing mothers – local policy should be applied

Side Effects
- Slight increase in bilirubin & blood iron
  - Headaches (9%)
  - Nausea (4%)
  - Vomiting (2%)
  - GI upset < 1%
  - Rash <1%
  - 2 reported deaths (in 500,000)

Gadolinium and Nephrogenic Systemic Fibrosis

Studies have indicated a correlation between the use of Gadolinium based contrast agents and NSF. There have been no reports in patients with normal renal function. Virtually all cases have occurred in dialysis dependant patients. Most cases have occurred at higher doses such as those used for MRA examinations. Cumulative doses also need consideration.

All reported cases have occurred with an associated pro-inflammatory event-surgery, systemic infection, thromboembolic event (eg limb ischaemia). Liver transplant patients are also at risk.

Due to this recent complication prudent use of contrast in MRI is needed. It is particularly important to look for patients with renal dysfunction as this adversely affects the patient’s ability to excrete the Gadolinium post injection. For specific guidelines it is important to refer to your MRI Senior and company policy. However a useful reference is listed below. (Zappia, Daniel 2007)

**Gadolinium Administration Guidelines**

*Any gadolinium given should be at the lowest clinically useful dose friendly.*

1. **Severe renal failure** – GFR <30ml/min/1.73m
   - Absolute contra-indication unless contrast administration is essential and the benefits out way the risks.

2. **Moderate renal failure** – GFR = 30-60 ml/min/1.73m
   - gadolinium given with caution if inpatient and pro-inflammatory event.
3. Renal Function >60 ml min/1.73m

- Gadolinium administration can occur
  In at risk patients doses should not be repeated within the same 7 day period.
  In at risk patients the FDA recommends dialysis despite the fact that current studies are yet to prove it to be of benefit. Dialysis in at risk patients should be performed within two hours.

There are many useful GFR calculators on the internet the following address provides one example. - http://nephron.com/cgi-bin/LMDRG_GFR.cgi
(Zappia, Daniel 2007)

13.5 Australian Approval

Adults
- IV only
- intra-cranial lesions with abnormal bbb or vascularity
- spinal lesions
- whole body applications - neck, thorax, abdomen, pelvis, female breast, musculo-skeletal system (Lennon-George, J)

Children (including neonates)
- intra-cranial lesions with abnormal bbb or vascularity
- spinal lesions. (Lennon-George, J)

13.6 RANZCR Guidelines on the use of Gadolinium-Containing MRI Contrast Agents

The Royal Australian and New Zealand College of Radiologist (RANZCR) publishes the Guidelines on the use of Gaolinium-Containing MRI Contrast Agents, the current version of the guidelines were published in 2007. The AIR recommends that radiographers involved MRI examinations involving the administration of intravenous contrast for MRI read and makes themselves familiar with the content of these guidelines.

A copy of the guidelines is available from the documents section of the RANZCR website www.ranzcr.edu.au
* Please note that registration with RANZCR or the website is required to access this document
Section 14. Hardware

14.1 Magnetism

Magnetic susceptibility is a measure of the extent to which a substance becomes magnetised when placed in a magnetic field. These interactions will concentrate or disperse the lines of the magnetic field and either increase or decrease the local field strength.

Paramagnetic substances

- magnetise in the direction of the applied field ie. have a positive susceptibility (+10)
- increase the local field strength
- include chelates of metals such as iron, copper and gadolinium.

