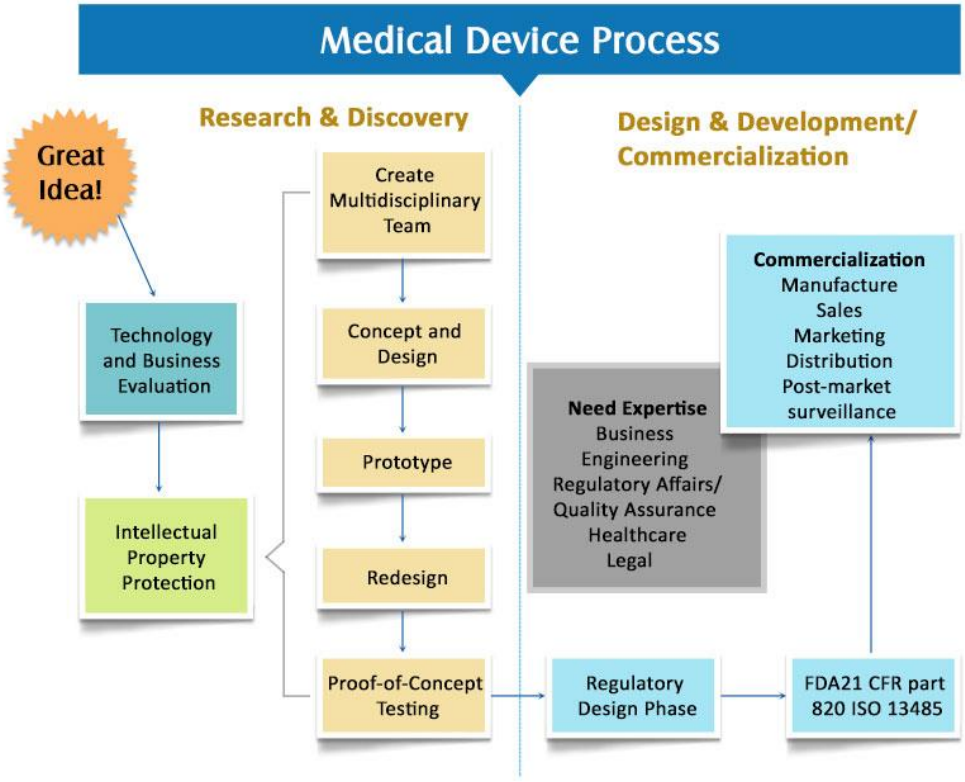
Lecture one

Concepts of Engineering Design

Introduction

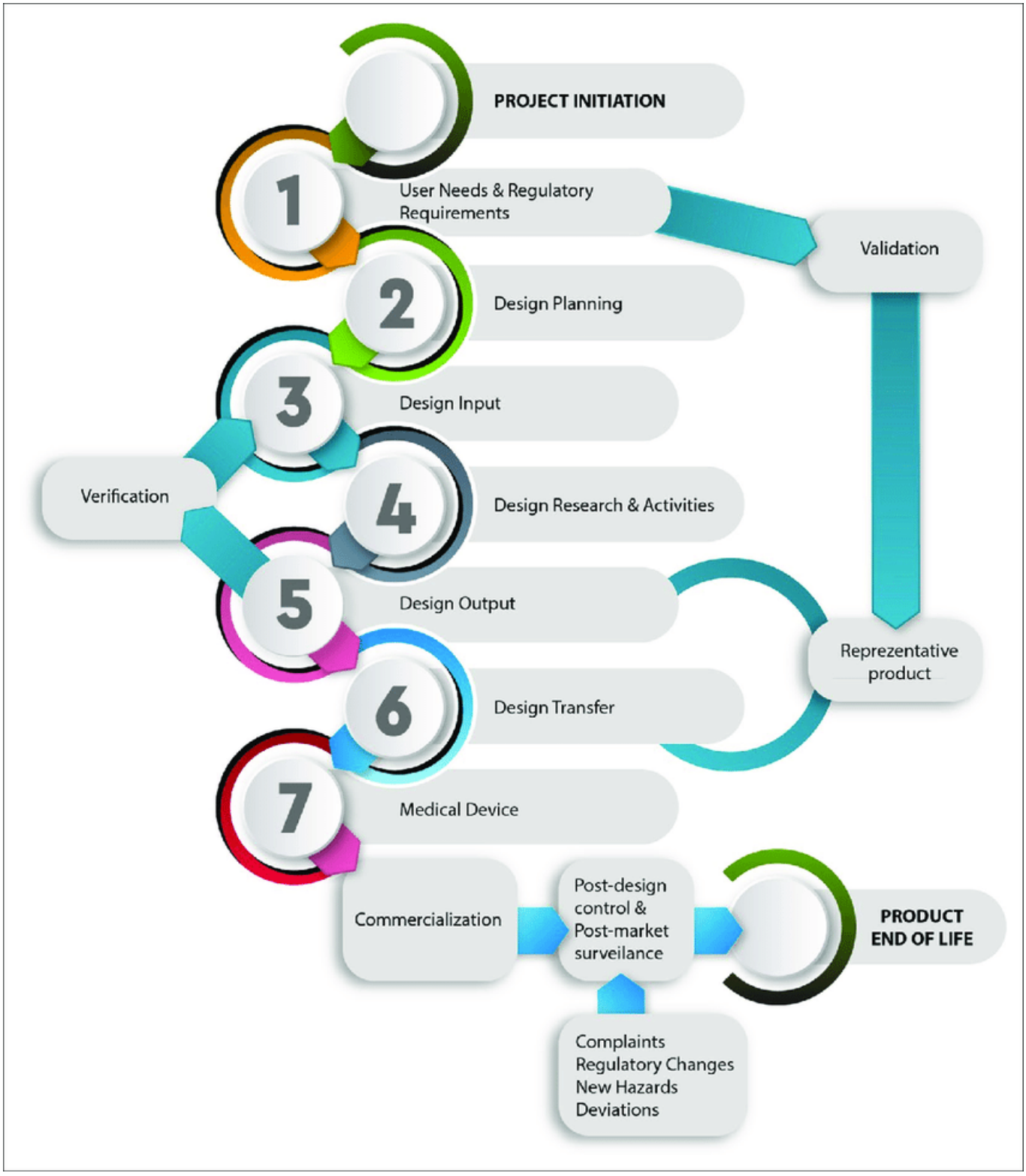
According to The Accreditation Board for Engineering and Technology (**ABET**), The Engineering design is the process of devising a system, component, or process to meet desired needs. Among the fundamental elements of the design process are the establishment of objectives and criteria, synthesis, analysis, construction, testing and evaluation.

In bio/biomedical engineering, design involves applying principles of engineering, biology, human physiology, chemistry, calculus-based physics, mathematics and statistics; in solving bio/biomedical engineering problems, including those associated with the interaction between living and non-living systems; analyzing, modeling, designing, and realizing bio/biomedical engineering devices, systems, components, or processes; and making measurements on and interpreting data from living systems.



Life cycle of medical devices

Q: What is meant by: customer-focused design and manufacturing partner?



MRI Design: 1. Magnetism and electromagnetism

Magnetic susceptibility

The **magnetic susceptibility** of a substance is the ability of external magnetic fields to affect the nuclei of a particular atom, and is related to the electron configurations of that atom. The nucleus of an atom, which is surrounded by paired electrons, is more protected from, and unaffected by, the external magnetic field than the nucleus of an atom with unpaired electrons. There are three types of magnetic susceptibility: **paramagnetism**, **diamagnetism** and **ferromagnetism**.

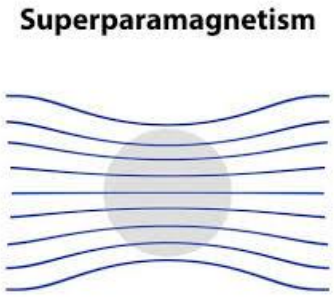
Paramagnetism

Paramagnetic substances contain unpaired electrons within the atom that induce a small magnetic field about themselves known as the **magnetic moment**. With no external magnetic field, these magnetic moments occur in a random pattern and cancel each other out. In the presence of an external magnetic field, paramagnetic substances align with the direction of the field and so the magnetic moments add together. Paramagnetic substances affect external magnetic fields in a positive way, resulting in a local increase in the magnetic field (Figure 1.1). An example of a paramagnetic substance is oxygen.

Figure 1.1 Paramagnetic properties.

Super-paramagnetism

Super-paramagnetic substances have a positive susceptibility that is greater than that exhibited by paramagnetic substances, but less than that of ferromagnetic materials. Examples of a super-paramagnetic substance are iron oxide contrast agents.



Diamagnetism

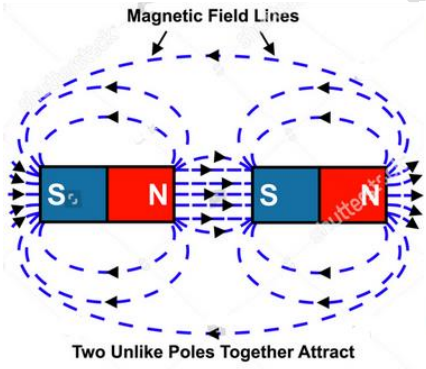
With no external magnetic field present, diamagnetic substances show no net magnetic moment, as the electron currents caused by their motions add to zero. When an external magnetic field is applied, diamagnetic substances show a small magnetic moment that opposes the applied field. Substances of this type are therefore slightly repelled by the magnetic field and have negative magnetic susceptibilities (Figure 1.2). Examples of diamagnetic substances include water and inert gasses.

Figure 1.2 Diamagnetic properties.

Ferromagnetism

When a ferromagnetic substance comes into contact with a magnetic field, the results are strong attraction and alignment. They retain their magnetization even when the external magnetic field has been removed. Ferromagnetic substances remain magnetic, are permanently magnetized and subsequently become permanent magnets. An example of a ferromagnetic substance is iron.

Magnets are **bipolar** as they have two poles, north and south. The magnetic field exerted by them produces magnetic field lines or lines of force running from the magnetic south to the north poles of the magnet (Figure 1.3). They are called **magnetic lines of flux.** The number of lines per unit area is called the **magnetic flux density**. The strength of the magnetic field, expressed by the notation **(B)** - or, in the case of more than one field, the primary field **(B0**) and the secondary field **(B_1)** - is measured in one of three units: **gauss (G)**, **kilogauss (kG)** and **tesla (T)**. If two magnets are brought close together, there are forces of attraction and repulsion between them depending on the orientation of their poles relative to each other. Like poles repel and opposite poles attract.



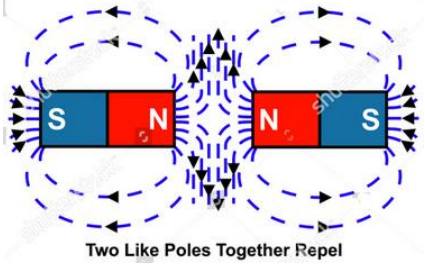


Figure 1.3 Ferromagnetic properties.

Electromagnetism

A magnetic field is generated by a moving charge (electrical current). The direction of the magnetic field can either be clockwise or counter-clockwise with respect to the direction of flow of the current. **Ampere's law** or **Fleming's right-hand rule** determines the magnitude and direction of the magnetic field due to a current (Figure 1.4).

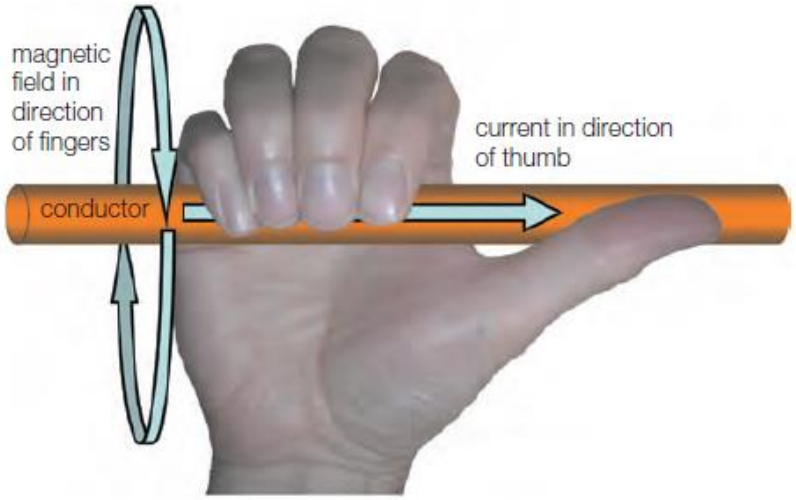


Figure 1.4 The right-hand thumb rule.

Just as moving electrical charge generates magnetic fields, changing magnetic fields generate electric currents. When a magnet is moved in and out of a closed circuit, an oscillating current is produced, which ceases the moment the magnet stops moving. Such a current is called an **induced electric current** (Figure 1.5).

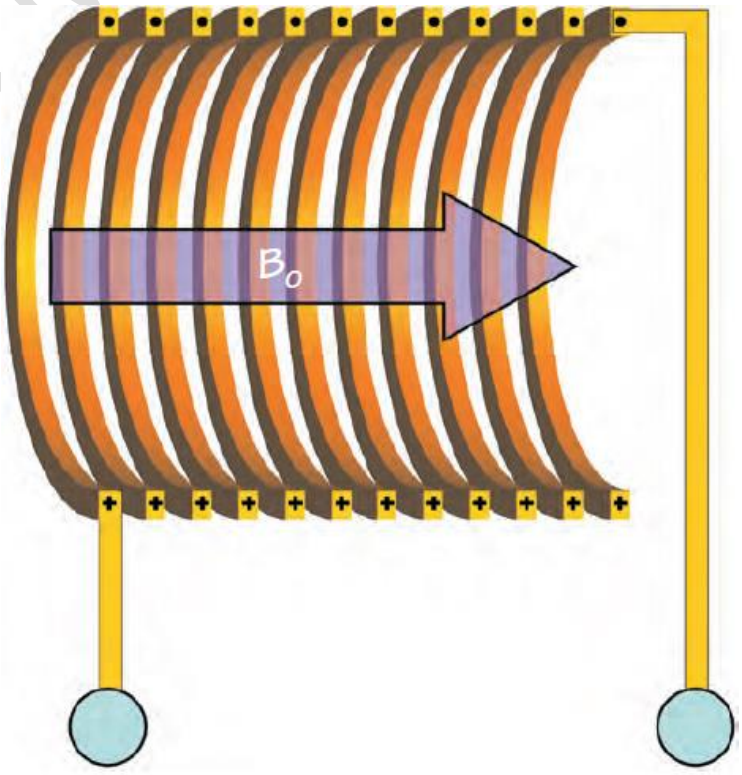


Figure 1.5 A simple electromagnet.

According to **Faraday's law of induction,** the change of magnetic flux through a closed circuit induces an **electromotive force (emf)** in the circuit. The emf is defined as the energy available from a unit of charge travelling once around a loop of wire. The emf drives a current in the circuit and is the result of a changing magnetic field inducing an electric field. The laws of electromagnetic induction (Faraday) state that the induced emf:

- is proportional to the rate of change of magnetic field and the area of the circuit;
- is proportional to the number of turns in a coil of wire Table 1.1);
- is in a direction so that it opposes the change in magnetic field which causes it (**Lenz's law**).

MRI Design: 2. Atomic structure

Atoms make up all matter in the universe and also therefore in the human body. Most of the human body (96%) is made up of just four elements: hydrogen, oxygen, carbon and nitrogen. Hydrogen is the most common element in the universe and in humans. The atom consists of the following particles:

Protons: in the nucleus, are positively charged

Neutrons: in the nucleus, have no charge

Electrons: orbit the nucleus, are negatively charged (Figure 2.1).

The following terms are used to characterize an atom:

· **Atomic number:** number of protons in the nucleus and determines the type of element the atoms make up.

· **Mass number:** sum of the neutrons and protons in the nucleus.

Atoms of the same element having a different mass number are called **isotopes**. In a stable atom the number of negatively charged electrons equals the number of positively charged protons. Atoms with a deficit or excess number of electrons are called **ions** and the process of removing electrons from the atom is called **ionization.**

Motion within the atom

There are three types of motion of particles in the atom:

· Negatively charged electrons spinning on their own axis.

· Negatively charged electrons orbiting the nucleus.

· Particles within the nucleus spinning on their own axes (Figure 2.1).

Each type of motion produces a magnetic field. In MRI we are concerned with the motion of particles within the nucleus and the nucleus itself.

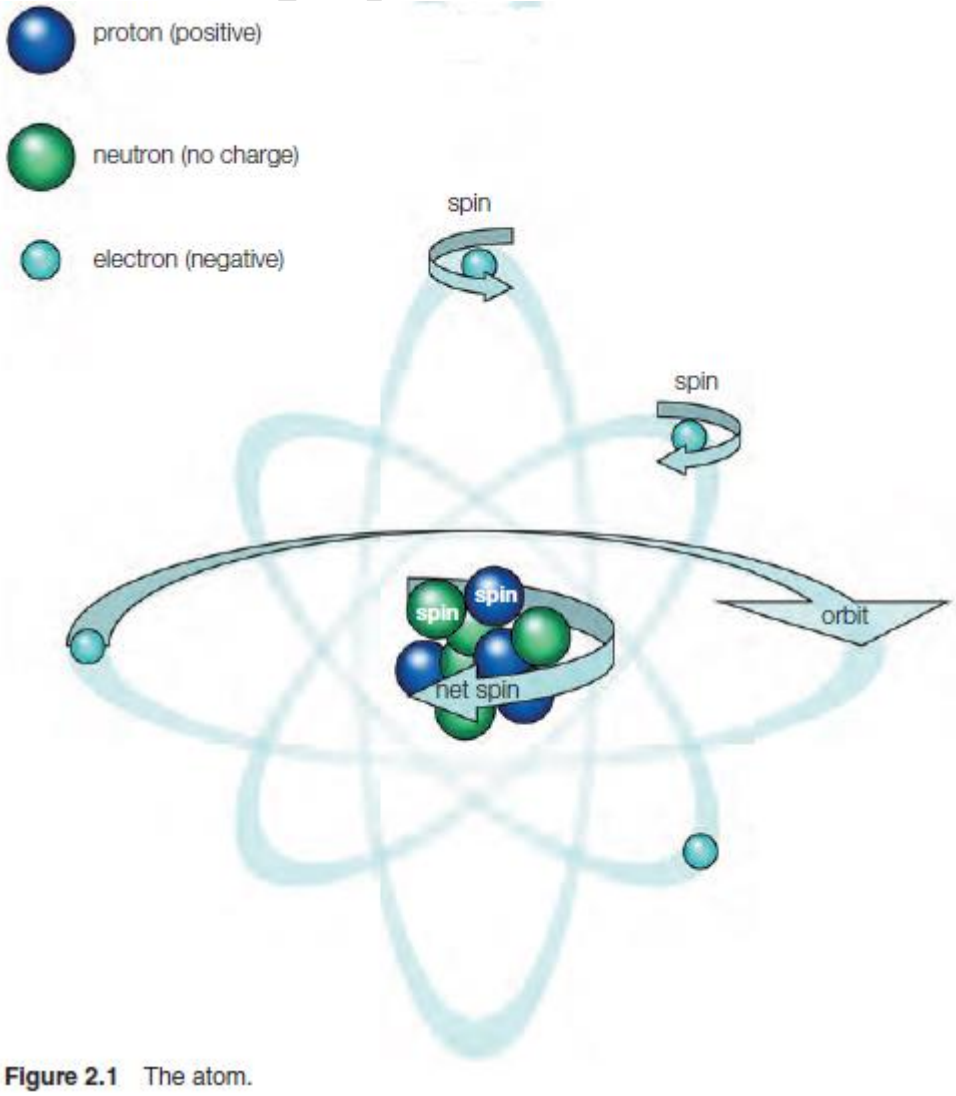


Figure 2.1 The atom.

MR active nuclei

Protons and neutrons spin about their own axis within the nucleus. The direction of spin is random, so that some particles spin clockwise and others anticlockwise.

- When a nucleus has an *even mass number*, the spins cancel each other out so the nucleus has *no net spin.*
- When a nucleus has an *odd mass number*, the spins do not cancel each other out and the *nucleus spins.*

As protons have charge, a nucleus with an odd mass number has a net charge as well as a net spin. Due to the laws of electromagnetic induction, a moving unbalanced charge induces a magnetic field around itself. The direction and size of the magnetic field are denoted by a magnetic moment (Figure 2.2). The total magnetic moment of the nucleus is the vector sum of all the magnetic moments of protons in the nucleus. The length of the arrow represents the magnitude of the magnetic moment. The direction of the arrow denotes the direction of alignment of the magnetic moment.

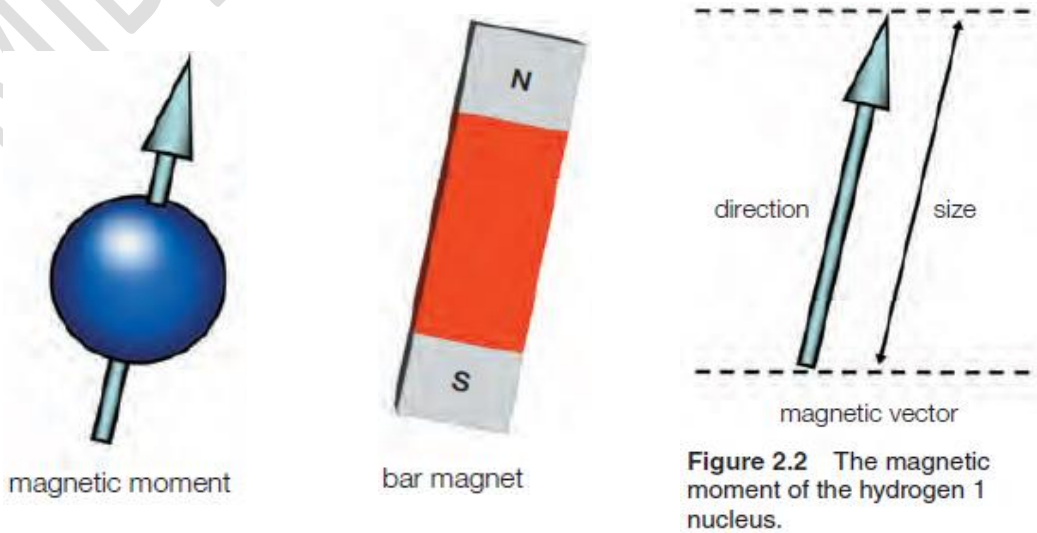


Figure 2.2 The magnetic moment of the hydrogen 1 nucleus.

Nuclei with an odd number of protons are said to be **MR active**. They act like tiny bar magnets. There are many types of elements that are MR active. They all have an odd mass number. The common MR active nuclei, together with their mass numbers and spin characteristics are shown in Table 2.1.

Table 2.1 Constants of selected MR active nuclei.

Element	Protons	Neutrons	Nuclear spin	% Natural abundance
^{1}H (protium)	1	0	1/2	99.985
^{13}C (carbon)	6	7	1/2	1.10
^{15}N (nitrogen)	7	8	1/2	0.366
^{17}O (oxygen)	8	9	5/2	0.038

*The isotope of hydrogen called **protium** is the MR active nucleus used in MRI*, as it has a mass and atomic number of 1. The nucleus of this isotope consists of a single proton and has no neutrons. It is used for MR imaging because:

- it is abundant in the human body (e.g. in fat and water);
- the solitary proton gives it a relatively large magnetic moment because there are no neutrons present in this type of nucleus.

Neutrons tend to decrease the relative size of the nuclear magnetic field, so if they are not present, the magnetic field is maximized (Table 2.1). In the rest of this course MR active nuclei, and specifically protium, are referred to as *spins*.

Key points.

- Hydrogen is the most abundant element in the human body.
- The nuclei that are available for MRI are those that exhibit a net spin (because their mass number is an odd number).
- As all nuclei contain at least one positively charged proton, those that also spin have a magnetic field induced around them.
- An arrow called a magnetic moment denotes the magnetic field of a nucleus.

MRI DESIGN: 3. Alignment

In a normal environment the magnetic moments of MR active nuclei (spins) point in a random direction, and produce no overall magnetic effect. When spins are placed in an external magnetic field, their magnetic moments line up with the magnetic field flux lines. This is called **alignment**. Alignment is described using two theories.

The classical theory

This uses the direction of the magnetic moments to illustrate alignment.

- **Parallel alignment**: alignment of magnetic moments in the *same* direction as the main field.
- **Anti-parallel alignment**: alignment of magnetic moments in the *opposite* direction to the main field (Figure 3.1).

At room temperature there are always more spins with their magnetic moments aligned parallel than anti-parallel. The net magnetism of the patient (termed the **net magnetization vector**; **NMV**) is therefore aligned parallel to the main field.

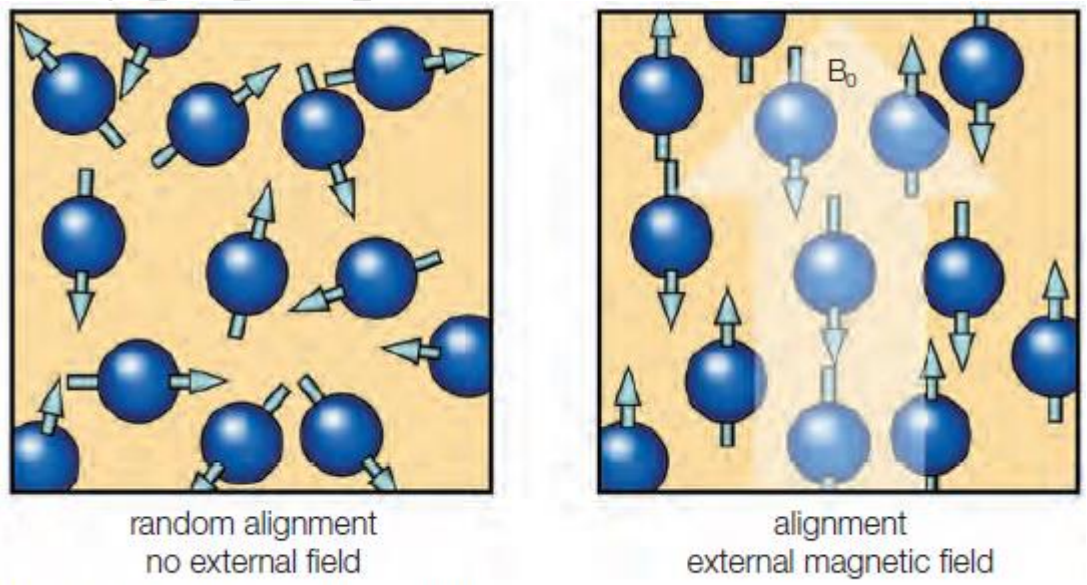


Figure 3.1 Alignment: classical theory.

The quantum theory

This uses the energy level of the spins to illustrate alignment. According to the quantum theory, protons of hydrogen nuclei interact with the external magnetic field of the scanner and cause a discrete number of energy states. For hydrogen nuclei there are only two possible energy states.

- **Spin-up** nuclei have low energy and do not have enough energy to oppose the main field. These are nuclei that align their magnetic moments parallel to the main field in the classical description.
- **Spin-down** nuclei have high energy and have enough energy to oppose the main field. These are nuclei that align their magnetic moments anti-parallel to the main field in the classical description.

The difference in energy between these two states is proportional to the strength of the external magnetic field (B_0). The magnetic moments of the spins actually align at an angle to B_0 due to the force of repulsion between B_0 and the magnetic moments.

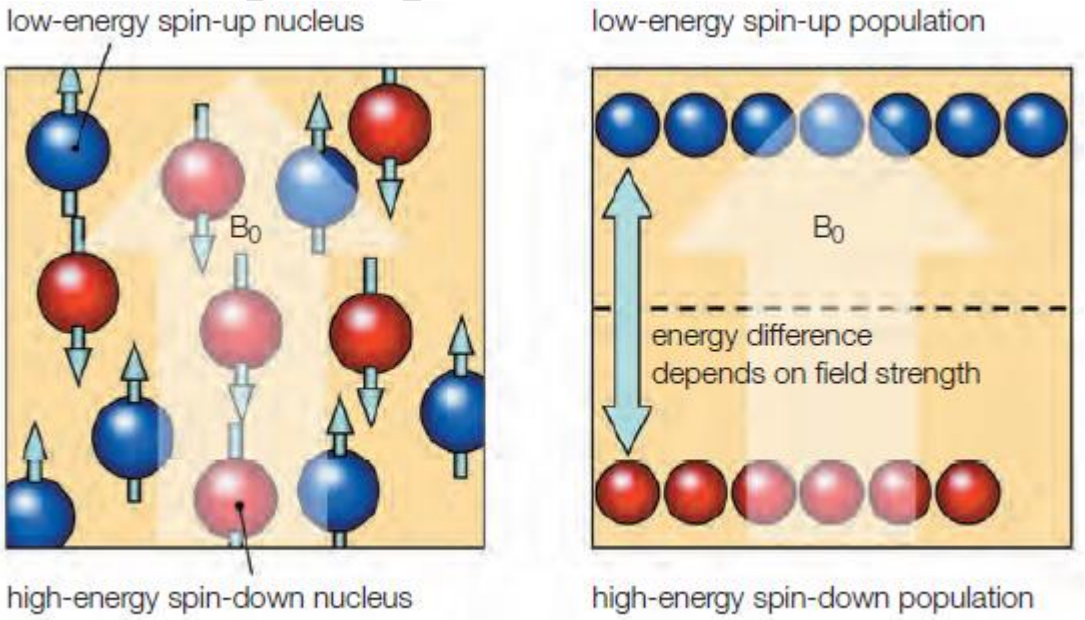


Figure 3.2 Alignment: quantum theory.

What do the quantum and classical theories tell us?

- Hydrogen only has two energy states: high or low. Therefore, the magnetic moments of hydrogen spins only align in the parallel or anti-parallel directions. The magnetic moments of hydrogen spins cannot orientate themselves in any other direction.
- The patient's temperature is an important factor that determines whether a spin is in the high or low energy population. In clinical imaging, thermal effects are discounted, as we assume the patient's temperature is the same inside and outside the magnetic field (thermal equilibrium).
- The magnetic moments of hydrogen spins are constantly changing their orientation because they are constantly moving between high and low energy states. The spins gain and lose energy and their magnetic moments therefore constantly alter their alignment relative to B_0.
- In thermal equilibrium, at any moment there are a greater proportion of spins with their magnetic moments aligned with the field than against it. Th is excess aligned with B_0 produces a net magnetic effect called the NMV that aligns with the main magnetic field (Figure 3.3).

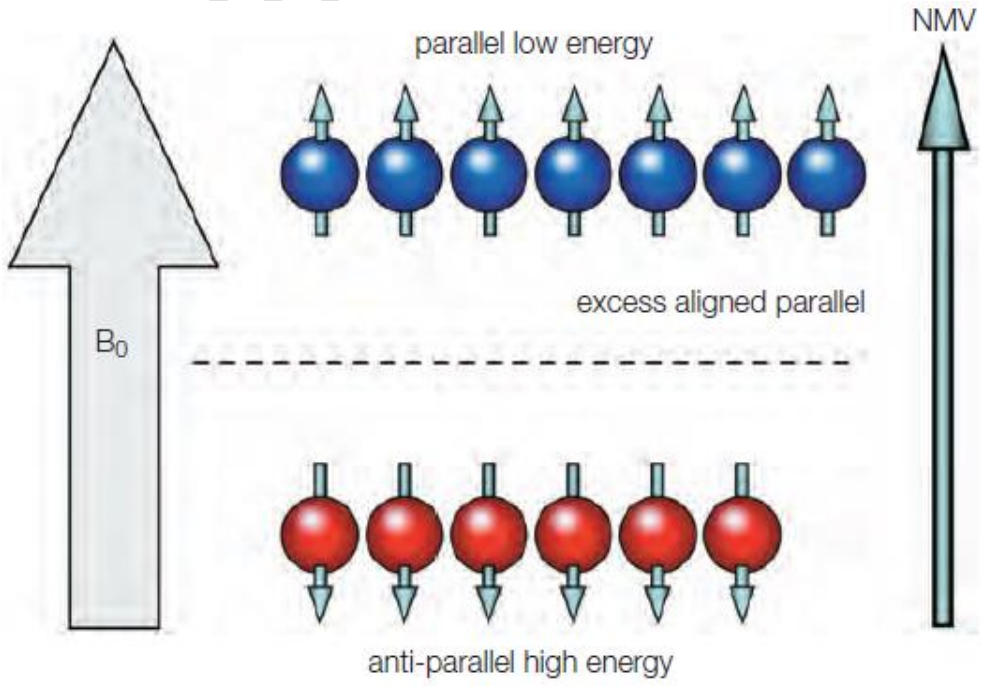


Figure 3.3 The net magnetization vector (NMV).

- As the magnitude of the external magnetic field increases, more magnetic moments line up in the parallel direction, because the amount of energy the spins must possess to align their magnetic moments in opposition to the stronger field and line up in the anti-parallel direction is increased. As the field strength increases, the low-energy population increases and the high energy population decreases. As a result, the NMV increases.

Key points.

- When placed in an external magnetic field, the magnetic moments of hydrogen either align in a spin-up, low-energy or spin-down, high-energy orientation.
- At thermal equilibrium, there are more spin-up, low-energy than spin-down, high-energy spins, so the net magnetization of the patient (NMV) is orientated in the same direction as B_0.
- The difference in energy between these populations is determined by the strength of B_0.
- As B_0 increases the energy difference between the two populations also increases, as the number of spin-up, low-energy spins increases relative to the number of spin-down, high-energy spins.
- The signal to noise ratio (SNR) increases at higher values of B_0.

MRI DESIGN: 4. Precession

Every MR active nucleus is spinning on its own axis. The magnetic field exerts a torque on the magnetic moments of all MR active nuclei, causing a secondary spin (Figure 4.1). This spin is called **precession** and causes the magnetic moments of all MR active nuclei (spin up and spin down) to describe a circular path around B_0 (Figure 4.2). The speed at which the magnetic moments spin about the external magnetic field is called the **precessional frequency**.

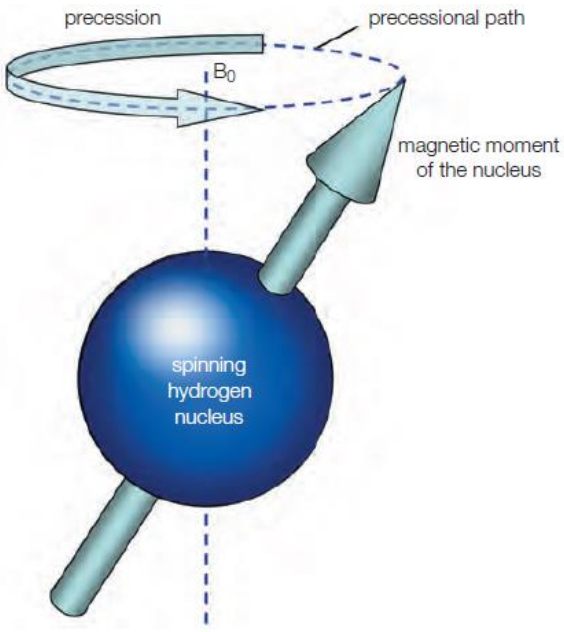


Figure 4.1 Precession.

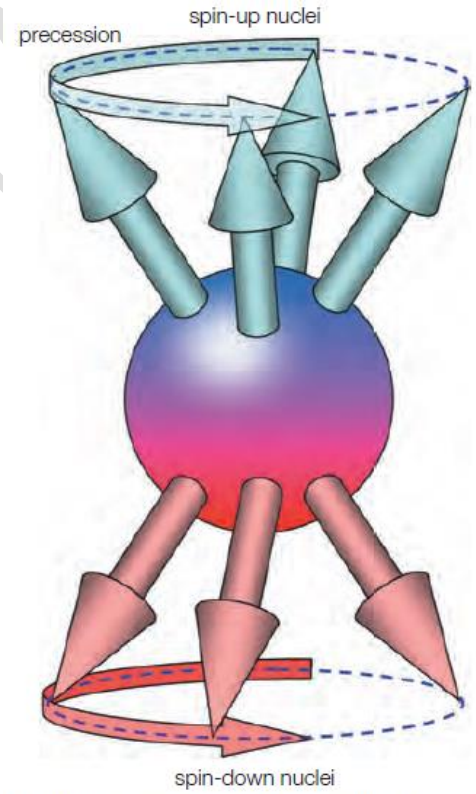


Figure 4.2 Precession of the spin-up and spin-down populations.

Precessional (Larmor) frequency

The **Larmor equation** is used to calculate the frequency or speed of precession for the magnetic moments of a specific nucleus in a specific magnetic field strength. The Larmor equation is simply stated as follows:

$$\omega_0 = \gamma B_0$$

- The precessional frequency is denoted by ω_0 and expressed in megahertz (MHz).
- The strength of the external field is expressed in tesla (T) and denoted by the symbol B_0 (Table 4.1).
- The **gyromagnetic ratio** is the precessional frequency of the magnetic moments of a specific nucleus at 1T and has units of MHz/T. It is denoted by γ. As it is a constant of proportionality, the precessional frequency or Larmor frequency is proportional to the strength of the external field and can be calculated for any type of MR active nucleus and field strength (Table 4.2).

Table 4.1 Common equations of precession.

Equations		
$\omega_0 = \gamma B_0/2\pi$ *simplified to* $\omega_0 = \gamma B_0$	ω_0 is the precessional of Larmor frequency (MHz) γ is the gyromagnetic ratio (MHz/T) B_0 is the strength of the external magnetic field (T)	This is the Larmor equation. The 2π function enables the conversion of ω_0 from angular to cyclical frequency. As γ is a constant, for a given MR active nucleus ω_0 is proportional to B_0.

^{15}N (nitrogen)	1/2	4.3173	6.4759
^{17}O (oxygen)	5/2	5.7743	8.6614

The precessional frequencies of the magnetic moments of hydrogen spins (gyromagnetic ratio 42.57 MHz/T) commonly found in clinical MRI are:

- 21.285 MHz at 0.5 T
- 42.57 MHz at 1 T
- 63.86 MHz at 1.5 T (Table 4.2)

Table 4.2 Spin characteristics of selected MR active nuclei.

Element	Nuclear spin	Gyromagnetic ratio (MHz/T)	Larmor frequency at 1.5T (MHz)
^{1}H (protium)	1/2	42.5774	63.8646
^{13}C (carbon)	1/2	10.7084	16.0621
^{15}N (nitrogen)	1/2	4.3173	6.4759
^{17}O (oxygen)	5/2	5.7743	8.6614

The precessional frequency corresponds to the range of frequencies in the electromagnetic spectrum of **radiowaves** (Figure 4.3). Therefore, the magnetic moments of hydrogen spins precess at a relatively low radio frequency (RF) compared to other types of electromagnetic radiation. This is why from the perspective of the energies used, MRI is thought to be safe. RF energy is not sufficiently energetic to ionization.

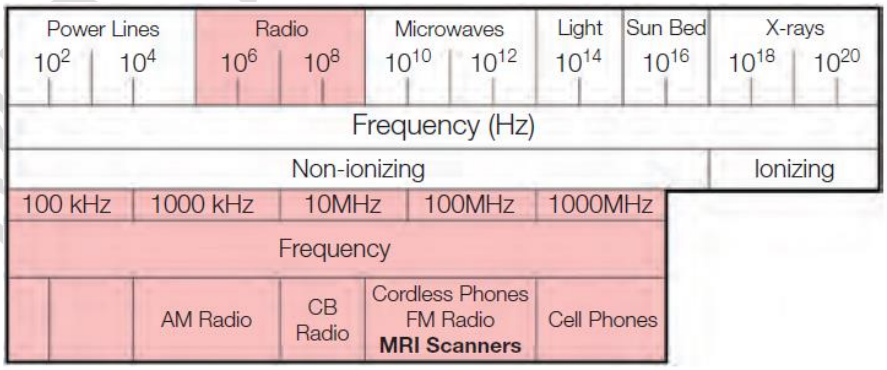


Figure 4.3 The electromagnetic spectrum.

Precessional phase

Phase refers to the position of the magnetic moments of spins on their precessional path at any moment in time. Its units are radians. A magnetic moment travels through 360 radians during one rotation. In this context, frequency is the rate of change phase of magnetic moments; that is, it is a measure of how quickly the phase position of a magnetic moment changes over time.

In MRI, we are particularly interested in the relative phase position of all the magnetic moments of hydrogen spins in the tissue we are imaging.

- **Out of phase** or **incoherent** means that the magnetic moments of hydrogen spins are at different places on the precessional path at a moment in time.
- **In phase** or **coherent** means that the magnetic moments of hydrogen spins are at the same place on the precessional path at a moment in time (Figure 4.4).

At rest (when the patient is simply placed inside the magnetic field and exposed to B_0), the magnetic moments of the hydrogen spins are out of phase with each other and therefore the NMV does not precess. The key points of this chapter are summarized in Table 4.3.

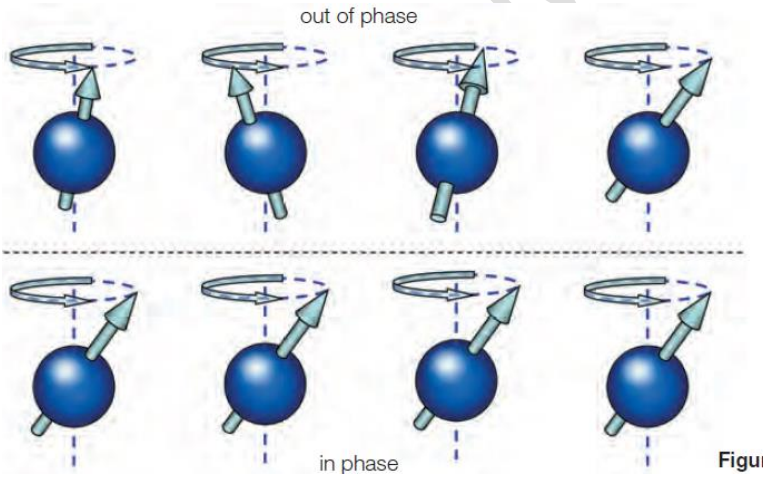


Figure 4.4 Coherent and incoherent phase positions

KEY POINTS

✓ The magnetic moments of all the spins precess around B_0 at the Larmor frequency that is proportional to B_0 for a given MR active nucleus. Frequency

therefore refers to how fast the magnetic moments of spins are precessing and is measured in MHz in MRI.

✓ For field strengths used in clinical imaging, the Larmor frequency of hydrogen is in the radiofrequency band of the electromagnetic spectrum.

✓ At rest the magnetic moments of the spins are out of phase with each other.

Lecture Two

MRI DESIGN: 5. Resonance and signal generation

Resonance is an energy transition that occurs when an object is subjected to a frequency the same as its own. Resonance is induced by applying a **radiofrequency (RF) pulse**:

- at the same frequency as the precessing magnetic moment's hydrogen spins;
- at 90° to B_0.

This causes the hydrogen spins to resonate (receive energy from the RF pulse), whereas other types of MR active nuclei do not resonate. As their gyromagnetic ratios are different from that of hydrogen, their precessional frequencies are also different from that of hydrogen. They only resonate if RF at their specific precessional frequency is applied. As RF is only applied at the same frequency as the precessional frequency of hydrogen, only hydrogen spins resonate. The other types of MR active nuclei do not. Two things happen to the hydrogen spins at resonance: *energy absorption and phase coherence.*

Energy absorption

The energy and frequency of electromagnetic radiation (including RF) are related to each other and, consequently, the frequency required to cause resonance is related to the difference in energy between the high- and low-energy populations and thus the strength of B_0 (Table 5.1). The spin-up, low-energy hydrogen spins absorb energy from the RF pulse (excitation pulse) and move into the high-energy population. At the same time, the spin-down, high-energy spins give energy away and return to the low-energy state. As there are more low-energy spins, the net effect is of energy absorption. This absorption of applied RF energy at 90° to B_0 causes a *net increase* in the number of high-energy, spin-down nuclei compared to the pre-resonant state (Figure 5.1).

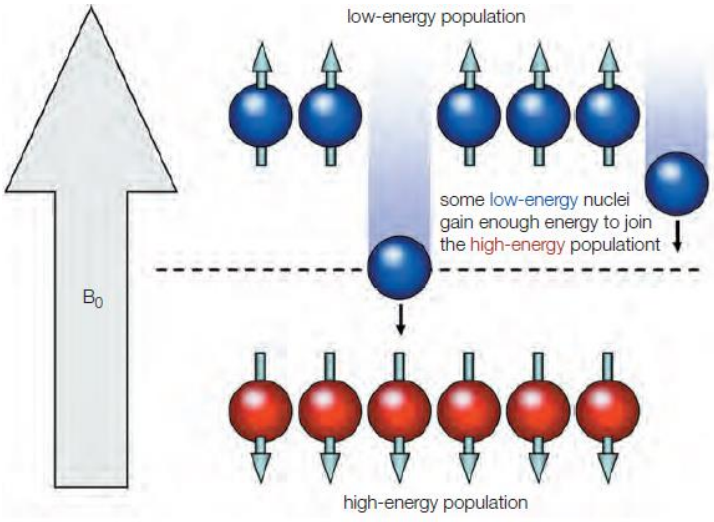


Figure 5.1 Energy transfer during excitation.

If just the right amount of energy is applied, the number of nuclei in the spin-up position equals the number in the spin-down position. As a result, the NMV (which represents the balance between spin-up and spin-down nuclei) lies in a plane at 90° to the external field (the **transverse plane**) as the net magnetization lies between the two energy states. As the NMV has been moved through 90° from B_0, it has a **flip or tip angle** of 90° (Figure 5.2).

Table 5.1 Common equations of resonance.

Equations		
$E = h\,\omega_0$	E is the energy of a photon (Joules, J) h is Planck's constant (6.626×10^{34} J/s) ω_0 is the frequency of an electromagnetic wave (Hz)	Planck's constant relates the energy of a photon of electromagnetic radiation to its frequency. Photons are both particles that possess energy and at the same time behave like waves that have frequency (wave particle duality).
$\Delta E = h\omega_0 = h\gamma B_0$	ΔE is the energy difference between the spin-up and spin-down populations h is Planck's constant (6.626×10^{34} J/s) ω_0 is the precessional or Larmor frequency (MHz) γ is the gyromagnetic ratio (MHz/T).	This equation shows that when the energy of the photon matches the energy difference between the spin-up and spin-down populations, energy absorption occurs. This is proportional to the magnetic field strength B_0.

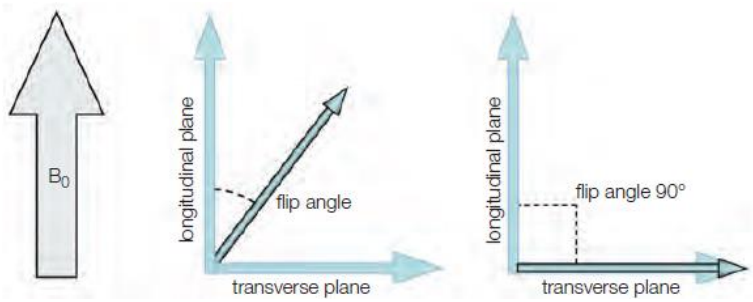


Figure 5.2 The flip angle. What flip angle gives maximum transverse magnetization?

Phase coherence

The magnetic moments of the spins move into phase with each other. As the magnetic moments of the spins are in phase in both the spin-up and spin-down positions and the spin-up nuclei are in phase with the spin-down nuclei, the net effect is one of precession, so the NMV precesses in the transverse plane at the Larmor frequency.

Important note:

When a patient is placed in the magnet and is scanned, hydrogen spins do not move. Spins are not flipped onto their sides in the transverse plane and neither are their magnetic moments. Only the magnetic moments of the spins move, aligning either with or against B_0. This is because hydrogen can only have two energy states, high or low. It is the NMV that lies in the transverse plane, not the magnetic moments, nor the spins themselves.

The MR signal:

A receiver coil is situated in the transverse plane. As the NMV rotates around the transverse plane as a result of resonance, it passes across the receiver coil, inducing a voltage in it. This voltage is the **MR signal** (Figure 5.3).

After a short period of time the RF pulse is removed. The signal induced in the receiver coil begins to decrease. This is because the in-phase component of the NMV in the transverse plane, which is passing across the receiver coil, begins to decrease as an increasingly higher proportion of spins become out of phase

with each other. The amplitude of the voltage induced in the receiver coil therefore decreases. This is called **free induction decay** or **FID**:

- 'free' because of the absence of the RF pulse;
- 'induction decay' because of the decay of the induced signal in the receiver coil.