Diamagnetic substances

- magnetise in the opposite direction to the applied field ie. a negative susceptibility (-1)
- reduce the local field strength
- include water and most organic molecules

Ferromagnetic substances

- have an extremely large positive susceptibility (+25,000)
- can substantially increase the induced field and so are used in electromagnets
- have a magnet ‘memory’ ie. will retain magnetisation after the applied field is removed
- include iron, nickel and cobalt alloys

Superparamagnetic substances

- have a large positive susceptibility (+5000)
- have no magnetic ‘memory’
- small Fe₃SO₄ particles
  (Lennon-George, J)

14.2 Magnets

Permanent Magnets

- vertically orientated field requiring solenoidal RF coils
- low field strength - up to around 0.3 Tesla
- very low fringe fields since the magnet frame provides a return path for the magnetic flux
- very heavy

Air Core Resistive Magnets

- horizontally oriented field produced by current flowing in a resistive coil of wire
- require a continual supply of electrical power
- low field strength - up to around 0.3 Tesla
- can be turned off at the end of each day

Iron Core Resistive Magnets

- hybrid of permanent and resistive designs
- vertically oriented field requiring solenoidal RF coils
- low field - up to around 0.6 Tesla
- low fringe fields

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Superconductive Magnets

- horizontally oriented field produced by current flowing in a coil wound with an alloy of Niobium-Titanium which loses all resistance when cooled to around 10 degrees K
- this is achieved by immersion in a bath of liquid helium at about 4 degrees K
- once ramped, no electrical power supply is required since the resistance is zero
- capable of producing high, stable and homogeneous magnetic fields
- significant fringe fields are also produced

14.3 Quench

If the coil temperature rises above the superconductivity threshold, the windings will become resistive which creates further heat. This will cause an sudden explosive boil-off of cryogens which should escape via the quench pipe.

There is, however, the potential for cryogens to escape into the examination room which will lower the temperature considerably and fill the room with an opaque mist. Although Helium is not toxic, it will displace the oxygen and present a risk of asphyxiation. The remaining oxygen will be found at floor level since it is heavier than Helium.

(Lennon-George, J)

14.4 Shimming

This is done when the magnet is installed and refers to adjustments made to improve the homogeneity of the static field. Current systems quote homogeneity figures of around 4 ppm over a 50 cm volume and 0.1 ppm over a 10 cm volume. The shimming process identifies areas where the field is too low or too high and compensates for these variations.

Passive shimming is done by gluing or bolting pieces of metal to a jacket which is located next to the coil windings.

Active shimming is done with resistive or superconductive shim coils. The amplitude and direction of the current flowing through these coils can be adjusted so that the small magnetic fields which they produce will smooth out the total field strength at each location.

14.5 Magnetic Shielding

This is done to reduce the size of the fringe field which may extend over a considerable area with a high field superconductive system. This will minimise both the exclusion zone for the general public as well as the area over which the homogeneity of the magnetic field can be affected by metallic objects such as lifts and motor vehicles.

Passive shielding is done with steel plates either as part of the magnet gantry or attached to the walls, ceiling and floor of the examination room.

Active shielding uses a second coil external to the windings of the main magnetic field. The field generated by this secondary coil is oriented to oppose the main field and reduce its spread.

14.6 RF Shielding

Magnetic shielding should not to be confused with RF shielding which keeps unwanted RF from entering the examination room and interfering with the detection of the MR signal. RF shielding involves placing a Faraday Cage around the scan room. This cage prevents any extraneous RF from entering or exiting the room.

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14.7 Gradient System

The gradient system is designed to create linear variations in the magnetic field to allow spatial localisation of the MR signal. Three separate coils permit a gradient to be applied in the X, Y or Z direction. Two gradients used in combination will permit oblique slices to be acquired.

Spatial resolution and speed depend on gradient performance which is usually stated in terms of maximum amplitude and rise time.
(Lennon-George, J)

**Maximum Amplitude**

This is the highest gradient strength that the amplifier and coils can sustain. Higher amplitudes will allow better resolution through smaller FOV’s and thinner slices. It is expressed in terms of mTesla per metre. (or sometimes Gauss per cm)
Typical values range from 10 to 25 mT/m       (1 - 2.5 Gauss/cm)

**Rise Time**

The rise time is the period necessary for the gradient to reach its maximum amplitude - e.g. to go from zero to 23mT/m.
The shorter this time the better - this will allow shorter TE’s, more slices per TR interval and shorter echo spacing on FSE sequences.
Rise times are expressed in mSec.