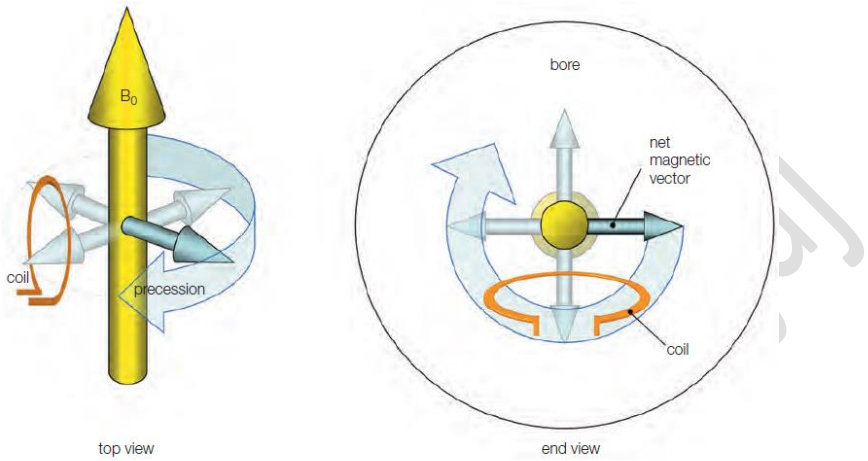


Figure 5.3 Generation of the MR signal. Why would you expect it to be alternating?

KEY POINTS

✓ The application of RF energy at the Larmor frequency causes a net absorption of energy (excitation) and changes the balance between the number of spins in the low- and high-energy populations.

✓ The orientation of the NMV to B_0 depends on this balance. If there are a similar number of spins in each population, the NMV lies in a plane at 90° to B_0 (transverse plane).

✓ Resonance also causes the magnetic moments of all spins to precess in phase. The result is coherent transverse magnetization that precesses in the transverse plane at the Larmor frequency.

✓ If a receiver coil (conductor) is placed in the transverse plane, the movement of the rotating coherent transverse magnetization causes a voltage to be induced in the coil.

✓ When the RF excitation pulse is removed, the magnetic moments of all spins dephase and produce a FID.

MRI DESIGN: 6. Contrast Mechanisms

What is contrast?

An image has contrast if there are areas of high signal (white on the image), as well as areas of low signal (dark on the image). Some areas have an intermediate signal (shades of grey, between white and black). The NMV can be separated into the individual vectors of the tissues present in the patient, such as fat, cerebrospinal fluid (CSF), grey matter and white matter (Figure 6.1). The contrast to noise ratio (CNR) is an important image quality parameter and relates to the difference in signal between two adjacent areas. Images demonstrating a good CNR contain large differences in signal intensity. Images demonstrating poor CNR do not.

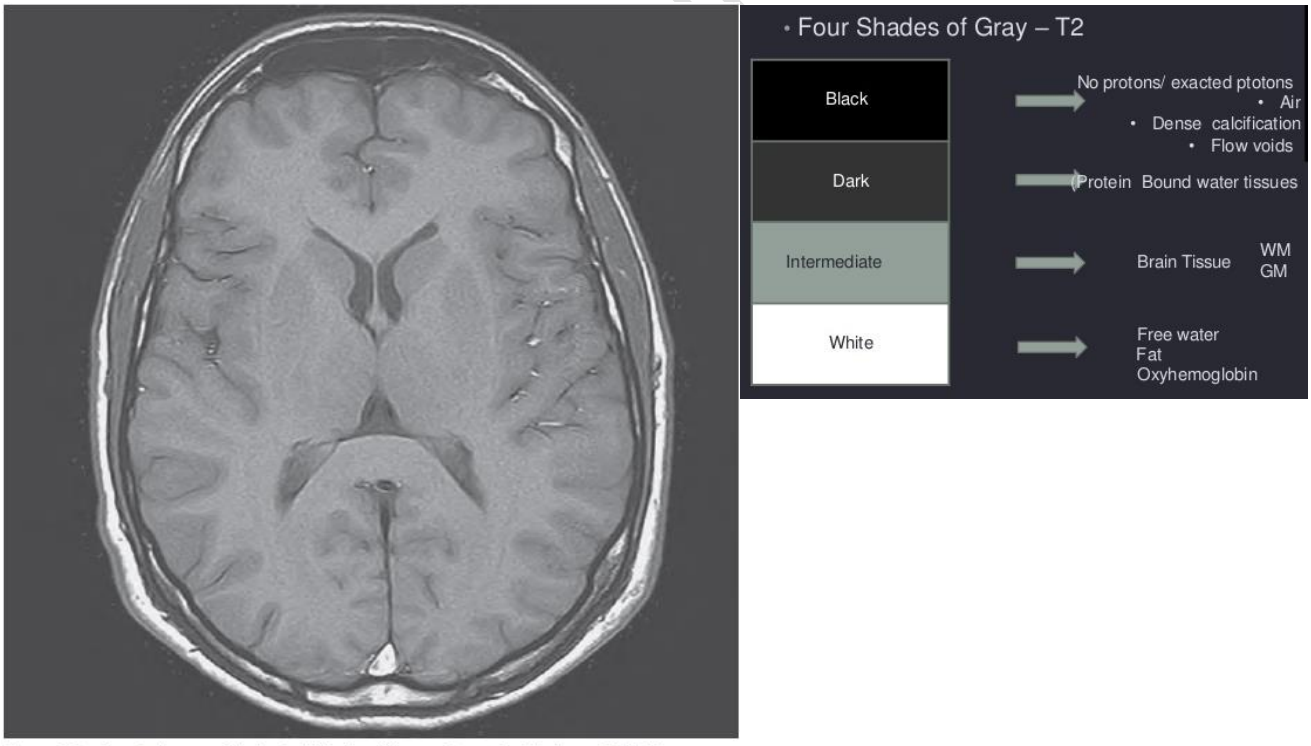


Figure 6.1 An axial image of the brain. Note the difference in contrast between CSF, fat, grey and white matter.

The determinant of signal intensity in MRI is the magnitude of precessing coherent transverse magnetization that cuts through the windings of the receiver coil when the signal is measured. This is because the amplitude of voltage induced in a conductor depends on the amplitude of the transverse magnetic field.

A tissue has a *high signal (white, hyper-intense)* if it has a *large transverse component of magnetization* when the signal is measured. If there is a large component of transverse magnetization, the amplitude of the magnetization that cuts the coil is large, and the signal induced in the coil is also large.

A tissue has a *low signal (black, hypo-intense)* if it has a *small transverse component of magnetization* when the signal is measured. If there is a small component of transverse magnetization, the amplitude of the magnetization that cuts the coil is small, and the signal induced in the coil is also small.

A tissue has an intermediate signal (grey, iso-intense) if it has a medium transverse component of magnetization when the signal is measured.

Image contrast is determined by the difference in signal intensity between tissues. This is controlled by various parameters (Table 6.1).

Table 6.1 Common equations of contrast mechanisms.

Equations		
$SI = PD\, e^{-TE/T2} (1-e^{-TR/T1})$	SI is the signal intensity in a tissue PD is proton density TE is the echo time (ms) T2 is the T2 relaxation time of the tissue (ms) TR is the repetition time (ms) T1 is the T1 relaxation time in the tissue (ms)	This equation shows why the signal intensity from a tissue depends on intrinsic and extrinsic contrast parameters. In gradient echo sequences the flip angle is added to this equation and T2 is referred to as T2*

Extrinsic contrast parameters

These parameters are controlled by the operator. They are:

• **Repetition time (TR)**: This is the time from the application of one RF pulse to the application of the next for a particular slice. It is measured in milliseconds (ms). The TR affects the length of a relaxation period in a particular slice after the application of one RF excitation pulse to the beginning of the next (see Figure 6.2).

• **Time to echo (TE)**: This is the time between an RF excitation pulse and the collection of the signal. The TE affects the length of the relaxation period after the removal of an RF excitation pulse and the peak of the signal received in the receiver coil. It is also measured in ms (Figure 6.2).

• **Flip angle**: This is the angle through which the NMV is moved as a result of an RF excitation pulse (Figure 5.2).

• **Turbo-factor (TF)** or **echo train length (ETL)**: the number of echoes acquired after each excitation.

• **Time from inversion (TI)**: the time between the 180° inverting pulse and the 90°-pulse

• **'b' value**: parameter that defines gradient strength and duration, hence determines the degree of diffusion weighting.

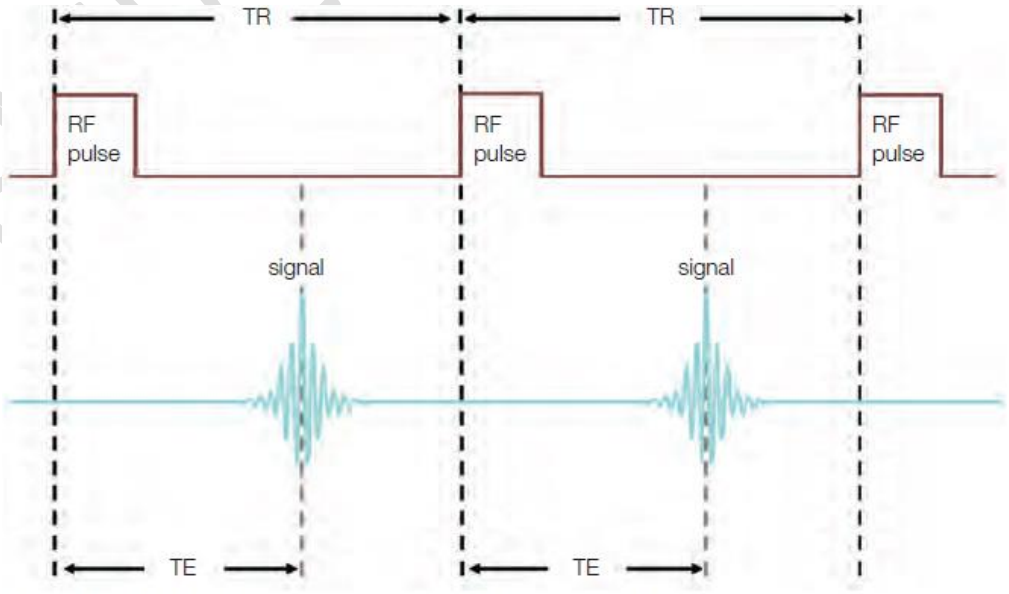


Figure 6.2 A basic pulse sequence showing TR and TE intervals.

Intrinsic contrast mechanisms

These parameters are inherent to the tissue and are not controlled by the operator. They are:

- T1 recovery time.
- T2 decay time.
- proton density.
- flow: the quantity of blood (cm^3/sec) passes a point at certain time.
- apparent diffusion coefficient (ADC): is a measure of the magnitude of diffusion (of water molecules) within tissue, and is commonly clinically calculated using MRI with diffusion-weighted imaging (DWI)

The composition of fat and water

All substances possess molecules that are constantly in motion. This molecular motion is made up of rotational and transitional movements and is called **Brownian motion**. The faster the molecular motion, the more difficult it is for a substance to release energy to its surroundings.

Fat comprises hydrogen atoms mainly linked to carbon that make up large molecules. The large molecules in fat are closely packed together and have a slow rate of molecular motion due to the inertia of the large molecules. They also have low inherent energy which means they are able to absorb energy efficiently.

Water comprises hydrogen atoms linked to oxygen. It consists of small molecules that are spaced far apart and have a high rate of molecular motion. They have a high inherent energy which means they are not able to absorb energy efficiently.

Because of these differences, tissues that contain fat and water produce different image contrast. This is because there are different **relaxation** rates in each tissue.

KEY POINTS

✓ The Contrast between tissues occurs because there is a different signal intensity between different tissues.

✓ Signal intensity depends on the amplitude of the signal.

✓ Resonance also causes the magnetic moments of all spins to precess in phase. The result is coherent transverse magnetization that precesses in the transverse plane at the Larmor frequency.

✓ If a receiver coil (conductor) is placed in the transverse plane, the movement of the rotating coherent transverse magnetization causes a voltage to be induced in the coil.

✓ When the RF excitation pulse is removed, the magnetic moments of all spins dephase and produce a FID.

Note:

Precession: is the slow movement of the axis of a spinning body around another axis due to a torque acting to change the direction of the first axis.

Lecture Three

MRI DESIGN: 7. Relaxation Mechanisms

Relaxation is a general term that refers to a loss of energy. In MRI this is energy delivered to the spins via excitation. After the RF excitation pulse has been applied and resonance and the desired flip angle achieved, the RF pulse is removed. The signal induced in the receiver coil begins to decrease. This is because the coherent component of NMV in the transverse plane, which is passing across the receiver coil, begins to gradually decrease as an increasingly higher proportion of spins become out of phase with each other. The amplitude of the voltage induced in the receiver coil therefore gradually decreases. This is called **free induction decay** or **FID**. The NMV in the transverse plane decreases due to:

- relaxation processes;
- field inhomogeneities and susceptibility effects.

The cumulative dephasing of spin-spin interactions and inhomogeneities is called T2* decay (Table 7.1).

Table 7.1 Common equations of relaxation mechanisms.

Equations		
$1/T2^* = 1/T2 + 1/2\ \gamma\Delta B_0$	T2 and T2* are the tissues' T2 and T2* relaxation times (ms) γ is the gyromagnetic ratio (MHz/T) ΔB_0 is the variation in magnetic field (inhomogeneities) (parts per million, ppm)	This equation shows how T2 and T2* are related to each other. Poor field inhomogeneities result in T2* being much shorter than T2, and fast decaying signal.

Relaxation processes

The magnetization in each tissue relaxes at different rates. This is one of the factors that create image contrast. The withdrawal of the RF produces several effects:

• Spins emit energy absorbed from the RF pulse through a process known as **spin-lattice energy transfer** and shift their magnetic moments from the *high-energy state* to the *low-energy state*. The NMV recovers and realigns to B_0. This relaxation process is called ***T1 recovery***.

• The magnetic moments of the spins lose precessional coherence or dephase, and the *NMV decays in the transverse plane*. The dephasing relaxation process is called ***T2 decay***.

The magnetic moments of the spins lose their coherence by:

• interactions of the intrinsic magnetic fields of adjacent nuclei (**spin-spin**) causing **T2 decay**;

• **inhomogeneities** of the external magnetic field.

Field inhomogeneities

Despite attempts to make the main magnetic field as uniform as possible via shimming, inhomogeneities of the external magnetic field are inevitable and slightly alter the magnitude of B_0; that is, some small areas of the field have a magnetic field strength of slightly more or less than the main field strength.

Due to the Larmor equation, the precessional frequency of the magnetic moment of a spin is proportional to B_0, so spins that pass through inhomogeneities experience a precessional frequency and phase change, and the resulting signal decays exponentially. This results in a change in dephasing of the transverse magnetization due to a loss in phase coherence (Figure 7.1). The resulting signal decays exponentially and is called an FID.

T2 decay is irreversible because spin-spin interactions occur at the atomic or molecular level. However, T2* decay, particularly caused by field inhomogeneity, can be compensated for and is desirable (Table 7.1). In order to produce images where T2 contrast is visualized, ideally there must be a mechanism to rephase spins and compensate for magnetic field inhomogeneities. *Pulse sequences* are mechanisms that perform this function.

A **pulse sequence** is defined as a series of RF pulses, gradient applications and intervening time periods. They enable control of the way in which the system applies RF pulses and gradients. By selecting the intervening time periods, image weighting is controlled. Pulse sequences are required because, without a mechanism of refocusing spins, there is insufficient signal to produce an image. This is because dephasing occurs almost immediately after the RF excitation pulse has been removed.

The main purposes of pulse sequences are to:

- rephase spins and remove inhomogeneity effects and therefore produce a signal or echo that contains information about the T2 decay characteristics of tissue alone;
- enable manipulation of the TE and TR to produce different types of contrast.

Spins are rephased by using (Table 7.2):

- a 180° RF pulse (used in all spin echo sequences);
- a gradient (used in all gradient echo sequences).

Table 7.2 Pulse sequences and their rephasing mechanisms.

Use RF pulses to rephase spins	Use gradients to rephase spins
Conventional spin echo	Coherent gradient echo
Fast or turbo spin echo	Incoherent gradient echo
Inversion recovery	Steady-state free precession
STIR	Balanced gradient echo
FLAIR	EPI

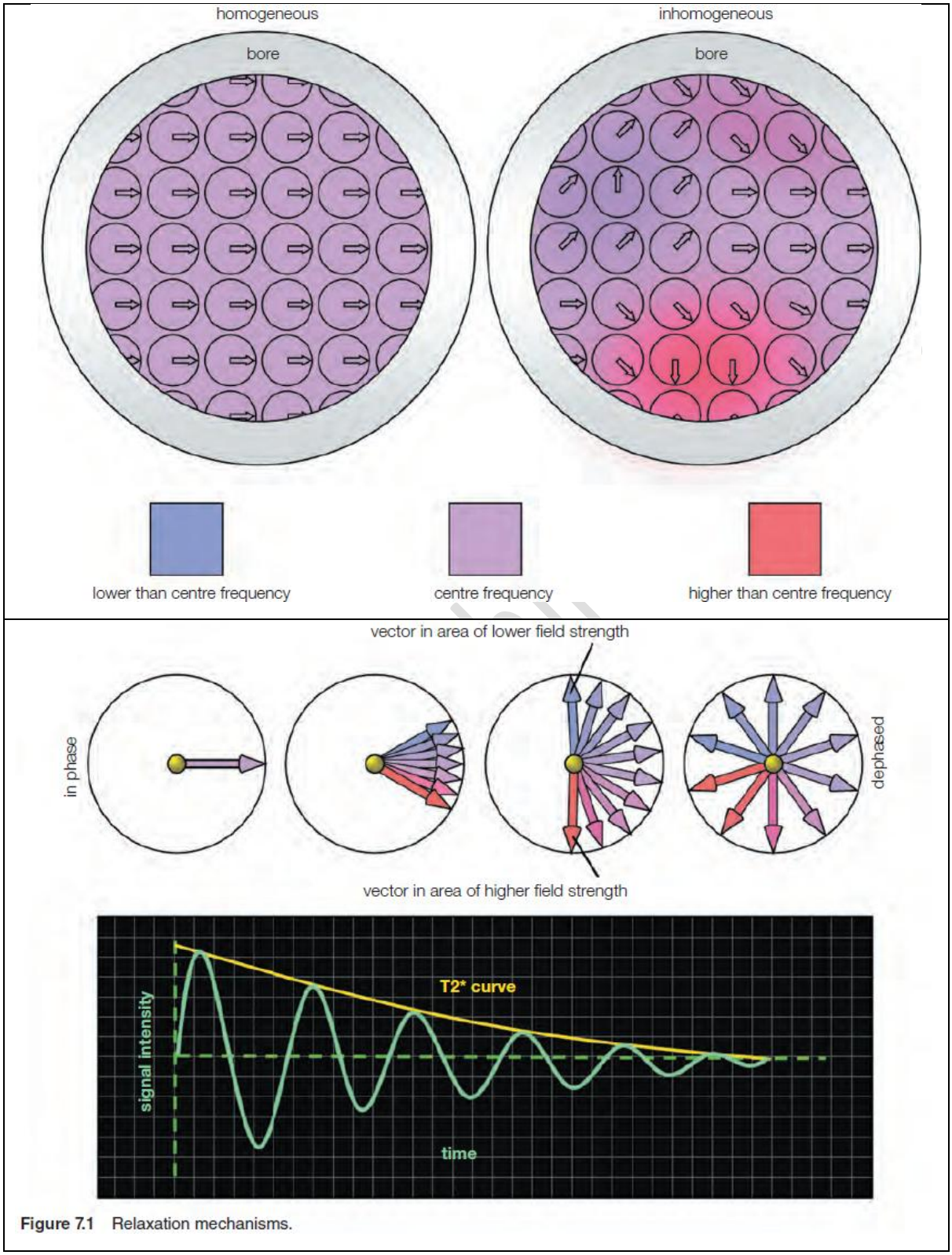


Figure 7.1 Relaxation mechanisms.

Q: What is RF pulse in MRI? And what is gradient in MRI?

KEY POINTS

✓ Relaxation is a general term that refers to a loss of energy. In MRI, this is energy delivered to the spins via excitation.

✓ Relaxation and inhomogeneities result in a FID signal.

✓ Spin lattice energy transfer is a relaxation process where spins give up the energy absorbed through excitation to the surrounding molecular lattice of the tissue. It is called T1 recovery.

✓ T2 decay is an irreversible loss of phase coherence due to spin-spin interactions on an atomic and molecular level.

✓ Pulse sequences are mechanisms that permit refocusing of spins so that images can be acquired with different types of contrast.

MRI DESIGN: 8. T1 Recovery

T1 recovery is caused by the exchange of energy from spins to their surrounding environment or lattice. It is called **spin lattice energy transfer**. As the spins dissipate their energy their magnetic moments relax or return to B_0; that is, they regain their longitudinal magnetization. Th e rate at which this occurs is an exponential process and it takes place at different rates in different tissues.

The T1 recovery time of a particular tissue is an intrinsic contrast parameter that is inherent to the tissue being imaged. It is a constant for a particular tissue and is defined as the time it takes for 63% of the longitudinal magnetization to recover in that tissue (Figure 8.1 and Table 8.1). The period of time during which this occurs is the time between one excitation pulse and the next or the **TR**. The TR therefore determines how much T1 recovery occurs in a particular tissue.

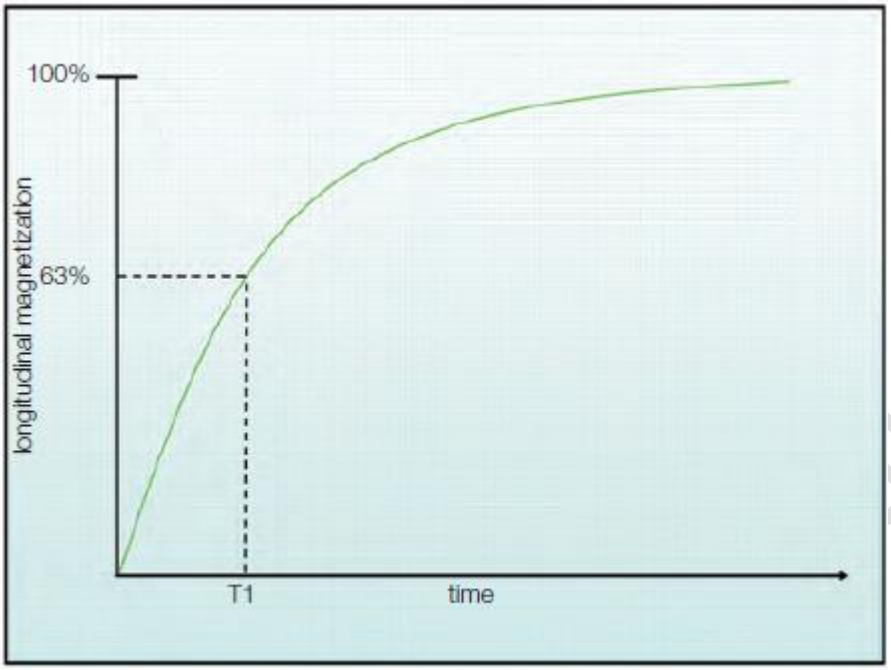


Figure 8.1 The T1 recovery curve.

Table 8.1 Equations of T1 recovery.

Equations		
$Mz_t = Mz\,(1-e^{-t/T1})$	Mz_t is the amount of longitudinal magnetization at time t after the removal of the excitation pulse. Mz is full longitudinal magnetization. T1 is the T1 recovery time (ms) and is the time taken to increase the longitudinal magnetization by a factor of e.	This equation plots the size of the recovering NMV as a function of time after the removal of the excitation pulse and the T1 recovery time. When t=T1, 63% of the longitudinal magnetization recovers. When t=2xT1, 86% recovers and when t=3xT1, 95% recovers. It usually takes between 3 and 5 T1 recovery times for full recovery to occur.

T1 recovery in fat

T1 relaxation occurs as a result of spins exchanging the energy given to them by the RF pulse to their surrounding environment. The efficiency of this process determines the T1 recovery time of the tissue in which they are situated. Due to the fact that fat is able to absorb energy quickly, the *T1 recovery time of fat is very short*; that is, spins dispose of their energy to the surrounding fat tissue and return to B_0 in a short time (Figure 8.2; Table 8.2).

Table 8.2 T1 relaxation times of brain tissue at 1 T.

Tissue	T1 relaxation time (ms)
Water	2500
Fat	200
CSF	2000
White matter	500

T1 recovery in water

Water is very inefficient at receiving energy from spins. *The T1 recovery time of water is therefore quite long*; that is, spins take a lot longer to dispose of their energy to the surrounding water tissue and return to B_0 (Figure 8.3; Table 8.2). In addition, the efficiency of spin lattice energy transfer depends on how closely molecular motion of the molecules matches the Larmor frequency. If there is a good match between the rate of molecular tumbling and the precessional frequency of spins, energy is efficiently exchanged between hydrogen and the surrounding molecular lattice. The Larmor frequency is relatively slow and therefore fat is much better at this type of energy exchange than water, whose molecular motion is much faster than the Larmor frequency. *This is another reason why fat has a shorter T1 recovery time than water.*

T1 recovery is affected by the strength of the external magnetic field. The precessional frequency of spins within a tissue varies slightly, but efficient energy exchange due to molecular motion only occurs at the Larmor frequency. The Larmor frequency is proportional to B_0 and therefore T1 recovery takes longer as B_0 increases, because there are fewer molecules moving at relaxation-causing frequencies.

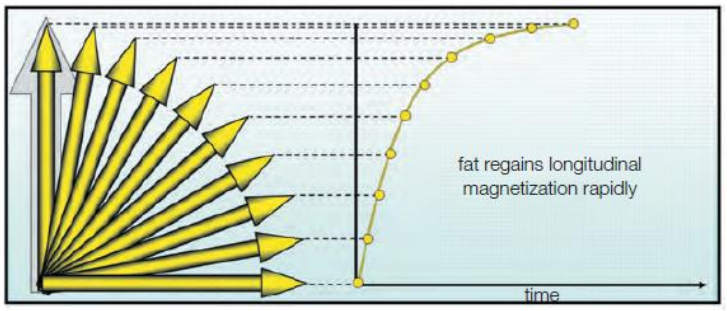


Figure 8.2 T1 recovery in fat.

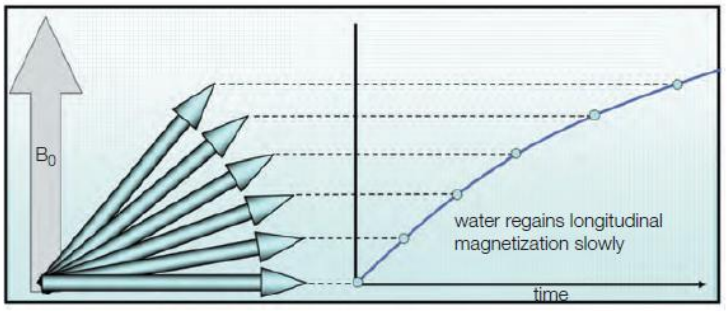


Figure 8.3 T1 recovery in water.

Control of T1 recovery

The TR controls how much of the NMV in fat or water recovers before the application of the next RF pulse. *A short TR* does not permit full longitudinal recovery in most tissues, so that there are different longitudinal components in fat and water. These different longitudinal components are converted to different transverse components after the next excitation pulse has been applied.

As the NMV does not recover completely to the positive longitudinal axis, they are pushed beyond the transverse plane by the succeeding 90° RF pulse. This is called **saturation**. When saturation occurs, there is a contrast difference between fat and water due to differences in their T1 recovery times (Figures 8.4 and 8.5).

A long TR allows full recovery of the longitudinal components in most tissues. There is no difference in the magnitude of their longitudinal components. There is no contrast difference between fat and water due to differences in T1 recovery times when using a long TR. Any differences seen in contrast are due to differences in the number of protons or **proton density** of each tissue. The proton density of a particular tissue is an intrinsic contrast parameter and is therefore inherent to the tissue being imaged.

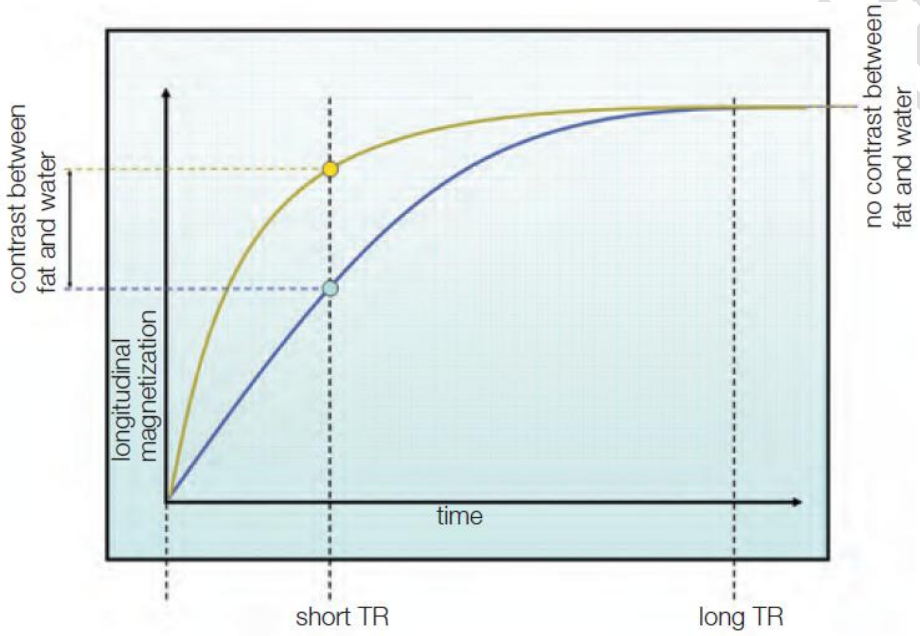


Figure 8.4 T1 recovery of fat and water.

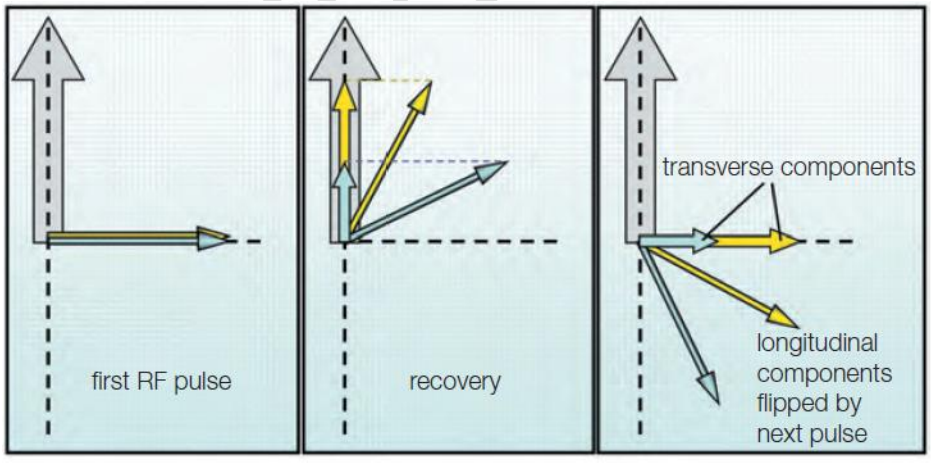


Figure 8.5 T1 contrast generation.

KEY POINTS

- ✓ Fat has a short T1 recovery time.
- ✓ Water has a long T1 recovery time.
- ✓ T1 recovery is caused by spin lattice energy transfer. The efficiency of this process depends on the inherent energy of the tissue and how well the rate of molecular tumbling matches Larmor.
- ✓ T1 recovery times are dependent on magnetic field strength. As field strength increases, tissues take longer to relax.
- ✓ The TR controls T1 contrast. For good T contrast, the TR must be short.

MRI DESIGN: 9. T2 Decay

T2 decay is caused by the interaction between the magnetic fields of neighbouring spins. It is called **spin-spin**. It occurs as a result of the intrinsic magnetic fields of the nuclei interacting with each other. This produces a loss of phase coherence or dephasing, and results in decay of the NMV in the transverse plane. It is an exponential process and occurs at different rates in different tissues.

The T2 decay time of a particular tissue is an intrinsic contrast parameter and is inherent to the tissue being imaged. It is the time it takes for 63% of the transverse magnetization to be lost due to dephasing; that is, transverse magnetization is reduced by 63% of its original value (37% remains; Figure 9.1 and Table 9.1). The period of time during which this occurs is the time between the excitation pulse and the MR signal or the **TE**. The TE therefore determines how much T2 decay occurs in a particular tissue.

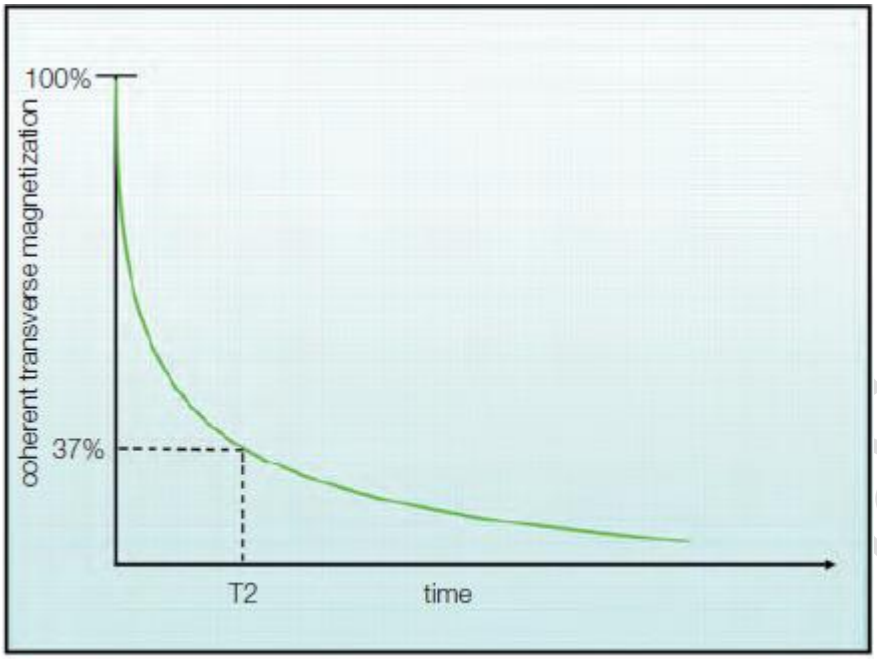


Figure 9.1 The T2 decay curve.

Table 9.1 T2 decay times of brain tissue at 1 T.

Equations		
$Mxy_t = Mxy\ e^{-t/T2}$	Mxy_t is the amount of transverse magnetization at time t (ms) after the removal of the excitation pulse. Mxy is full transverse magnetization. T2 is the T2 decay time (in ms) and is the time taken to reduce the transverse magnetization by a factor of e.	This equation plots the size of the decaying transverse magnetization as a function of time after the removal of the excitation pulse and the T2 decay time. When t=T2, 63% of the coherent transverse magnetization has decayed and 37% remains.

T2 decay in fat

T2 relaxation occurs as a result of the spins of adjacent nuclei interacting with each other. The efficiency of this process depends on how closely packed the molecules are to each other. In fat, the molecules are closely packed together so that spin-spin is efficient. *The T2 time of fat is therefore very short* (Figure 9.2).

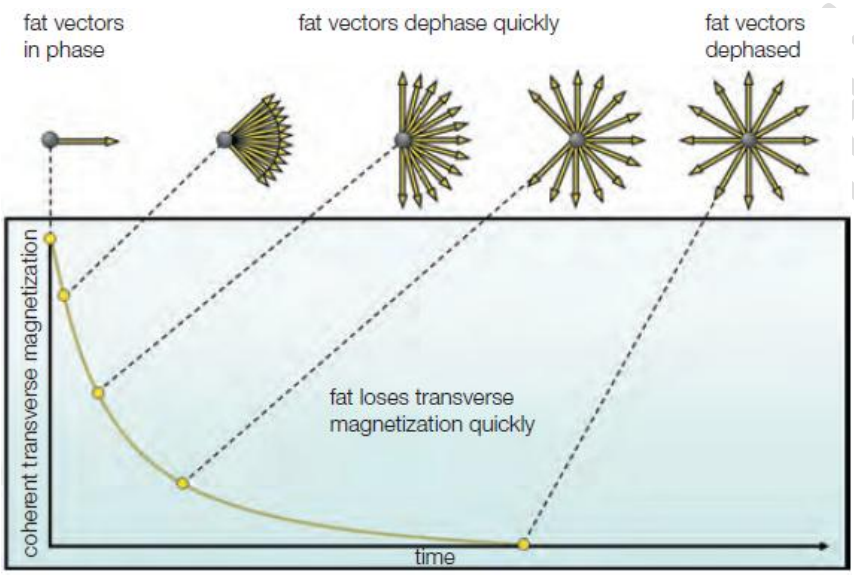


Figure 9.2 T2 decay in fat.

T2 decay in water

In water the molecules are spaced apart so that spin-spin is not efficient. *The T2 time of water is therefore very long* (Figure 9.3).

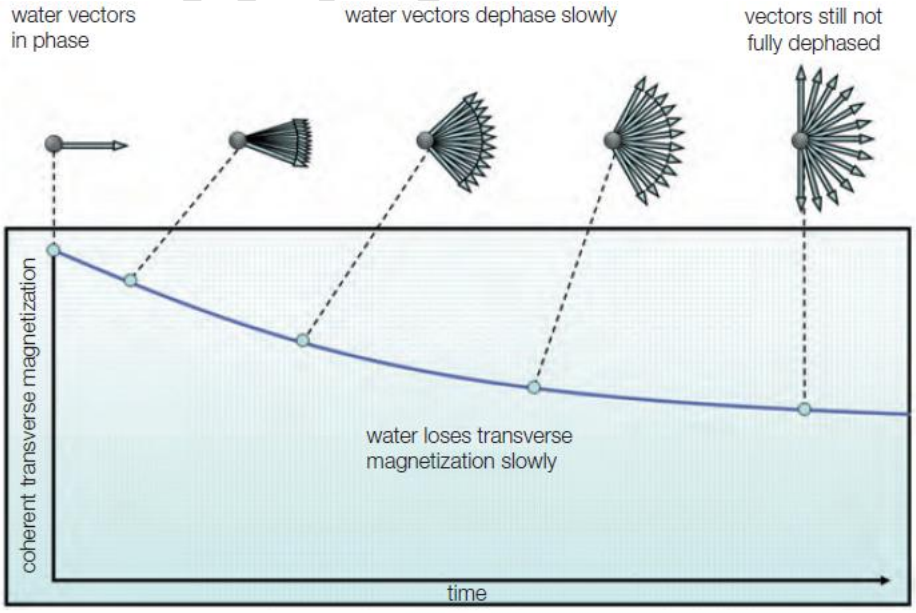


Figure 9.3 T2 decay in water.

Control of T2 decay

The **TE** controls how much transverse magnetization has been allowed to decay in fat and water when the signal is read.

A *short TE* does not permit full dephasing in either fat or water, so their coherent transverse components are similar. There is little contrast difference between fat and water due to differences in T2 decay times using a short TE.

A *long TE* allows dephasing of the transverse components in fat and water. There is a contrast difference between fat and water due to differences in T2 decay times when using a long TE (Figure 9.4).

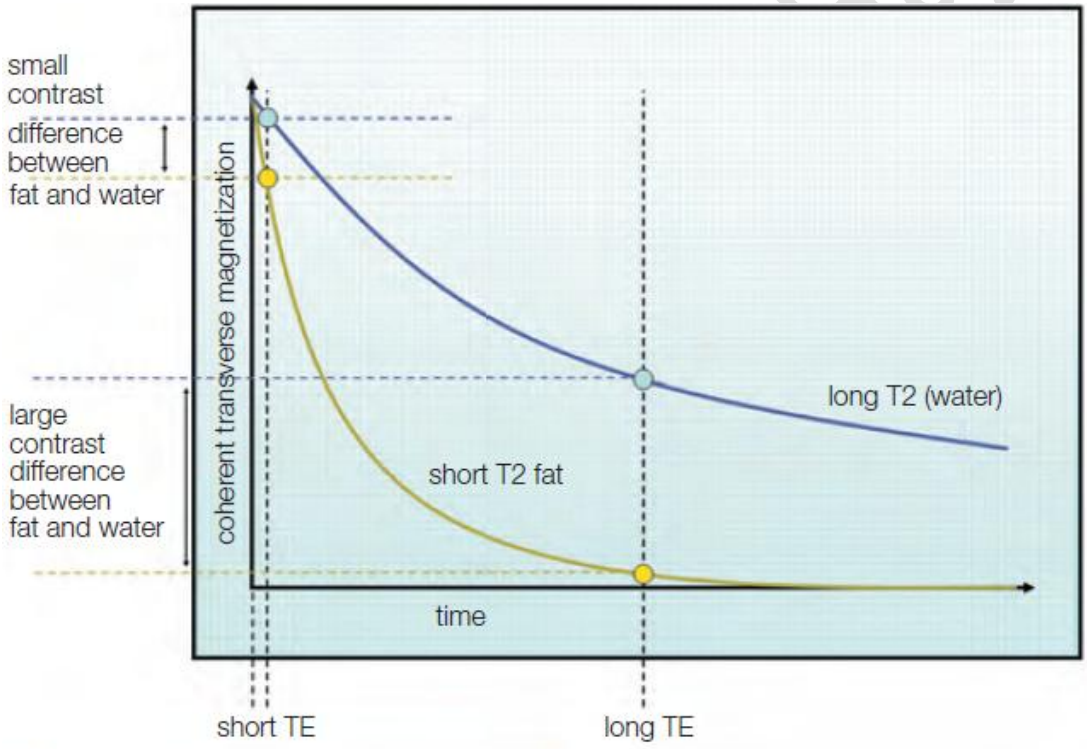


Figure 9.4 T2 decay curves in fat and water.

T2 decay is affected by the strength of the external magnetic field. Spin-spin processes are more efficient when molecular motion occurs at the Larmor frequency. The Larmor frequency is proportional to B_0 and therefore T2 decay takes longer as B_0 increases, because there are fewer molecules moving at relaxation-causing frequencies. It should be noted that fat and water represent the extremes in image contrast. Other tissues, such as muscle, grey matter and white matter, have contrast characteristics that fall between fat and water (Table 9.2).

Table 9.2 Equations of T2 decay.

Tissue	T2 decay time (ms)
Water	2500
Fat	100
CSF	300
White matter	100

Notes:

The **stationary frame** of reference refers to the observer (i.e. you) viewing something moving. You and the room you are situated in are stationary and what you are observing moves.

The **rotating frame of reference** refers to the observer viewing this from a different perspective. Imagine you are the thing that moves, what would the room look like? You would appear stationary and the room would appear to move.

A good example of this is to imagine the NMV relaxing back to B_0. If you were to observe this from the stationary frame of reference, as the NMV relaxes it also precesses around B_0. If you are looking at this from the rotating frame, however, *you* become the NMV as if you 'ride along' with it. From this perspective the room moves around you (the NMV) and you just smoothly relax back to B_0. In other words, from the rotating frame of reference it is the room that precesses relative to the NMV, rather than the NMV precessing relative to the room (as with the stationary frame of reference).

KEY POINTS

- ✓ Fat has a short T2 decay time.
- ✓ Water has a long T2 decay time.
- ✓ T2 decay is caused by spin-spin energy transfer. The efficiency of this process depends on how closely the molecules are packed together.
- ✓ T2 decay times are dependent on magnetic field strength. As field strength increases, tissues take longer to dephase.
- ✓ The TE controls T2 contrast. For good T2 contrast, the TE must be long.

Lecture Four

MRI DESIGN: T1 Weighting

All intrinsic contrast mechanisms affect image contrast, regardless of the pulse sequence used. For example, tissues with a low proton density, and air, are always dark on an MR image, and tissues in which spins move may be dark or bright depending on their velocity and the pulse sequence used (An **MRI pulse sequence** is a programmed set of changing magnetic gradients. Each sequence will have a number of parameters, and multiple sequences grouped together into an MRI protocol).

In order to produce images where the contrast is predictable, parameters are selected to weight the image towards one contrast mechanism and away from the others. This is achieved by understanding how extrinsic contrast parameters determine the degree to which intrinsic contrast parameters are allowed to affect image contrast. Extrinsic contrast parameters must be manipulated to accentuate one intrinsic contrast parameter and diminish the others. Proton density effects cannot be changed. T1 and T2 influences are manipulated by changing the TR and TE in the following way.

In a **T1 weighted image**, differences in the T1 relaxation times of tissues are accentuated and T2 effects are reduced. To achieve this, a TR is selected that is short enough to ensure that the NMV in neither fat nor water has had time to fully relax back to B_0 before the application of the next excitation pulse. The NMV in both fat and water is saturated. If the TR is long, the NMV in both fat and water recovers and the respective T1 relaxation times can no longer be distinguished.

A T1 weighted image is an image whose contrast is predominantly due to the differences in T1 recovery times of tissues. For T1 weighting, differences between the T1 times of tissues are exaggerated and to achieve this the *TR must be short*. At the same time, however, T2 effects must be minimized to avoid mixed weighting. To diminish T2 effects the *TE must also be short*.

In T1 weighted images, tissues containing a high proportion of fat, with short T1 relaxation times, are bright (high signal, hyper-intense), because they recover most of their longitudinal magnetization during the short TR period and Therefore, more magnetization is available to be flipped into the transverse plane by the next RF pulse and contribute to the signal (Table 10.1).

Tissues containing a high proportion of water, with long T1 relaxation times, are dark (low signal, hypo-intense), because they do not recover much of their longitudinal magnetization during the short TR period and therefore less magnetization is available to be flipped into the transverse plane by the next RF pulse and contribute to the signal (Table 10.1).

Table 10.1 Signal intensities seen in T1 weighted images.

High signal	fat haemangioma intraosseous lipoma radiation change degeneration fatty deposition methaemoglobin cysts with proteinaceous fluid paramagnetic contrast agents slow-flowing blood
Low signal	cortical bone avascular necrosis infarction infection tumours sclerosis cysts calcification
No signal	air fast-flowing blood tendons cortical bone scar tissue calcification

T1 weighted images best demonstrate anatomy, but also show pathology if used after contrast enhancement (Figures 10.1, 10.2 and 10.3).

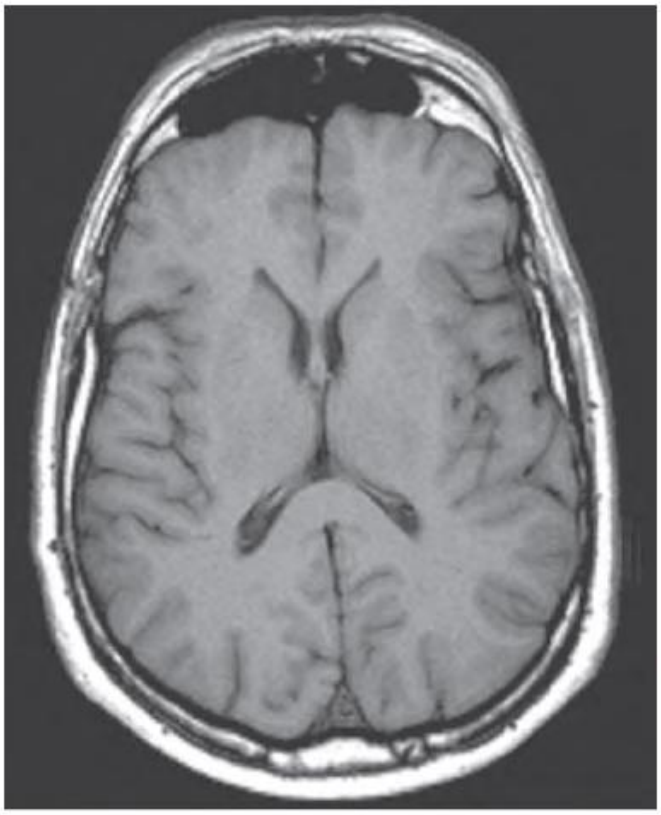


Figure 10.1 Axial T1 weighted image of the brain.

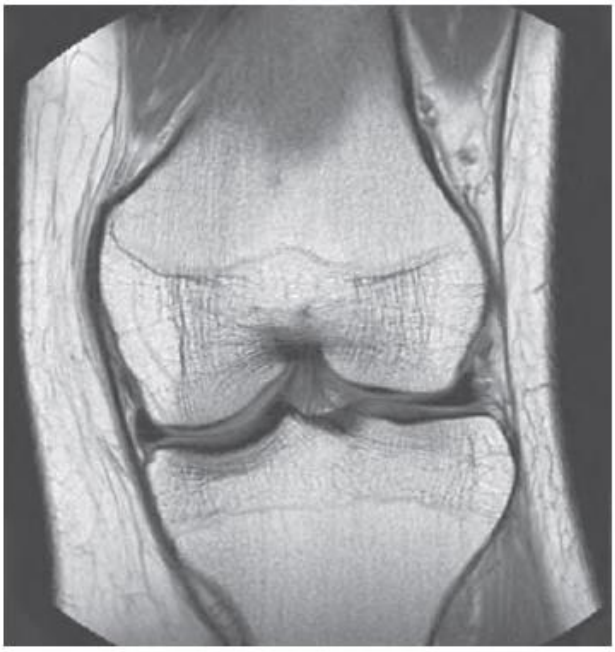


Figure 10.2 Coronal T1 weighted image of the knee.