**Slew Rate**

The slew rate is described as the strength of the gradient over distance as a function of time. Typical gradient slew rates are in the order of 70mT/m s. High speed gradients are generally even stronger at 120mT/m/s.
Comparisons should also take into account the ‘Duty Cycle’ - which is the percentage of time that the gradients can be switched on at maximum amplitude during a TR interval.

**Eddy Currents**

Gradient switching will cause eddy currents to be induced in conductors including the patient, the cables and most importantly, the scanner itself. Unless compensated for, they will distort the gradient wave-form which will impair the overall performance and create artefacts. The two common methods of minimising eddy currents are:

- eddy current compensation or pre-emphasis - whereby the gradient currents are deliberately altered to compensate for the distortions created by eddy currents.
- shielded gradients which incorporate a second set of gradient windings which are energised in the opposite direction to counteract the effect of the imaging gradient.

Gradient echo and particularly fast gradient echo sequences have the greatest potential to produce eddy currents because of their intensive use of rapid gradient switching.
(Lennon-George, J)
RF coils are required to:
Deliver the rotating $B_1$ field for excitation and rephasing and receive the MR signal.

linear and Quadrature (or Circular Polarised) Coils

The rotating magnetisation can be seen as having one component moving left and right along the horizontal axis and a second component moving up and down along the vertical axis. Linear coils will receive only one of these components. Quadrature coils use two receivers to detect both components which are phase shifted by 90 degrees. When the signal in the first coil is maximum, it will be minimal in the second coil (and vice versa). The result is the detection of two signals, one at the phase of the RF oscillator in the scanner and one phase shifted by 90 degrees. These are known as the ‘real’ and ‘imaginary’ components respectively. Compared to linear design, this results in a 1.4 (square root 2) increase in the SNR of the signal received and a much more efficient transfer of power to the patient in terms of the excitation and rephasing pulses.

During readout, both signals are detected and digitised so each k-space cell contains a real and an imaginary number. ‘Real’ or ‘imaginary’ images may be reconstructed by Fourier transformation of the relevant data, or more commonly, a ‘magnitude’ image is created using the square root of the sum of the squares of the real and imaginary components.

$$\text{Mag} = \sqrt{\text{Re}^2 + \text{Im}^2}$$

Volume Coils

- These coils can be used to transmit and to receive.
- They usually work in quadrature.
- The body coil and the head coil are examples.
- The provide a very uniform SNR over the volume.
(Lennon-George, J)

Surface Coils

- These are receive only coils.
- They must be positioned perpendicular to the XY plane - ie. perpendicular to the rotating magnetisation.
- The body coil is used to transmit and at these times surface coils must be de-coupled.
- They can be linear or quadrature.
- The signal drops off with the square of the distance from the coil.
- They are designed to improve SNR over the area of interest - often purpose built for specific studies.
- Noise and motion artefact from anatomy outside of the coil's sensitive area will not degrade the image.
- The SNR gain will allow higher resolution studies to be performed.
### Phased Array Coils

- These consist of two or more coils, each with its own receiver channel.
- After processing, the signals are combined to produce a single image with a larger FOV.
- They can be either linear or quadrature.
- They combine the SNR advantage of small coils with:
  - the ability to cover a longer area - eg. for spine imaging
  - a more uniform signal intensity across the volume eg. for abdominal scans.

(Lennon-George, J)
References

Kaut Roth, Carolyn, 2002, MRI: Imaging Procedures, Patient Care and Safety, Springer Publishing, Verlag, USA.


Lennon-George Jon, Perrett Medical Imaging, MRI Radiographer Accreditation Study Guide


Radiographics 2006, MR Pulse Sequences: What every radiologist wants to know but is afraid to ask


Image References

Lennon-George Jon, Perrett Medical Imaging, MRI Radiographer Accreditation Study Guide
Figures 9,10,11,12,13,14,21,22,23,24,25,39

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Figures 1,4,5,6,17,18,19,20,30,32 also (Table 1)