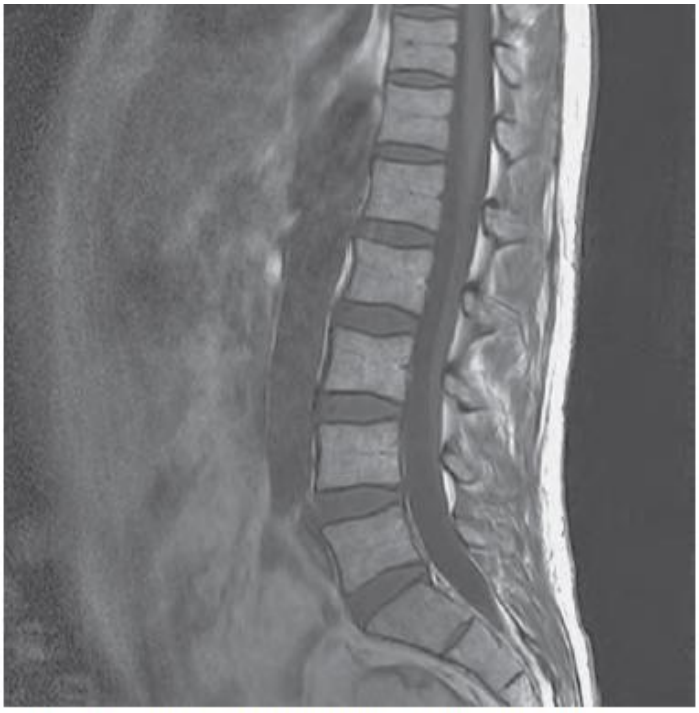


Figure 10.3 Sagittal T1 weighted image of the lumbar spine.

Typical values

• TR: 400–700 ms (shorter in gradient echo sequences).

• TE: 10–30 ms (shorter in gradient echo sequences).

The principal pulse sequences that are capable of producing T1 weighted images are:

• spin echo (produced by two successive RF pulses);

• turbo spin echo (produced by rapid acquisition with relaxation enhancement);

• inversion recovery (a conventional spin echo (SE) sequence preceded by a 180° inverting pulse);

• incoherent gradient echo (the utilization of gradient fields to generate transverse magnetization flip angles of less than 90°).

KEY POINTS

✓ All intrinsic contrast parameters contribute to image contrast.

✓ Extrinsic contrast parameters are used to control how much influence each intrinsic parameter has on image contrast.

✓ TR controls T1 contrast. TE controls T2 contrast.

✓ To produce a T1 weighted image it is necessary to create contrast in which the differences in the T1 recovery times of the tissues dominate image contrast.

✓ A short TR (e.g. 400ms) combined with a short TE (e.g. 10ms) maximizes T1 and minimizes T2 contrast respectively.

✓ T1 weighted images are used for anatomy and pathology post contrast enhancement.

MRI DESIGN: 11. T2 Weighting

All intrinsic contrast parameters affect image contrast, regardless of the pulse sequence, TR and TE used. For example, tissues with a low proton density, and air, are always dark on an MR image, and tissues in which nuclei move may be dark or bright depending on their velocity and the pulse sequence used.

Therefore, parameters are selected to weight the image towards one contrast mechanism and away from the others. This is achieved by understanding how extrinsic contrast parameters determine the degree to which intrinsic contrast parameters are allowed to affect image contrast. Extrinsic contrast parameters must be manipulated to accentuate one intrinsic contrast parameter and diminish the others. Proton density effects cannot be changed. T1 and T2 influences are manipulated by changing the TR and TE in the following way.

In a **T2 weighted image** the differences in the T2 relaxation times of tissues are accentuated and T1 effects are reduced. To achieve this, a long TE is selected to ensure that the NMV in both fat and water has had time to decay. If the TE is too short, the NMV in neither fat nor water has had time to decay and the respective T2 times cannot be distinguished. A T2 weighted image is an image whose contrast is predominantly due to the differences in the T2 decay times of tissues. For T2 weighting the differences between the T2 times of tissues are exaggerated, therefore the *TE must be long*. At the same time, however, T1 effects must be minimized to avoid mixed weighting. To diminish T1 effects *the TR must be long*.

Tissues containing a high proportion of fat, with a short T2 decay time, are dark (low signal, hypo-intense) because they lose most of their coherent transverse magnetization during the TE period (Table 11.1).

Table 11.1 Signal intensities seen in T2 weighted images.

High signal	water
	synovial fluid
	haemangioma
	infection
	inflammation
	oedema
	some tumours
	haemorrhage
	slow-flowing blood
	cysts
Low signal	cortical bone
	bone islands
	deoxyhaemoglobin
	haemosiderin
	calcification
	T2 paramagnetic agents
No signal	air
	fast-flowing blood
	tendons
	cortical bone
	scar tissue
	calcification

Tissues containing a high proportion of water, with a long T2 decay time, are bright (high signal, hyper-intense), because they retain most of their transverse coherence during the TE period (Table 11.1). T2 weighted images best demonstrate pathology, as most pathology has increased water content and is therefore bright on T2 weighted images (Figures 11.1, 11.2 and 11.3).

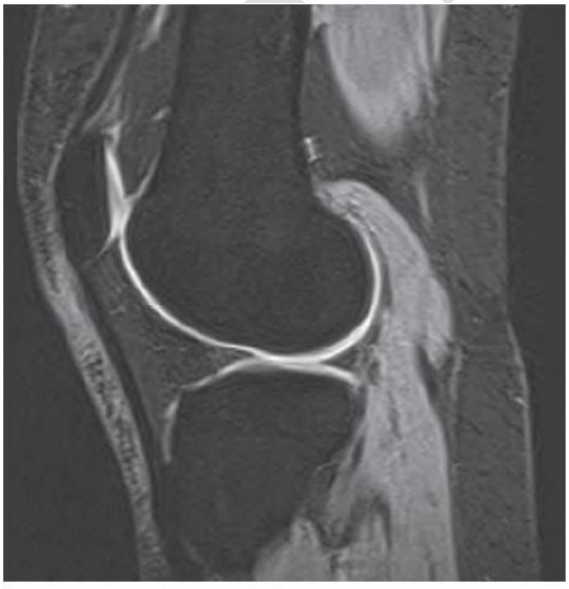


Figure 11.1 Axial T2 weighted image of the brain.

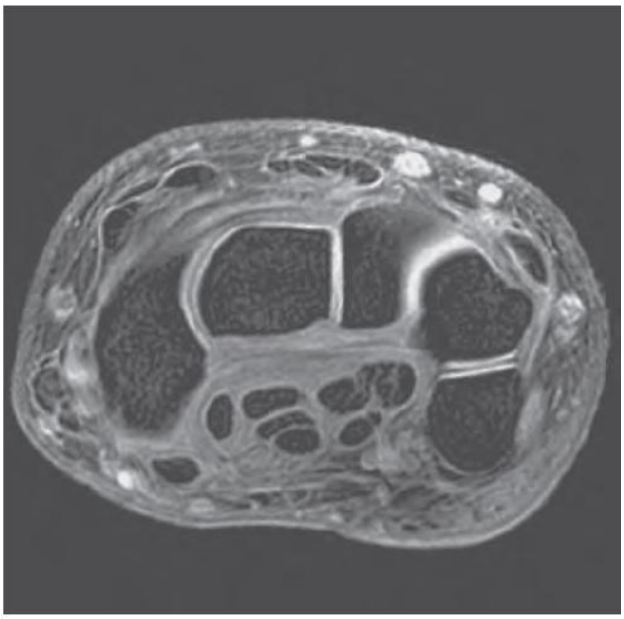


Figure 11.2 Axial T2 weighted image of the wrist.

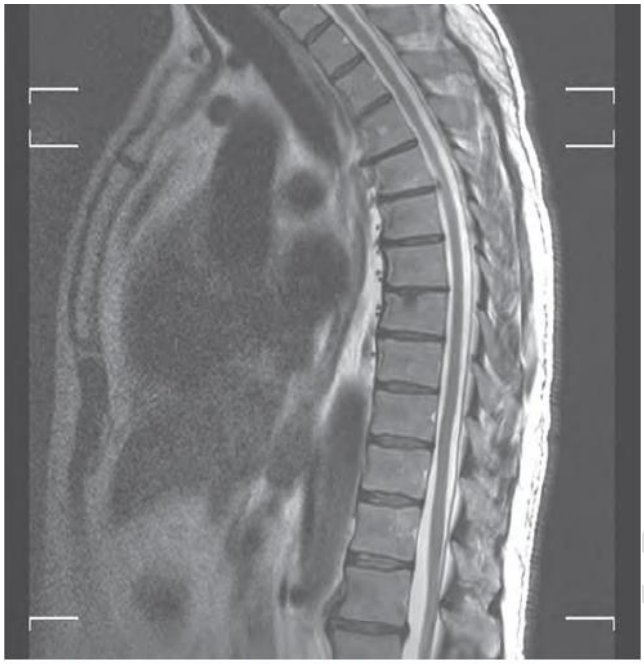


Figure 11.3 Sagittal T2 weighted image of the thoracic spine.

Typical values

• TR: 2000+ ms (much shorter in gradient echo sequences)

• TE: 70+ ms (shorter in gradient echo sequences)

The principal pulse sequences that are capable of producing T2 weighted images are:

• spin echo.

• turbo spin echo.

• STIR/FLAIR (STIR: stands for Short-T1 Inversion Recovery and is typically used to null the signal from fat. FLAIR: Fluid Attenuated Inversion Recovery).

The following pulse sequences produce T2* weighting that has similar characteristics in that water is bright. However, contrast in other tissues may be different.

• coherent gradient echo.

• balanced gradient echo.

KEY POINTS

- ✓ All intrinsic contrast parameters contribute to image contrast.
- ✓ Extrinsic contrast parameters are used to control how much influence each intrinsic parameter has on image contrast.
- ✓ TR controls T1 contrast. TE controls T2 contrast.
- ✓ To produce a T2 weighted image it is necessary to create contrast in which the differences in the T2 decay times of the tissues dominate image contrast.
- ✓ A long TR (e.g. 4000ms) combined with a long TE (e.g. 100ms) minimizes T1 and maximizes T2 contrast respectively.
- ✓ T2 weighted images are used for pathology.

MRI DESIGN: 12. PD Weighting

All intrinsic contrast parameters affect image contrast, regardless of the pulse sequence, TR and TE used. For example, tissues with a low proton density, and air, are always dark on an MR image, and tissues in which nuclei move may be dark or bright depending on their velocity and the pulse sequence used.

Therefore, parameters are selected to weight the image towards one contrast mechanism and away from the others. This is achieved by understanding how extrinsic contrast parameters determine the degree to which intrinsic contrast parameters are allowed to affect image contrast. Extrinsic contrast parameters must be manipulated to accentuate one intrinsic contrast parameter and diminish the others.

In a **proton density (PD) weighted image**, differences in the proton densities (number of hydrogen protons in the tissue) are demonstrated. To achieve this, both T1 and T2 effects are diminished. Selecting a long TR reduces T1 effects and T2 effects are diminished by selecting *a short TE*.

A proton density weighted image is an image whose contrast is predominantly due to differences in the proton density of the tissues.

Tissues with a low proton density are dark (low signal, hypointense) because the low number of protons results in a small component of transverse magnetization. Tissues with a high proton density are bright (high signal, hyper-intense) because the high number of protons results in a large component of transverse magnetization (Table 12.1).

Table 12.1 Signal intensities seen in PD weighted images.

High signal	CSF synovial fluid slow-flowing blood infection inflammation oedema cysts fat
Low or no signal	air fast-flowing blood tendons cortical bone scar tissue calcification

Cortical bone and air are always dark on MR images regardless of the weighting, as they have a low proton density and therefore return little signal. Proton density weighted images show anatomy and some pathology (Figures 12.1, 12.2 and 12.3).

Typical values

- TR: 2000ms+
- TE: 10–30ms

The main pulse sequences that are used to obtain PD weighting are:

- spin echo;
- turbo spin echo.

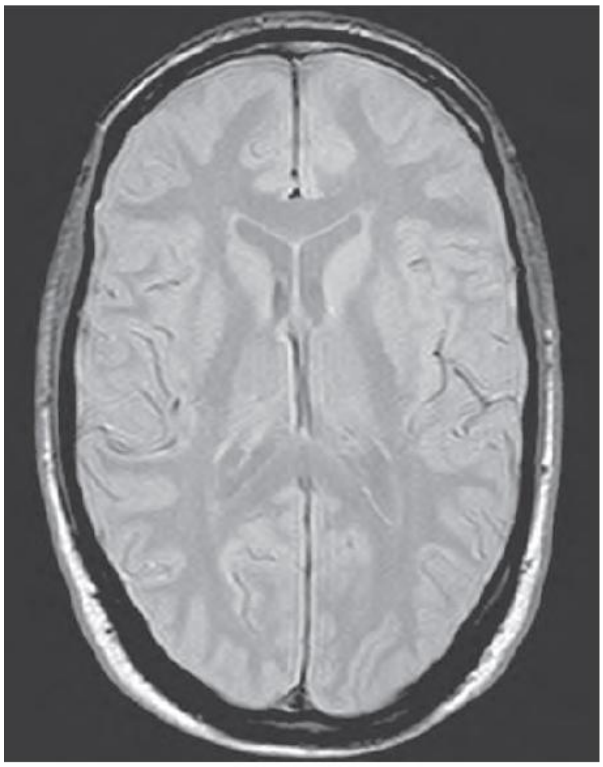


Figure 12.1 Axial proton density weighted image of the brain.

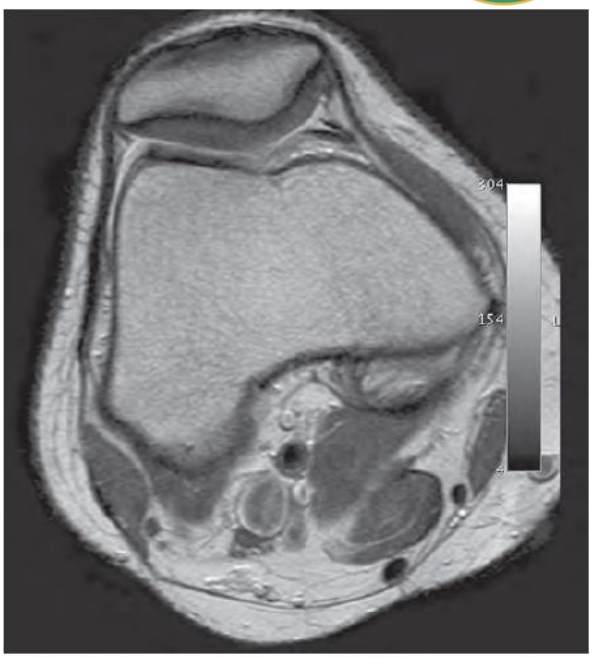


Figure 12.2 Axial proton density weighted image of the knee.

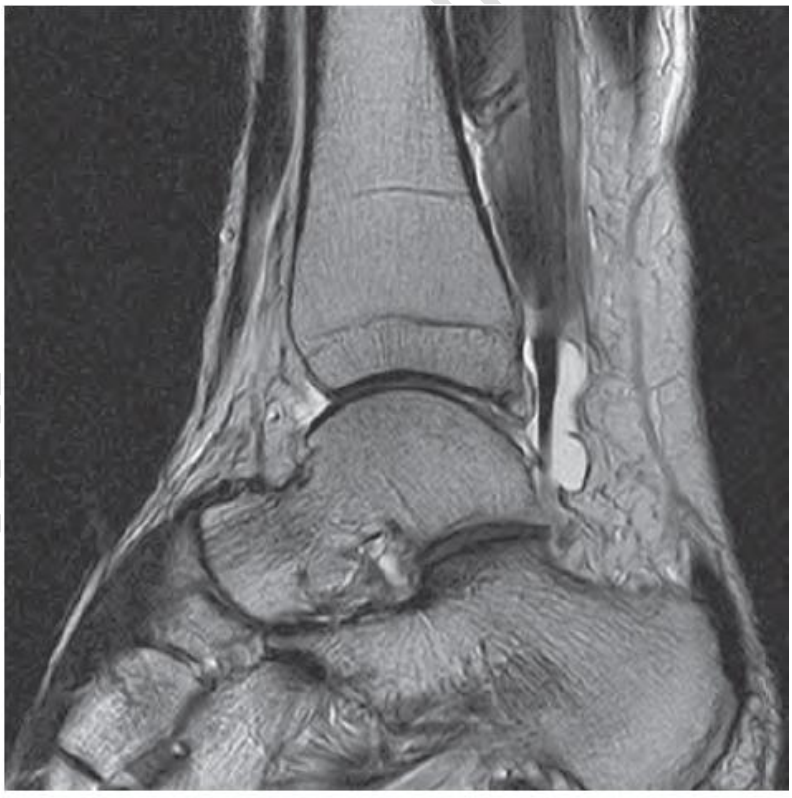


Figure 12.3 Sagittal proton density weighted image of the ankle.

Other types of weighting

Flow and the ADC of a tissue also affect weighting, as they are intrinsic contrast mechanisms. Flow mechanisms can be used to weight the image specifically to flowing spins. Flow-related weighting is achieved in MR angiography techniques. ADC-related weighting is achieved in diffusion weighting.

KEY POINTS

- ✓ All intrinsic contrast parameters contribute to image contrast. Extrinsic contrast parameters are used to control how much influence each intrinsic parameter has on image contrast.
- ✓ TR controls T1 contrast. TE controls T2 contrast.
- ✓ To produce a PD weighted image it is necessary to create contrast in which the differences in the proton densities of the tissues dominate image contrast.
- ✓ A long TR (e.g. 4000 ms) combined with a short TE (e.g. 20ms) minimizes T1 and T2 contrast respectively so that PD can dominate.
- ✓ PD weighted images are used for anatomy and pathology.

MRI DESIGN: 13. Conventional Spin Echo

Pulse sequences are defined as a series of RF pulses, gradient applications and intervening time intervals. All pulse sequences contain these elements. They differ only in the way they are coordinated and timed.

Conventional spin echo (SE or CSE) pulse sequences are used to produce T1, T2 or proton density weighted images and are one of the most basic pulse sequences used in MRI. In a spin echo pulse sequence, there is a 90° excitation pulse followed by a 180° rephasing pulse followed by an **echo**.

Mechanisms of CSE

After the application of the 90° RF pulse, the magnetic moments of the spins lose precessional coherence because of an increase or decrease in their precessional frequency caused by the magnetic field inhomogeneities. This results in a decay of coherent magnetization in the transverse plane and the ability to generate a signal is lost.

Magnetic moments that experience an increase in precessional frequency gain phase relative to those that experience a decrease in precessional frequency, which lag behind. Dephasing can be imagined as a 'fan' where magnetic moments that lag behind precess more slowly, and those that gain phase precess more quickly.

A 180° RF pulse flips magnetic moments of the dephased spins through 180°. The fast edge of the fan is now positioned behind the slow edge. The fast edge eventually catches up with the slow edge, therefore **rephasing** the spins (Figure 13.1).

The coherent signal in the receiver coil is regenerated and can be measured. This regenerated signal is called an **echo** and, because an RF pulse has been used to generate it, it is specifically called a **spin echo**.

Rephasing the spins eliminates the effect of the magnetic field inhomogeneities. Whenever a 180° RF rephasing pulse is applied, a spin echo results. Rephasing pulses may be applied either once or several times to produce either one or several spin echoes.

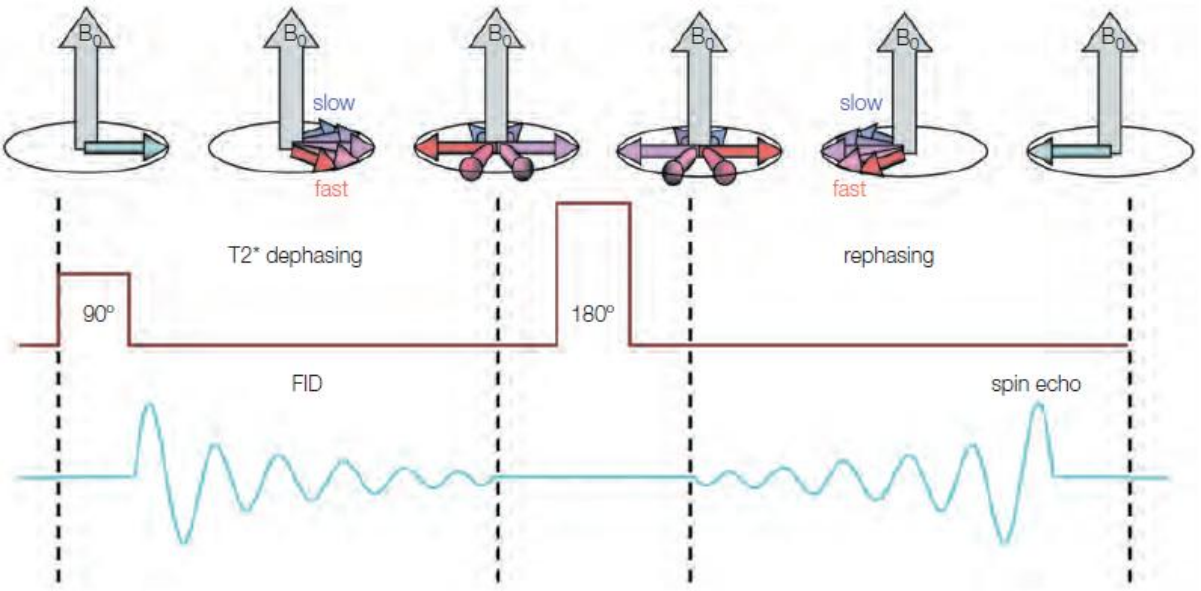


Figure 13.1 180° RF rephasing.

Contrast

CSE is usually used in one of two ways:

• A **single spin echo** pulse consists of a single 180° RF pulse applied after the excitation pulse to produce a single spin echo (Figure 13.2). This a typical sequence used to produce a T1 weighted set of images.

The **TR** is the length of time from one 90° RF pulse to the next 90° RF pulse in a particular slice. For T1 weighted imaging a short TR is used.

The **TE** is the length of time from the 90° RF pulse to the midpoint or peak of the signal generated after the 180° RF pulse; that is, the spin echo. For T1 weighted imaging a short TE is used.

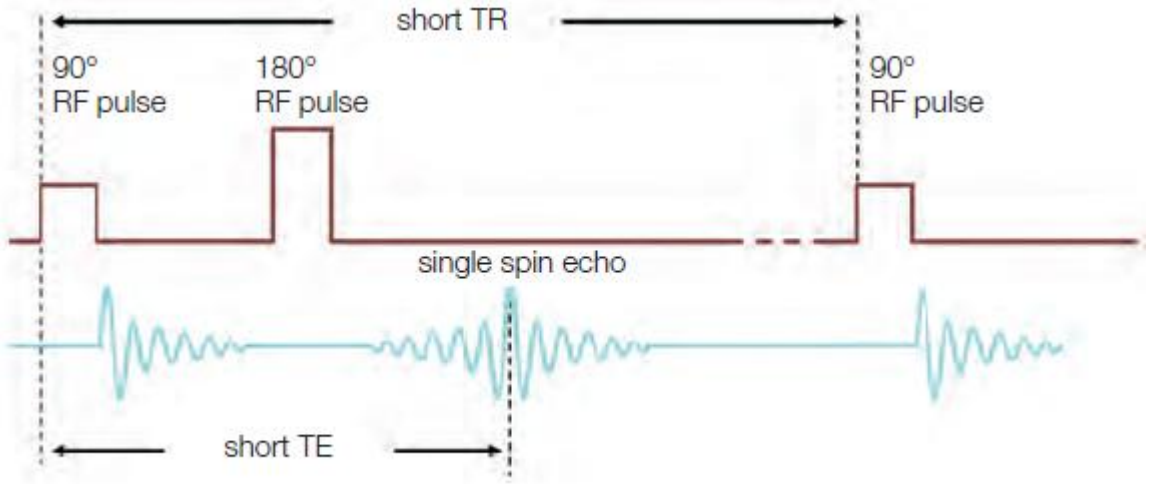


Figure 13.2 Single-echo spin echo sequence.

• A **dual echo sequence** consists of two 180° pulses applied to produce two spin echoes. This is a sequence that provides two images per slice location: one that is proton density weighted and one that is T2 weighted (Figure 13.3). The first echo has a short TE and a long TR and results in a set of proton density weighted images. The second echo has a long TE and a long TR and results in a T2 weighted set of images. This echo has less amplitude than the first echo because more T2 decay has occurred by this point.

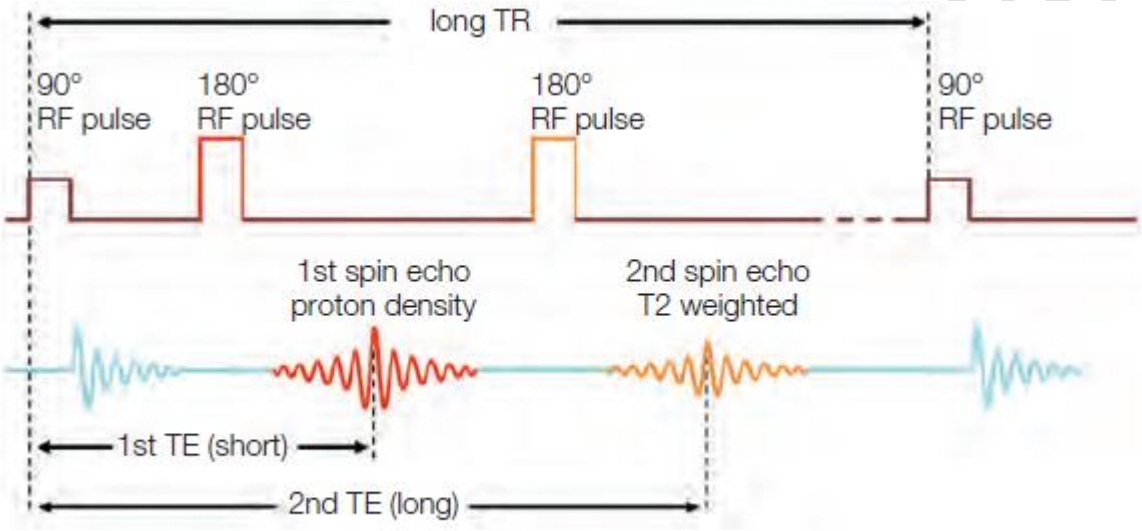


Figure 13.3 Dual-echo spin echo sequence.

Typical values

Single echo (for T1 weighting)

• TR: 400–700ms

• TE: 10–30ms

Dual echo (for PD/T2 weighting)

• TR: 2000+ms

• TE1: 20ms

• TE2: 80ms

Uses

Spin echo sequences are still considered the 'gold standard' (Table 13.1) in that the contrast they produce is understood and is predictable. They produce T1, T2 and PD weighted images of good quality and may be used in any part of the body, for any indication (Figures 13.4 and 13.5). However, due to relatively long scan times, PD and T2 weighted images are now usually acquired using fast or turbo spin echo.

Table 13.1 Advantages and disadvantage of conventional spin echo.

Advantages	Disadvantage
Good image quality Very versatile True T2 weighting Available on all systems Gold standard for image contrast and weighting	Long scan times

KEY POINTS

- ✓ Spin echo sequences are characterized by 180°RF rephasing pulses that refocus the magnetic moments of spins to produce an echo.
- ✓ T1, T2 and PD weighting are all achievable using conventional spin echo.
- ✓ Conventional spin echo is traditionally used to acquire one or two echoes to achieve T1, T2 or PD weighting.
- ✓ Although quite old sequences, they are still considered the gold standard and can be used to image anatomy and pathology in all body areas.

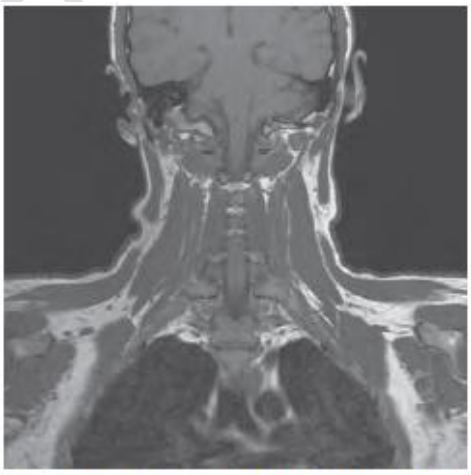


Figure 13.4 Coronal T1 weighted SE image of the brachial plexus.

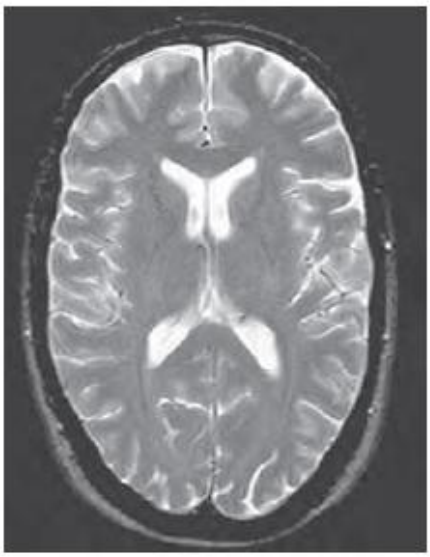


Figure 13.5 Axial T2 weighted SE image of the brain.

Lecture Five

MRI DESIGN: 27. Gradient Functions

Gradients are coils of wire that, when a current is passed through them, alter the magnetic field strength of the magnet in a controlled and predictable way. They add to or linearly subtract from the existing field so that the magnetic field strength at any point along the gradient is known (Figure 27.1).

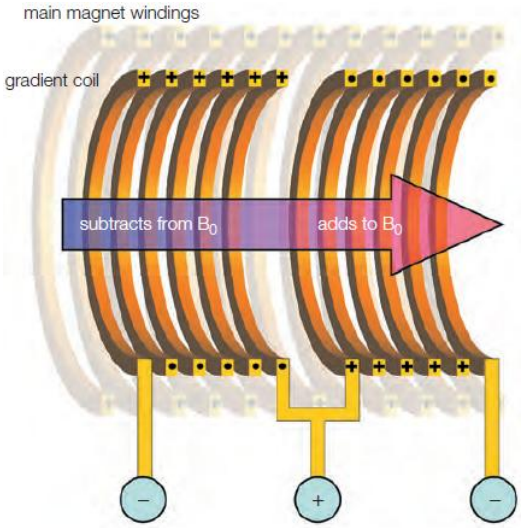


Figure 27.1 A gradient coil.

When a gradient is applied the following occur. At **magnetic isocentre** (the centre of all three gradients), the field strength remains unchanged even when the gradient is switched on. At a certain distance away from isocentre, the field strength either increases or decreases. The magnitude of the change depends on the distance from isocentre and the strength of the gradient (Figure 27.2).

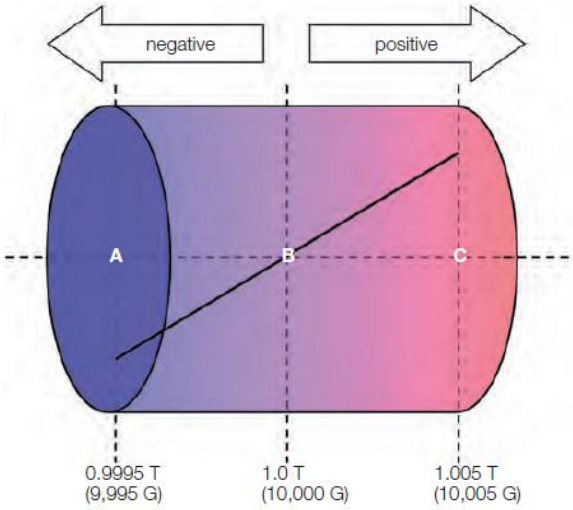


Figure 27.2 Gradients and changing field strength.

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The slope of the gradient signifies the rate of change of the magnetic field strength along its length. The strength or *amplitude* of the gradient is determined by *how much current* is applied to the gradient coil. Larger currents create steeper gradients, so that the change in field strength over distance is greater. The reverse is true of smaller currents.

The polarity of the gradient determines which end of the gradient produces a higher field strength than isocentre (positive) and which a lower field strength than isocentre (negative). The *polarity* of the gradient is determined by the *direction of the current* flowing through the coil. As coils are circular, current either flows clockwise or anticlockwise.

The *maximum amplitude* of the gradient determines the maximum achievable *resolution*. Therefore, if at least one (and sometimes all three) gradients are steep, small voxels are achieved.

The *maximum speeds* at which gradients can be switched on and off are called the **rise time** and **slew rate**. Both of these factors determine the maximum scan speeds of a system. Therefore, in fast sequencing the gradients have high slew rates.

How gradients work

The precessional frequency of the magnetic moments of nuclei is proportional to the magnetic field strength experienced by them (as stated by the Larmor equation). The frequency of signal received from the patient can be changed according to its position along the gradient. The precessional phase is also affected, as faster magnetic moments gain phase compared with their slower neighbours. Imposing a gradient magnetic field therefore:

• changes the field strength in a linear fashion across a distance in the patient;

• changes the precessional **frequency** of magnetic moments of nuclei in a linear fashion across a distance in the patient (Table 27.1);

Table 27.1 Frequency changes along a linear gradient.

Position along gradient	Field strength (gauss)	Larmor frequency (MHz)
isocentre	10000	42.5700
1 cm negative from isocentre	9999	42.5657
2 cm negative from isocentre	9998	42.5614
1 cm positive from isocentre	10001	42.5742
2 cm positive from isocentre	10002	42.5785
10 cm negative from isocentre	9990	42.5274

• changes the precessional **phase** of magnetic moments of nuclei in a linear fashion across a distance in the patient (Figure 27.3). These characteristics can be used to **encode** the MR signal in three dimensions. In order to do this, there must be three orthogonal sets of gradients situated within the bore of the magnet. They are named according to the axis along which they work:

• The *Z gradient* alters the magnetic field strength along the *Z axis.*

• The *Y gradient* alters the magnetic field strength along the *Y axis.*

• The *X gradient* alters the magnetic field strength along the *X axis.*

• The **magnetic isocentre** is the centre of all three gradients.

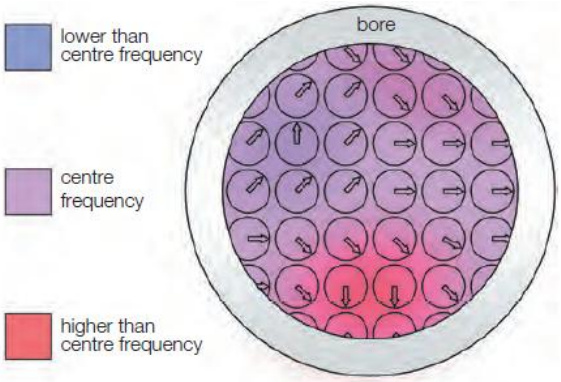


Figure 27.3 How gradients change frequency and phase.

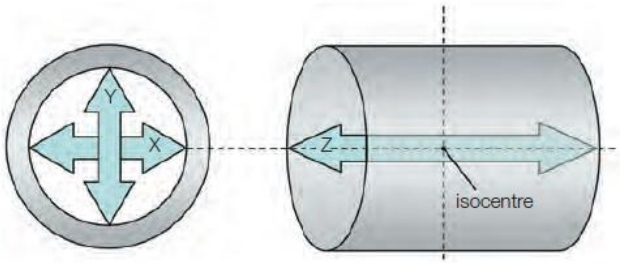


Figure 27.4 Gradient axes.

The field strength here does not change even when a gradient is applied (Figure 27.4). There are only three gradients, but they are used to perform many important functions during a pulse sequence. For example, in gradient echo sequences, a gradient is used to refocus spins and produce a gradient echo. One of these functions is **spatial encoding**; that is, spatially locating a signal in three dimensions. In order to do this, three separate functions are necessary. Usually each gradient performs one of the following tasks. The gradient used for each task depends on the plane of the scan and on which gradient the operator selects to perform frequency or phase encoding.

• **Slice selection** − locating a slice in the scan plane selected.

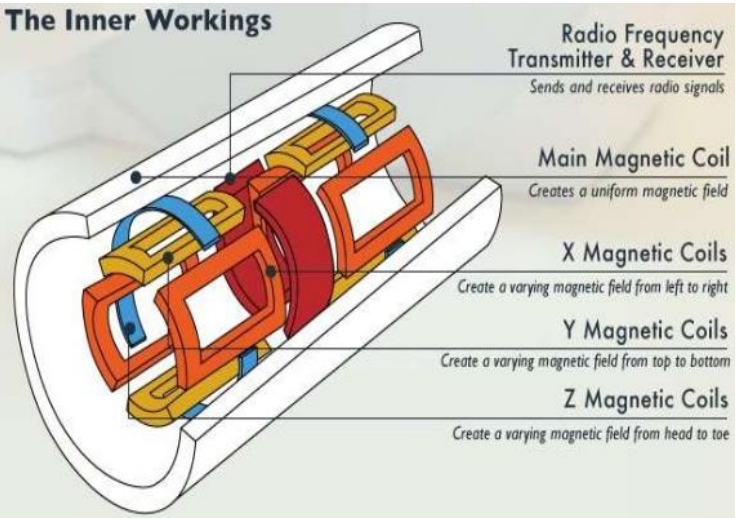
• Spatially locating signal along the short axis of the image. This is called **phase encoding**.

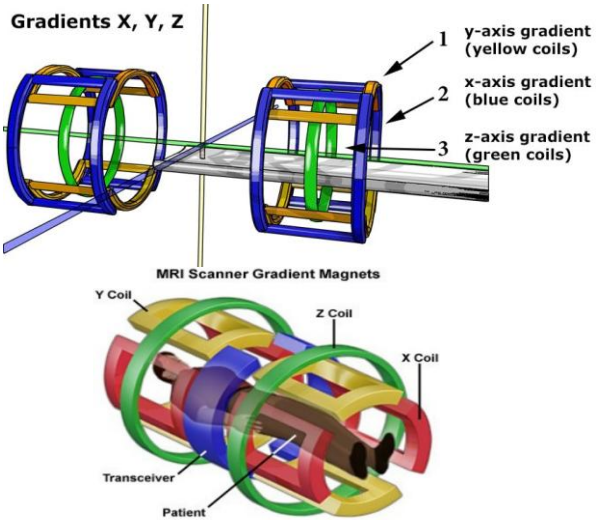
• Spatially locating signal along the long axis of the image. This is called **frequency encoding** (Table 27.2).

Table 27.2 Gradient axes in orthogonal imaging.

	Slice selection	Phase encoding	Frequency encoding
Sagittal	X	Y	Z
Axial (body)	Z	Y	X
Axial (head)	Z	X	Y
Coronal	Y	X	Z

(Where X is across the bore of the magnet from right to left)





KEY POINTS

✓ When a moving current is passed through a conductor, a magnetic field is induced around it.

✓ Gradient coils are conductors that cause a linear change in magnetic field strength along their axes when a current is passed through them.

✓ The amount of current passing through the coil determines the amplitude, strength or slope of the gradient.

✓ The direction of the current passing through the coil determines its polarity.

✓ When a gradient is switched on, it causes a linear change in magnetic field strength and, therefore, precessional frequency and phase of the magnetic moments of spins that lie along it.

MRI DESIGN: 28. Slice Selection

Mechanism

As a gradient alters the magnetic field strength of the magnet linearly, the magnetic moments of spins within a specific slice location along the gradient have a unique precessional frequency when the gradient is on. Transmitting RF at that unique precessional frequency, therefore, selectively excites a slice.

Example: a 1T field strength magnet with a gradient imposed that has changed the field strength between slices A and B, causing a change in precessional frequency between slices A and B of 2.6 MHz (Figure 28.1).

• The precessional frequency of magnetic moments between slices A and B has changed by 2.6 MHz.

• To excite nuclei in slice A, an RF pulse of 41.20 MHz must be applied.

• Slice B and all other slices are not excited because their precessional frequencies are different due to the influence of the gradient.

• To excite slice B, another RF pulse with a frequency of 43.80 MHz must be applied. Nuclei in slice A do not resonate after the application of this pulse because they are spinning at a different frequency.

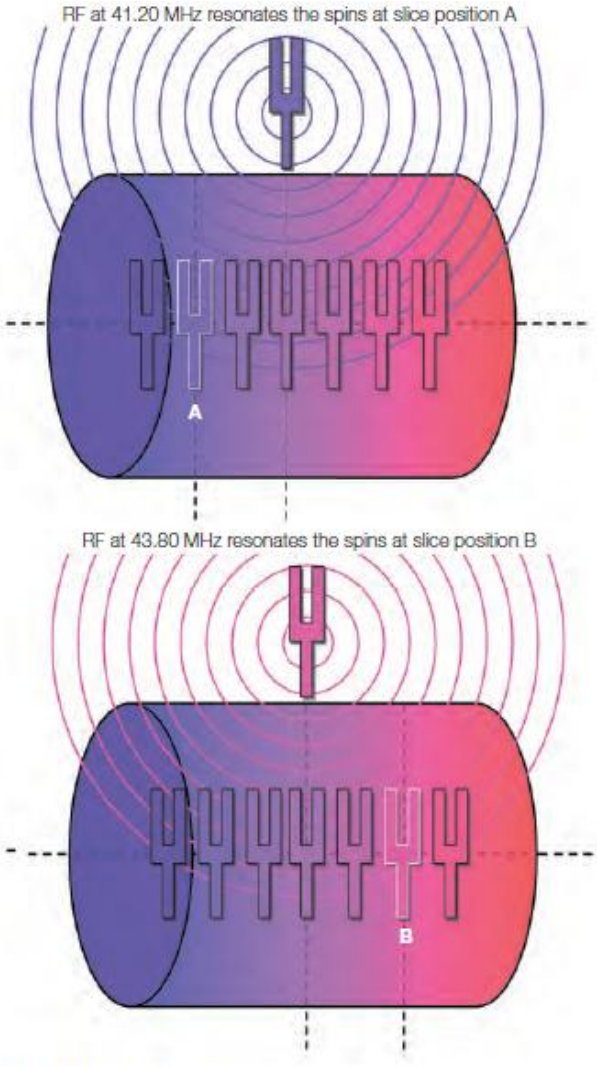


Figure 28.1 Slice selection.

The scan plane selected determines which gradient performs slice selection. In a superconducting system the following usually apply (in an open magnet system, the Z and Y axes are transposed and some manufacturers transpose X and Y):

• The Z gradient selects axial slices, so that nuclei in the patient's head spin at a different frequency to those in the feet.

• The Y gradient selects coronal slices, so that nuclei at the back of the patient spin at a different frequency to those at the front.

• The X gradient selects sagittal slices, so that nuclei on the righthand side of the patient spin at a different frequency to those on the left (Figure 28.2).

• A combination of any two gradients selects oblique slices.

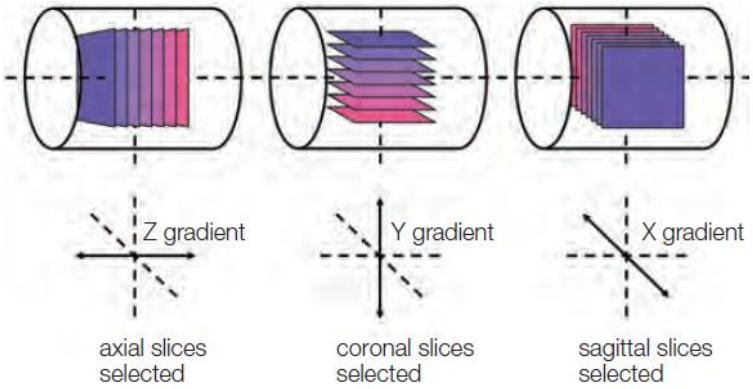


Figure 28.2 Using X, Y and Z gradients to select slices.

Slice thickness

In order to attain slice thickness, a range of frequencies must be transmitted to produce resonance across the whole slice (and therefore to excite the whole slice). This range of frequencies is called a bandwidth and because RF is being transmitted at this instant, it is specifically called the **transmit bandwidth**.

The slice thickness is determined by the slope of the slice select gradient and the transmit bandwidth. It affects inplane spatial resolution and SNR.

- Thin slices require a steep slope or a narrow transmit bandwidth, and improve spatial resolution.
- Thick slices require a shallow slope or a broad transmit bandwidth, and decrease spatial resolution (Figure 28.3).

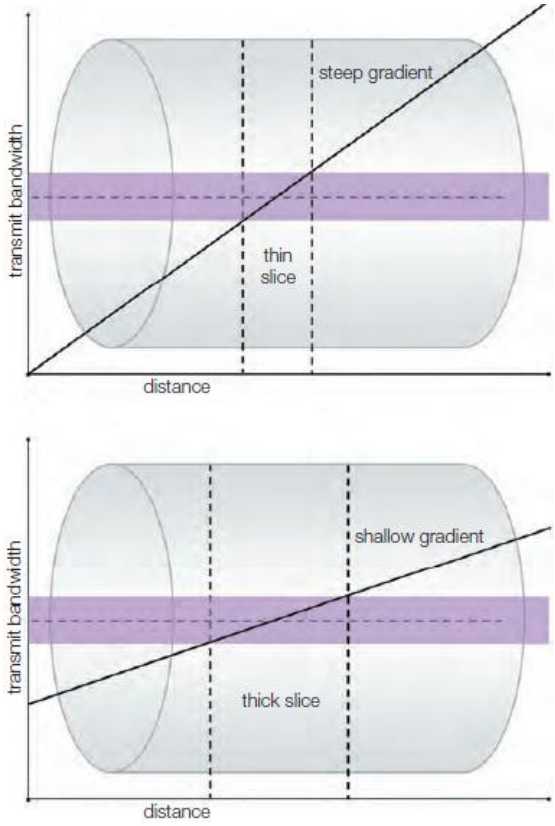
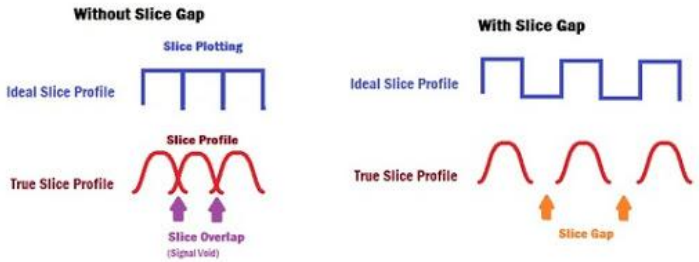


Figure 28.3 Transmit bandwidth, gradient slope and slice thickness

A slice is therefore excited by transmitting RF with a centre frequency corresponding to the middle of the slice, and a bandwidth and gradient slope according to the thickness of the slice required. The slice gap or skip is the space between slices. Too small a gap in relation to the slice thickness can lead to an artefact called **cross-talk**. This is caused because RF excitation pulses are Gaussian in shape (not exactly square). They have small 'tails' that overlap when RF pulses are too close together. This causes part of a slice to receive too much RF, resulting in cross-talk artefact.



The slice select gradient is always switched on during the delivery of the RF excitation pulse in the pulse sequence. It is switched on in the positive direction. The slice select gradient is also applied during the 180° pulse in spin echo sequences so that the RF rephasing pulse can be delivered specifically to the selected slice (Figure 28.4). Although not always shown, in all pulse sequences compensatory gradients are applied around each application of the slice select gradient. This is to compensate for the change of phase that the gradient imposes. This change of phase is not wanted in the slice selection process and is eliminated by these compensatory gradients.

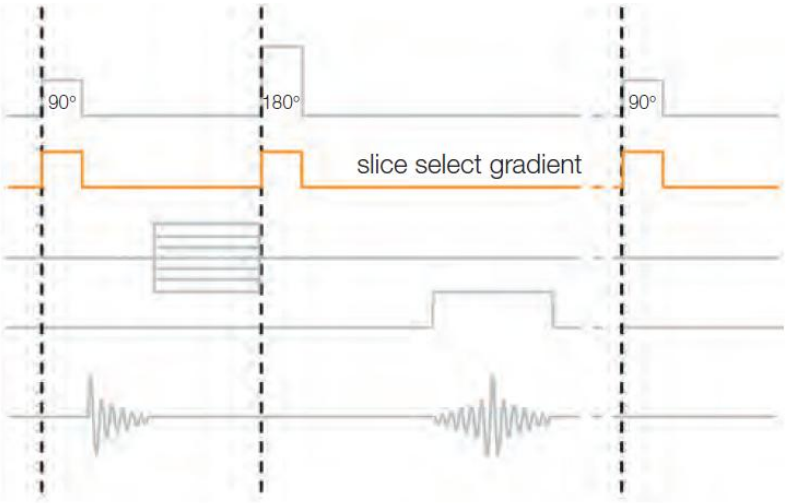


Figure 28.4 Timing of slice selection in a spin-echo pulse sequence.

KEY POINTS

✓ Slices are selected by applying a gradient at the same time as the RF excitation and rephasing pulse.

✓ The slice select gradient changes the magnetic field strength and therefore the precessional frequency of the magnetic moments of spins that lie along it.

✓ An RF pulse at the specific frequency of magnetic moments of spins in a particular slice on the gradient causes resonance of the slice.

✓ RF is transmitted width a bandwidth or range of frequencies on either side of the centre frequency of the slice.

✓ Slice thickness is altered by changing either the slope of the slice select gradient or the transmit bandwidth.

✓ Thin slices require either a steep slice select gradient slope or a narrow transmit bandwidth.

✓ Thick slices require either a shallow slice select gradient slope or a broad transmit bandwidth.

MRI DESIGN: 50. Contrast Agents

In order to increase contrast between pathology and normal tissue, enhancement agents may be introduced that selectively affect the T1 and T2 relaxation times in tissues. Both T1 recovery and T2 decay are influenced by the magnetic field experienced locally within the nucleus. The local magnetic field responsible for these processes is caused by:

• the main magnetic field;

• fluctuations as a result of the magnetic moments of nuclear spins in neighbouring molecules.

These molecules rotate or tumble, and the rate of rotation of the molecules is a characteristic property of the solution. It is dependent on:

• magnetic field strength;

• viscosity of the solution;

• temperature of the solution.

Molecules that tumble with a frequency at or near the Larmor frequency have more efficient T1 recovery times than other molecules (Figure 50.1).

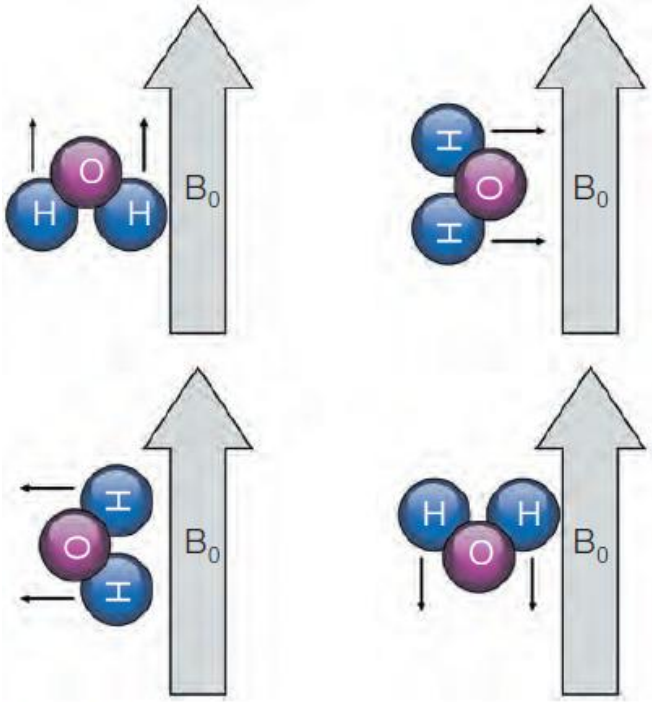


Figure 50.1 Tumbling of water molecules.

The phenomenon by which excited protons are affected by nearby excited protons and electrons is called dipole-dipole interaction. If a tumbling molecule with a large magnetic moment such as gadolinium is placed in the presence of water protons, local magnetic field fluctuations occur near the Larmor frequency. T1 relaxation times of nearby protons are therefore reduced and so they appear bright on a T1 weighted image. This effect on a substance whereby relaxation rates are altered is known as relaxivity.

Gadolinium

Gadolinium (Gd) is a paramagnetic agent. It is a trivalent lanthanide element that has seven unpaired electrons and an ability to allow rapid exchange of bulk water to minimize the space between itself and water within the body. It has a large magnetic

moment and, when it is placed in the presence of tumbling water protons, fluctuations in the local magnetic field are created near the Larmor frequency. The T1 relaxation times of nearby water protons are therefore reduced, resulting in an increased signal intensity on T1 weighted images. For this reason, gadolinium is known as a T1 enhancement agent.

Chelation

Gadolinium is a rare earth metal that cannot be excreted by the body and would cause long-term side effects as it binds to membranes. By binding the gadolinium ion to a chelate such as diethylene triaminepentaacetic acid (DTPA, a ligand), the chelate compound Gd-DTPA is formed, which can be safely excreted.

Administration

The effective dosage of Gd-DTPA is 0.1 millimoles per kilogram of body weight (mmol/kg) − approximately 0.2 ml/kg or 0.1 ml/lb − with a maximum dose of 20 ml.

Clinical applications

Gadolinium has proven invaluable in imaging the central nervous system because of its ability to pass through breakdowns in the blood–brain barrier (BBB). Clinical indications for gadolinium include:

- tumours (Figure 50.3);
- infection;
- arthrography (Figure 50.2);
- post-operation lumbar disc;
- breast disease;
- vessel patency and morphology.

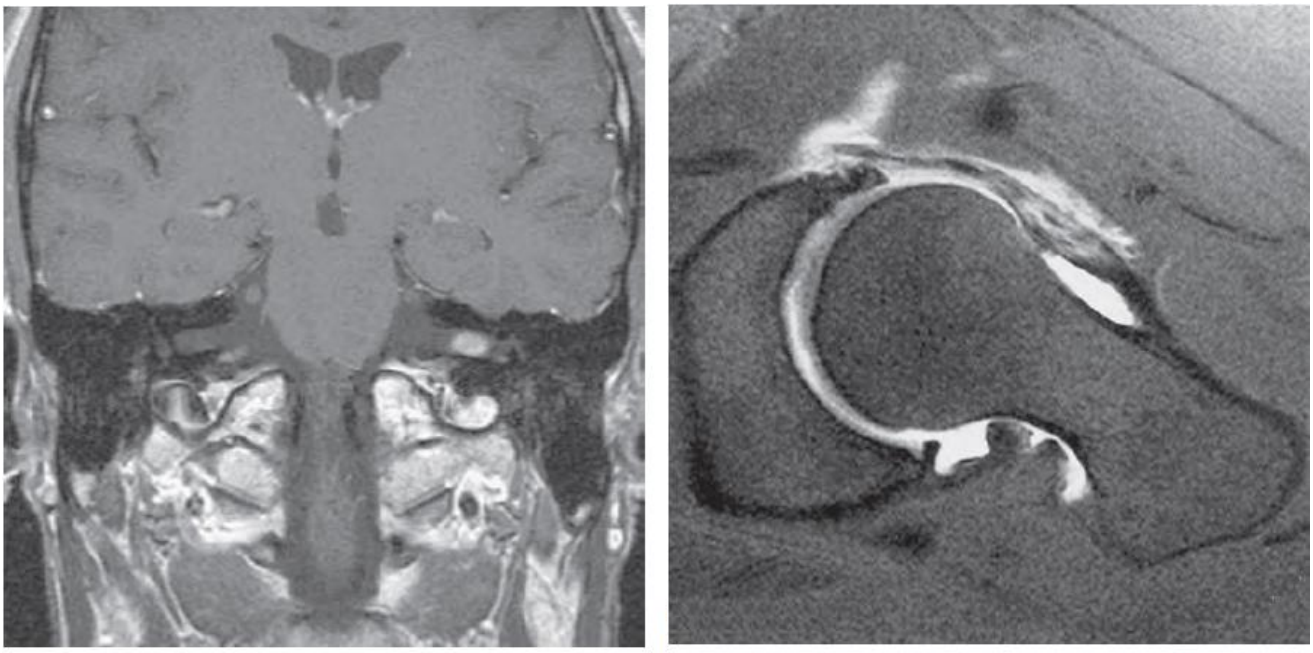


Figure 50.2 Axial arthrogram of the hip using gadolinium.

Iron oxide

Iron oxides shorten relaxation times of nearby hydrogen atoms and therefore reduce the signal intensity in normal tissues. This results in a signal loss on proton density weighted or heavily T2 weighted images. Super-paramagnetic iron oxides are known as T2 enhancement agents. Iron oxide is taken up by the reticuloendothelial system and excreted by the liver, so that normal liver is dark and liver lesions are bright on T2 weighted images.

Administration

The recommended dose of iron oxide is 0.56 mg of iron per kg of body weight. This should be diluted in 100 ml of 50% dextrose and given intravenously over 30 mins. The diluted drug is administered through a 5-micron filter at a rate of 2−4 mmol/min. This agent should be used within 8 hours following dilution.

Clinical applications

Iron oxide is mainly used in liver and biliary imaging (Figure 50.4).

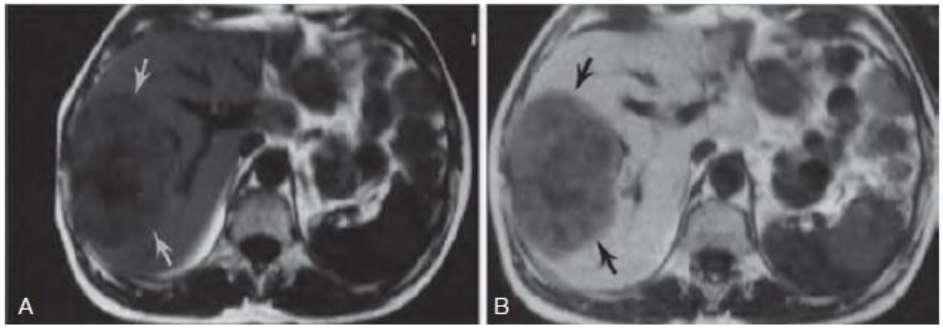


Figure 50.4 Axial T1 weighted image of the liver without (left) and with (right) manganese contrast.The enhanced image shows enhancement of normal liver so that the liver pathology is darker.

Other contrast agents

Gastrointestinal contrast agents are sometimes used for bowel enhancement. These include barium, ferromagnetic agents and fatty substances. However, due to constant peristalsis, these agents enhance bowel motion artefacts more often than enhancing pathological lesions. The use of anti-spasmodic agents helps to retard peristalsis to decrease these artefacts. Other agents include helium, which is inhaled and assists in the evaluation of lung perfusion.

Key points.

- The purpose of contrast agents is to ensure that pathology has a different contrast to surrounding normal anatomy.
- Contrast agents are either T1 or T2 enhancement agents.
    - The effect of altering the relaxivity rates of tissues by administering a contrast agent is called relaxivity.

Lecture Six

MRI DESIGN: 14. Fast or turbo spin echo

Fast or turbo spin echo (FSE or TSE) is a much faster version of conventional spin echo. It is sometimes called a rapid acquisition with relaxation enhancement (RARE) sequence. In spin echo sequences, only one phase of encoding is performed during each TR. The scan time is a function of TR, NSA (***number of excitations (NEX)***, also known on some systems as the ***number of signals averaged (NSA)***, is an important determinant of SNR) and phase matrix. One of the ways of speeding up a conventional sequence is to reduce the number of phase-encoding steps. However, this normally results in a loss of resolution. TSE overcomes this by still performing the same number of phase encodings, thereby maintaining the phase matrix, but more than one phase encoding is performed per TR, reducing the scan time.

Mechanism

TSE employs a train of 180° rephasing pulses, each one producing a spin echo. This train of spin echoes is called an **echo train**. The number of 180° RF pulses and resultant echoes is called the **echo train length (ETL)** or **turbo factor**. The spacing between each echo is called the **echo spacing**.

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After each rephasing, a phase-encoding step is performed and data from the resultant echo is stored in a different line of K space (k-space is an array of numbers representing spatial frequencies in the MR image) (Figure 14.1). Therefore, several lines of K space are filled every TR instead of one line as in conventional spin echo. As K space is filled more rapidly, the scan time decreases.

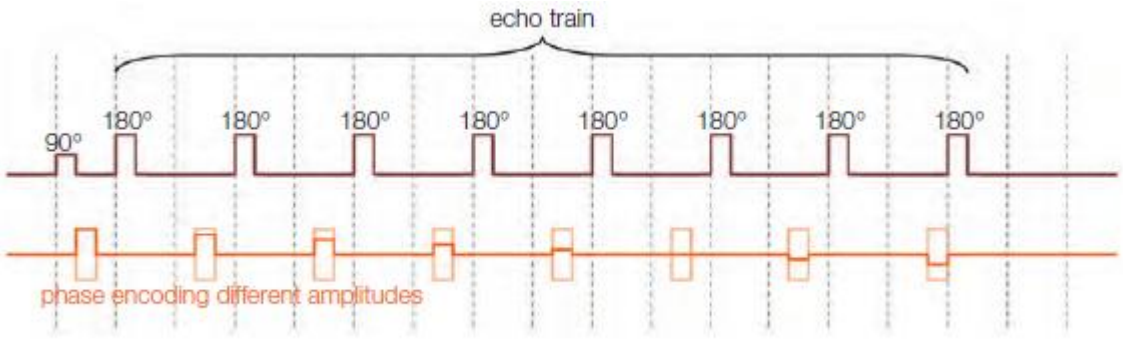


Figure 14.1 The echo train in TSE.

Typically, 2 to 30 180° RF pulses are applied during every TR, although many more can be applied if required. As several phase encodings are also performed during each TR, several lines of K space are filled each TR and the scan time is reduced. For example, if a factor of 16 is used, 16 phase encodings are performed per TR and therefore 16 lines of K space are filled per TR instead

of 1 as in conventional spin echo. Therefore, the scan time is 1/16 of the original scan time (Table 14.1). The *higher* the turbo factor the *shorter* the scan time (Table 14.2).

Table 14.1 TSE time-saving illustrations.

Pulse sequence	Scan time
SE, 256 phase encodings, 1NSA	256 × 1 × TR = 256 × TR
TSE, 256 phase encodings, 1 NSA, ETL 16	256 × 1 × TR/16 = 16 × TR

Table 14.2 Equations of TSE scan time.

Equations		
ST = TR × Matrix(P) × NSA/ ETL	ST is the scan time (s) TR is the repetition time (ms) Matrix(P) is the phase matrix NSA is the number of signal averages ETL is the echo train length or turbo factor	This equation enables the scanner to calculate the scan time in TSE. The longer the echo train, the shorter the scan time, but may result in fewer slices per TR

Contrast

Each echo has a different TE and data from each echo is used to produce one image. This is different from CSE, where several echoes may be generated with a different TE but each echo is used to produce a *different* image. In TSE multiple echoes with a different TE are used to produce the *same* image. This would normally result in a mixture of weighting. In TSE this problem is overcome by using **phase reordering.**

In any sequence, each phase-encoding step applies a different slope of phase gradient to phase shift each slice by a different amount. This ensures that data is placed in a different line of K space.

The very *steep* gradient slopes significantly *reduce the amplitude* of the resultant echo/signal, because they reduce the rephasing effect of the 180° rephasing pulse. *Shallow* gradients, on the other hand, do not have this effect and the *amplitude of the resultant echo/signal is maximized* (Figure 14.2).

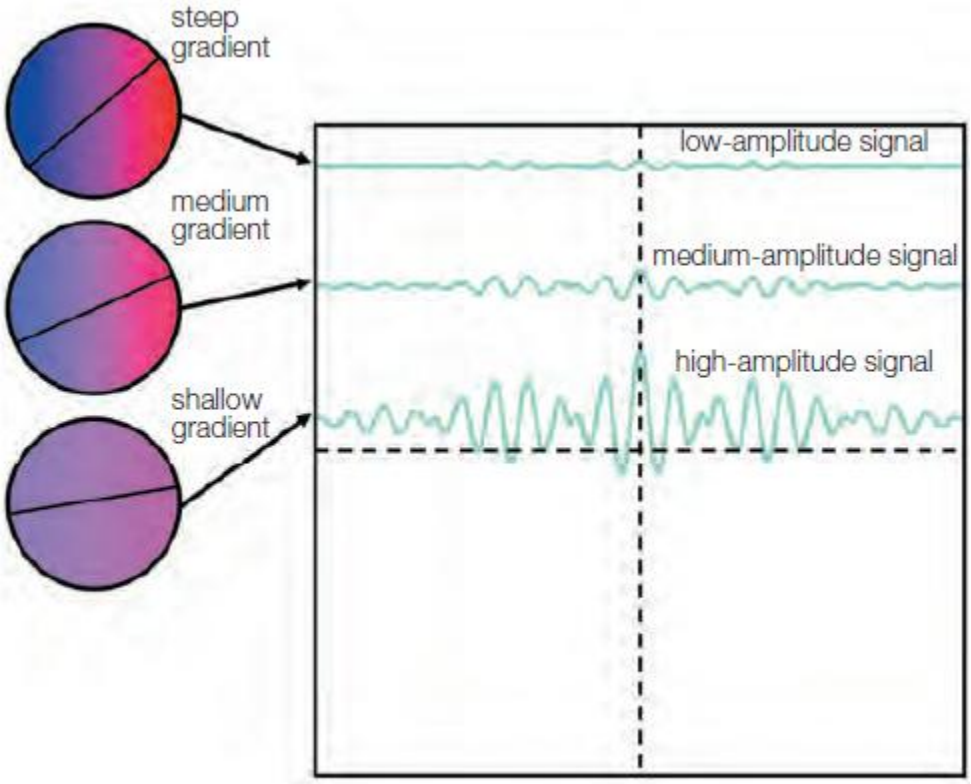


Figure 14.2 Phase encoding versus signal amplitude.

When the TE is selected (known as the **effective TE** in TSE sequences) the resultant image must have a weighting corresponding to that TE; that is, if the TE is set at 102 ms a T2 weighted image is obtained (assuming the TR is long).

The system therefore orders the phase encodings so that those that produce the most signal (the shallowest ones) are used on echoes produced from 180° pulses nearest to the effective TE selected. The steepest gradients (which reduce the signal) are reserved for those echoes that are produced by 180° pulses furthest

away from the effective TE. Therefore, the resultant image is mostly made from data acquired at approximately the correct TE, although some other data is present (Figure 14.3).

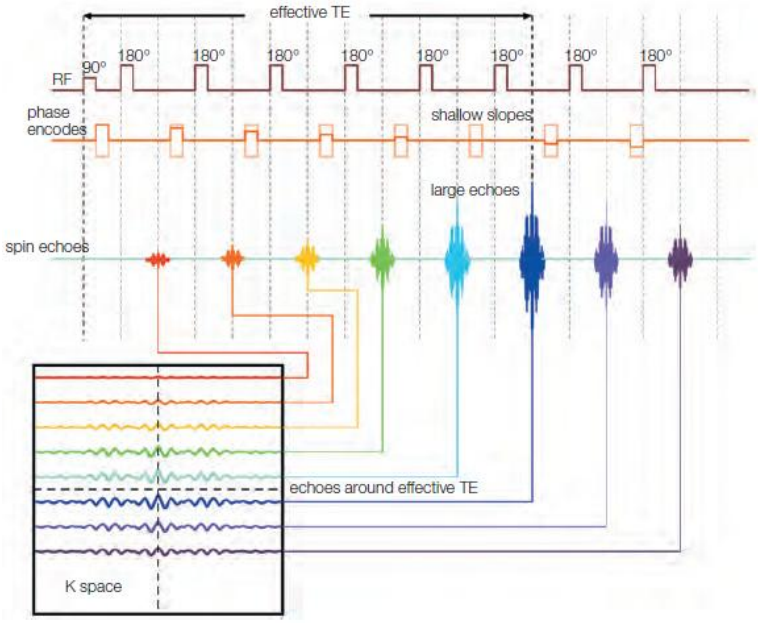
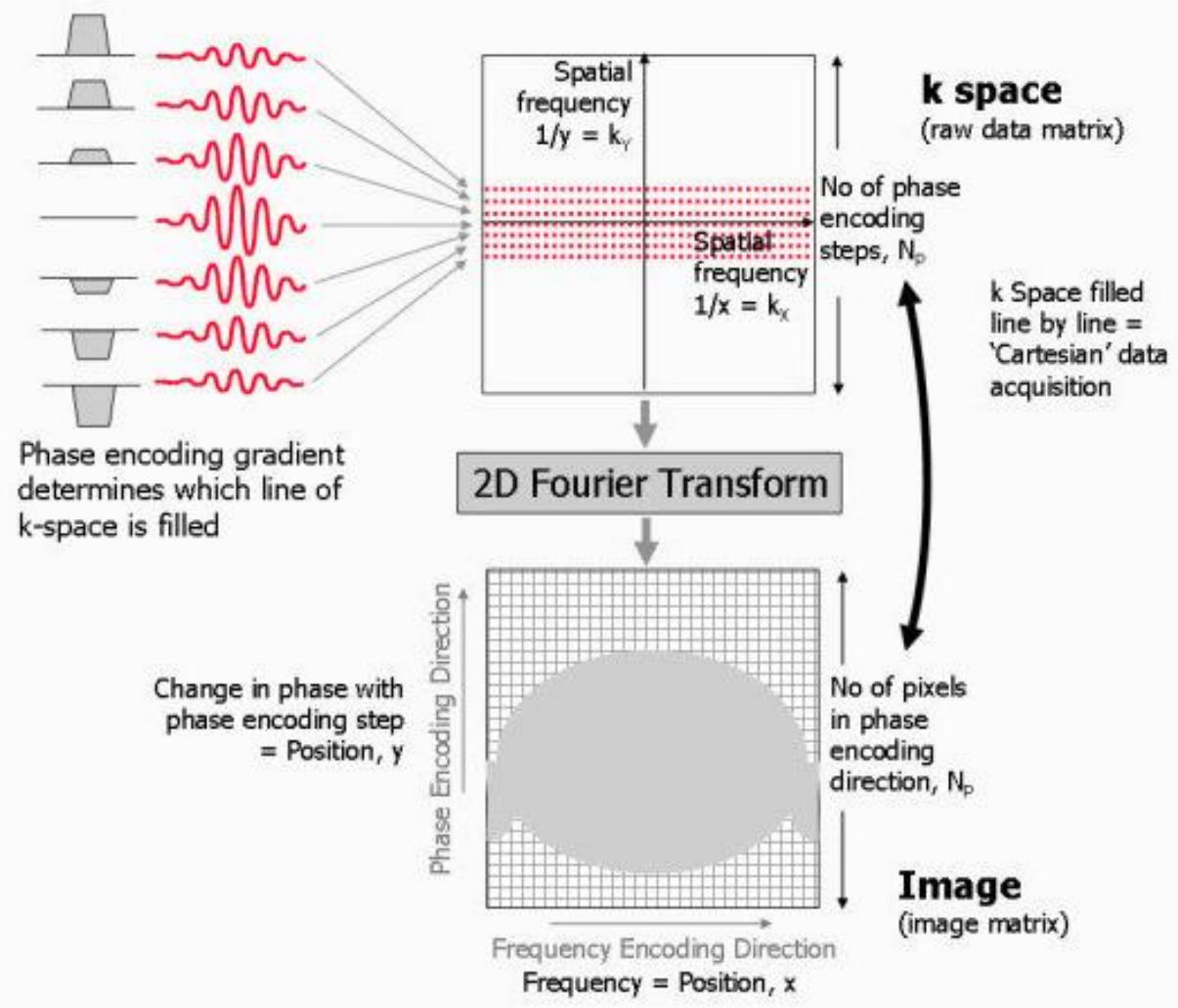


Figure 14.3 K space filling and phase reordering.

On many modern scanners it is possible to reduce the magnitude of the RF rephasing pulse from 180° (e.g. 150°). Rephasing still occurs, because RF energy is delivered at the Larmor frequency, but, as the amplitude of the RF is less, SAR is reduced. Reduction in SAR allows for more slices for a given TR.



KEY POINTS

- ✓ Turbo or fast spin echo sequences involve applying the phase encoding gradient multiple times in a TR period to varying amplitudes and polarity.
- ✓ This means that multiple lines of K space are selected per TR. The number is equal to the echo train length (ETL) or turbo factor.
- ✓ Multiple echoes are produced by multiple applications of an RF rephasing pulse and data from each echo is placed in a different line of K space.

- ✓ Scan times are reduced by a factor equal to the turbo factor or ETL.
- ✓ Image weighting is controlled by phase reordering so that data collected from echoes at or around the effective TE are placed in the signal and contrast areas of K space.
- ✓

MRI DESIGN: 15. Fast or turbo spin echo – how it is used

Due to different contrasts being present in the image, the contrast of TSE is unique. In T2 weighted scans, water and fat are hyperintense (bright). This is because the succession of 180° RF pulses reduces the spin-spin interactions in fat, thereby increasing its T2 decay time. Techniques such as STIR (short TI inversion recovery) and chemical pre-saturation that suppress fat signal are therefore usually required to differentiate fat from pathology in T2 weighted TSE sequences.

Muscle is often darker than in conventional spin echo T2 weighted images. This is because the succession of RF pulses increases magnetization transfer effects that produce saturation. In T1 weighted imaging, CNR is sometimes reduced so that the images look rather 'flat'. It is therefore best used when inherent contrast is good.

When used with a very long echo train, TSE can sometimes result in images that are blurred. This is particularly the case when combined with a long echo spacing value. Echoes with a *very long TE* are likely to have *low signal amplitude* because of T2 decay. If data from these small echoes is mapped into the resolution lines of K-space, *image blurring* can occur.

This is usually only a problem with a very long echo train, however. The number of slices is determined by the ETL and the echo spacing (see Table 15.1). Extending the TR lengthens the scan time, but this is more than compensated for by the use of long echo trains.

Table 15.1 Equations of TSE.

Equations		
$N\ slices = \frac{TR}{ETL \times E_S}$	N slices is the number of slices allowed per TR TR is the repetition time (ms) ETL is the echo train length or turbo factor E_S is the echo spacing (ms)	This equation shows how many slices are allowed in TSE and will be less than in conventional spin echo (see Scanning Tip 1).

Typical values

Dual echo	***Single echo T2 weighting***	***Single echo T1 weighting***
• TR: 2500–8000ms (for slice number) • Effective TE1: 17ms • Effective TE2: 102ms • Turbo factor 8: this may be split so that the PD image is acquired with the first four echoes and the T2 with the second four echoes	• TR: 4000–8000 ms • TE: 102 ms • Turbo factor: 20+	• TR: 600 ms • TE: 10 ms • Turbo factor: 4

Uses

TSE produces T1, T2 or proton density scans in a fraction of the time of CSE (Figures 15.1 and 15.2). Due to the fact that the scan times are reduced, phase matrix size can be increased to improve spatial resolution. TSE is normally used in the brain, spine,

joints, extremities and pelvis. As TSE is incompatible with phase-reordered respiratory compensation techniques, it can only be used in the chest and abdomen with respiratory triggering, breath-hold or multiple NSA.

Systems that have sufficiently powerful gradients can use TSE in a single-shot mode **(SS_TSE)** or half Fourier single-shot TSE **(HASTE)**. Both of these techniques permit image acquisition in a single breath-hold. In addition, using very long TEs and TRs permits very heavy T2 weighting (**watergrams**). An example of this technique is in gallbladder imaging, where only signal from bile in the biliary system is seen. Table 15.2 lists some advantages and disadvantages of TSE.

A modification of TSE that is sometimes called **fast recovery** or **DRIVE** adds an

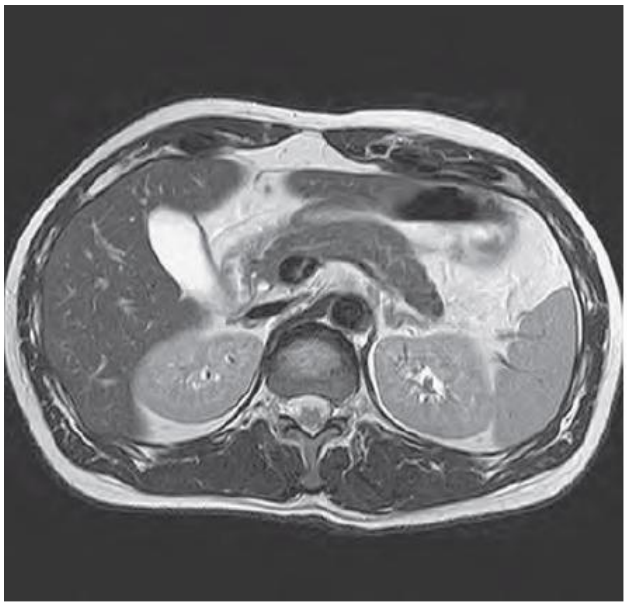


Figure 15.1 Axial T2 weighted TSE image of the abdomen.

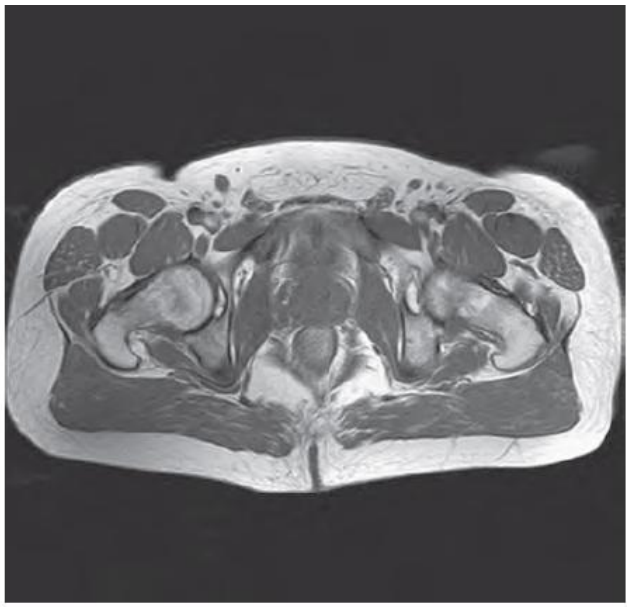


Figure 15.2 Axial T1 weighted TSE image of the male pelvis.

additional 'reset' pulse at the end of the TR period. This pulse 'drives' any residual magnetization in the transverse plane at the end of each TR back into the longitudinal plane (Figure 15.3).

This is then available to be flipped into the transverse plane by the next excitation pulse. This sequence provides high signal intensity in water even when using a short TR and therefore a short scan time (Figure 15.4). This is because water has a long T2 decay time; therefore, tissue with a high-water content has residual transverse magnetization at the end of each TR. Hence this is the main tissue that is driven back up to the longitudinal plane by the reset pulse and is therefore the dominant tissue in terms of signal.

Table 15.2 Advantages and disadvantages of TSE.

Advantages	Disadvantages
Short scan times High-resolution imaging Increased T2 weighting Magnetic susceptibility decreases*	Some flow artefacts increased Incompatible with some imaging options Some contrast interpretation problems Image blurring possible

* Also a disadvantage, e.g. haemorrhage not detected/delineated.

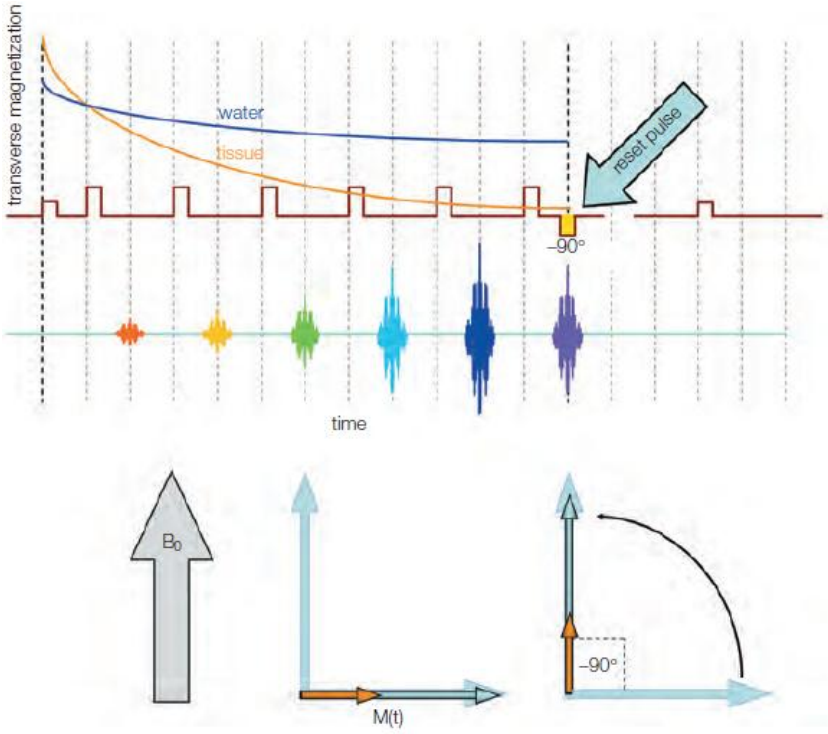


Figure 15.3 The fast recovery or 'DRIVE' sequence.

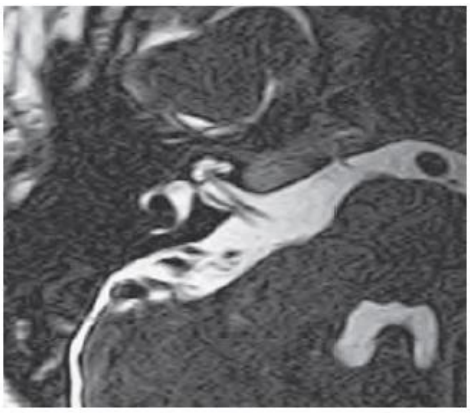


Figure 15.4 Fast recovery or 'DRIVE' image of the internal auditory meatus.

KEY POINTS

✓ In turbo spin echo sequences fat remains bright in T2 weighted images due to J coupling. Fat suppression techniques are commonly employed.

✓ The turbo factor or echo train length is an extrinsic contrast parameter unique to this sequence.

✓ Short turbo factors or echo train lengths are necessary for T1 and PD weighting so that echoes with long TEs do not affect image contrast.

✓ A long turbo factor or echo train length is needed for T2 weighting so that echoes with a long TE can affect contrast.

✓ The longer the echo train, the shorter the scan time.

✓ Turbo or fast spin echo has many applications in most body areas.

HW:

- In T2 weighted scans, water and fat are hyperintense (bright). Why?
- Muscle is often darker than in conventional spin echo T2 weighted images. Why?
- When used with a very long echo train and long TE, TSE can sometimes result in images that are blurred. Why?
- Extending the TR lengthens the scan time, but this is more than compensated for by the use of long echo trains. Discuss!

Lecture Seven

MRI DESIGN: MR safety and bio-effects

Static magnetic field bio-effects

Current guidelines recommend a maximum limit of 8T for clinical imaging, rising to 12T for research purposes and spectroscopy. Most clinical units operate below 3T.

The following points are *fundamentally* important with regard to the potentially harmful effects of the static magnetic field. The static field is always present (24 hours a day, 365 days a year, to infinity). It is switched on even when the system is out of use (Figure 54.1). The fringe field may extend several metres beyond the examination room and therefore any harmful effects or risks may come into play at some distance from the scanner (Figure 54.2).

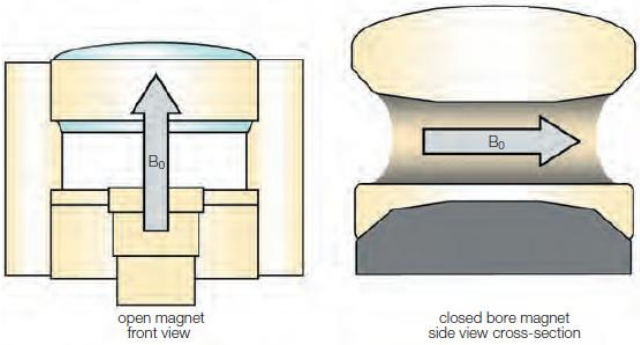


Figure 54.1 Static field in permanent and superconducting systems.

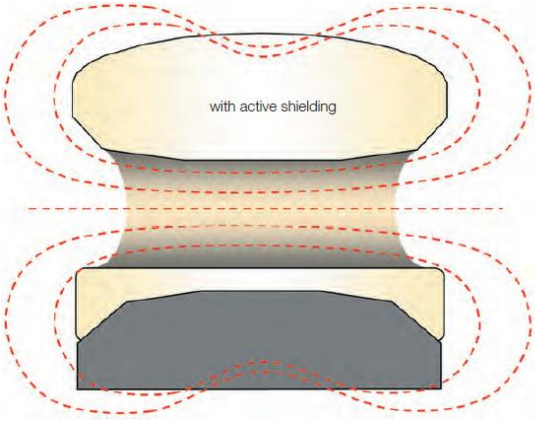


Figure 54.2 The fringe field.

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There is no conclusive evidence for irreversible or harmful bio-effects in humans below 2.5T. Reversible abnormalities may include:

• an increase in the amplitude of the T-wave that can be noted on an ECG due to the magnetic hydrodynamic effect (also known as the magnetic haemodynamic effect);

• heating of patients; • fatigue; • headaches; • hypotension;

• irritability.

Time-varying field bio-effects

Gradients create a time-varying magnetic field. This changing field occurs during the scanning sequence. It is not present at other times and therefore exposure is restricted to patients and to relatives who may be present in the scan room during the examination.

The health consequences are not related to the strength of the gradient field, but to changes in the magnetic field that cause induced currents. Nerves, blood vessels and muscles, which act as conductors in the body, may be affected. The induced current is greater in peripheral tissues, since the amplitude of the gradient
is higher away from magnetic isocentre. Time-varying bio-effects from gradient coils include:

• light flashes in the eyes; • alterations in the biochemistry of cells and fracture union;

• mild cutaneous sensations;

• involuntary muscle contractions; • cardiac arrhythmias.

RF transmit coils also produce time-varying fields. The predominant bio-effect of RF irradiation absorption is the potential heating of tissue. As an excitation pulse is applied, some nuclei absorb the RF energy and enter the high-energy state. As they relax,

nuclei give off this absorbed energy to the surrounding lattice. As excitation frequency is increased, absorbed energy is also increased, therefore heating of tissue is largely frequency dependent.

Energy dissipation can be described in terms of **specific absorption rate** or **SAR**. SAR is expressed in watts per kilogram (W/kg), a quantity that depends on:

- induced electrical field;
- pulse duty cycle;
- tissue density;
- conductivity;
- the size of the patient.

SAR is used to calculate an expected increase in body temperature during an average examination (Table 54.1). In the UK, it is recommended that this should not exceed 1°C during the examination. Studies show that patient exposure up to three times the recommended levels produces no serious adverse effects, despite elevations in skin and body temperatures. As body temperature increases, blood pressure and heart rate also increase slightly. Even though these effects seem insignificant, patients with compromised thermoregulatory systems may not be candidates for MRI.

Table 54.1 SAR limits in the USA.

Area	Dose	Time (mins)	SAR (W/kg)
Whole body	averaged over	15	4
Head	averaged over	10	3
Head or torso	per gram of tissue	5	8
Extremities	per gram of tissue	5	12

Radiofrequency fields can be responsible for significant burn hazards due to electrical currents that are produced in conductive loops. Equipment used in MRI, such as ECG leads and surface coils, should therefore be employed with extreme caution. When using a surface coil, the operator must be careful to prevent any electrically

conductive material (e.g. cable of surface coil) from forming a 'conductive loop' with itself or with the patient.

Site planning

There have been a number of fatal accidents in the MR environment. It is therefore vital that access to the MRI system and the magnetic field is controlled. The American College of Radiologists has produced a White Paper that recommends that all centres define the following zones (Figure 54.3).

Zone I includes all areas that are accessible to the public. All personnel are allowed in Zone I.

Zone II is the interface between Zone I and the controlled Zone III. There must be a lock or warning signs between Zones I and II. All personnel are allowed in Zone II, but a trained 'gatekeeper' should be present to prevent patients and non-MR personnel from inadvertently entering Zones III and IV.

Zone III is strictly restricted because free access by unscreened personnel and ferromagnetic objects may cause death or serious injury. This area must be strictly monitored and only MR-trained personnel and screened patients are permitted in this area.

Zone IV is only suitable for screened patients under direct and constant supervision from MR-trained personnel, as death and serious injury can occur. The patient may also experience heating, missile effects, RF antenna effects and anoxia in this zone.

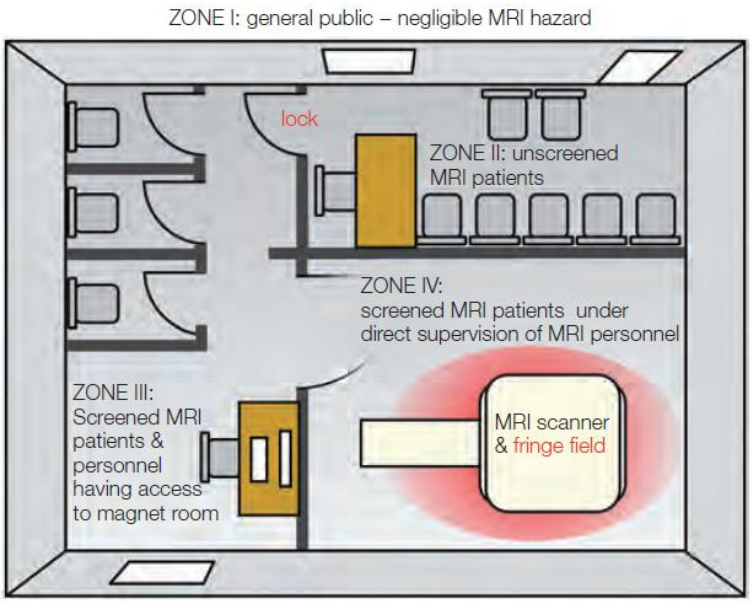

Figure 54.3 The zoning recommended by the American College of Radiology White Paper on MRI safety. Note that there has to be locked access between Zones II and III.

Key points.

- The magnetic field is present 24 hours a day, 365 days a year.
- There is no conclusive evidence that the static magnetic field is harmful up to 2.5T.
- Time-varying fields imposed by the gradients can have effects, especially when using fast sequences.
- RF pulses can cause heating. The SAR measures how much energy the patient's tissues absorb during the scan.
- Site planning requires the establishment of clearly marked zones.

MRI DESIGN: MR safety – projectiles

The projectile effect of a metal object exposed to the field can seriously compromise the safety of anyone sited between the object and the magnet system. *The potential harm cannot be over-emphasized.* In many ways the MR scan room is the most dangerous room in a hospital or imaging facility, because it is possible to seriously injure or even kill someone in a second. Even small objects such as paperclips and hairpins have a terminal velocity of 40 mph when pulled into a 1.5 T magnet, and therefore pose a serious risk to the patient and anyone else present in the scan room. Larger objects such as scissors travel at much higher velocities and may be fatal to any person in their path (Figure 55.1).

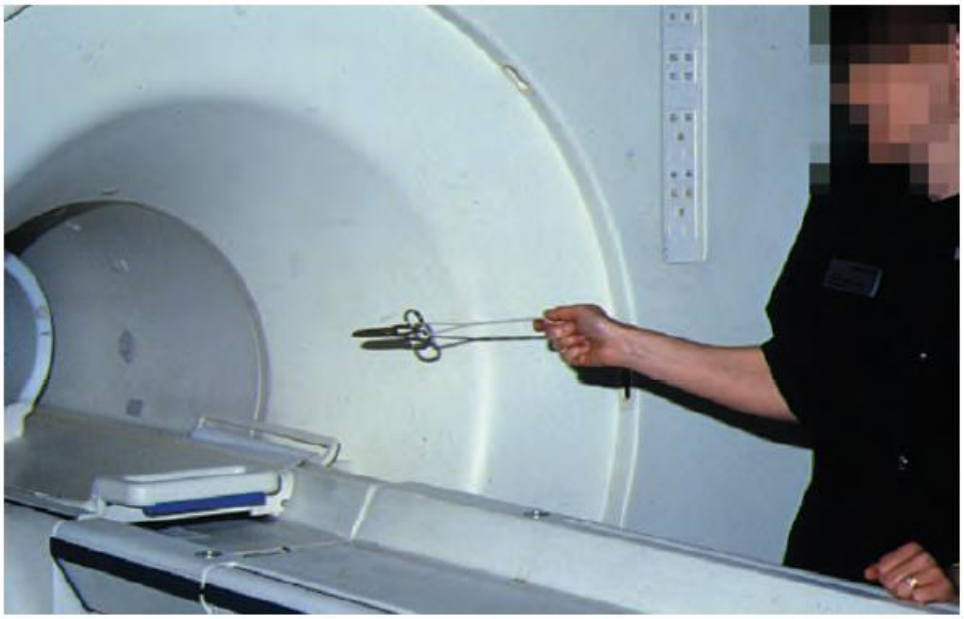

Figure 55.1 The pulling power of a pair of scissors in a 1.5 T system.

Many types of clinical equipment are ferromagnetic and should **never** be brought into the scan room. These include surgical tools, scissors, clamps and oxygen tanks.

Quenching

If an accident occurs where a patient or other person in the scan room is pinned to the magnet by a projectile that cannot be removed by hand, the magnetic field must be immediately quenched. Quenching is the process whereby there is a sudden loss of absolute zero of temperature in the magnet coils, so that they cease to be superconducting and become resistive. The magnetic field is therefore lost. Quenching can be initiated on purpose, usually by pressing a quench button in the control room, or it may happen accidently. Quenching causes helium to escape from the cryogen bath extremely rapidly. Quenching may cause severe and irreparable damage to the superconducting coils, and so all systems should have helium-venting equipment, which removes the helium to the outside environment in the event of a quench. However, if this fails, helium vents into the room and replaces oxygen. For this reason, all scan rooms should contain an oxygen monitor that sounds an alarm if the oxygen falls below a certain level.

Metallic implants and prostheses

Devices are tested on an ongoing basis for MR safety and many manufacturers have developed MR-safe devices. It is therefore vital to check the type of device or implant and whether it is safe before booking the appointment. Metallic implants and prostheses produce serious effects, which include torque or twisting in the field, heating effects and artefacts on MR images. The type of metal used in such implants is one factor that determines the force exerted on them in magnetic fields. While non-ferrous metallic implants may show little or no deflection to the field, they could cause significant heating due to their inability to dissipate the heat caused by radiofrequency absorption. Devices that might need to be taken into the scan room must be tested beforehand. There are

standard labels depending on whether the device is safe, unsafe or conditional on the field strength (Figure 55.2).

Figure 55.2 Standard labels associated with MR device testing.

What is not safe to scan?

Cochlear implants are attracted to the magnetic field and are magnetically or electronically activated. They are, therefore, unsafe to scan. It is not uncommon for patients who have worked with sheet metal to have metal fragments or slivers located in and around the eye. Since the magnetic field exerts a force on ferromagnetic objects, a metal fragment in the eye could move or be displaced and cause injury to the eye or surrounding tissue. Therefore, all patients with a suspected eye injury must be X-rayed before the MR examination.

Aneurysm clip motion may damage the vessel, resulting in haemorrhage, ischaemia or death. Currently, many intracranial clips are made of a non-ferromagnetic substance such as titanium. However, some of these may still deflect in a magnetic field. It is therefore recommended that imaging of patients with aneurysm clips is delayed, until the type of clip is emphatically identified as non-ferrous and non-deflectable. Intracranial

clips also cause severe magnetic susceptibility artefact, especially in gradient echo sequences.

Although MR-safe pacemakers have been developed, it is important to assume that most patients do not have this kind of pacemaker. Even field strengths as low as 10 gauss may be sufficient to cause deflection, programming changes, and closure of the reed switch that converts a pacemaker to asynchronous mode. Patients who have had their pacemaker removed may have pacer wires left within the body that could act as an antenna and (by induced currents) cause cardiac fibrillation.

What is probably safe to scan?

Prosthetic heart valves are deflected by the static magnetic field, but this is minimal compared to normal pulsatile cardiac motion. Although patients with most valvular implants are considered safe for MR, careful screening for valve type is advised.

Most orthopaedic implants show no deflection within the main magnetic field. A large metallic implant such as a hip prosthesis can become heated by currents induced in the metal by the magnetic and radiofrequency fields. It appears, however, that such heating is relatively low. The majority of orthopaedic implants have been imaged with MR without incident.

Abdominal surgical clips are generally safe for MR because they become anchored by fibrous tissue, but produce artefacts in proportion to their size and can distort the image.

Key points.

- The MR scan room can be considered the most dangerous room in the hospital or imaging facility.

- Objects are accelerated towards the magnet at very high velocity depending on their mass, the strength of the magnetic field and what they are made of.
- Anyone entering the scan room must have been thoroughly screened and checked by a trained MR practitioner.
- Implants and any devices being taken into the scan room must be checked beforehand.

Q: The heating of tissue is largely frequency-dependent. Discuss!!!