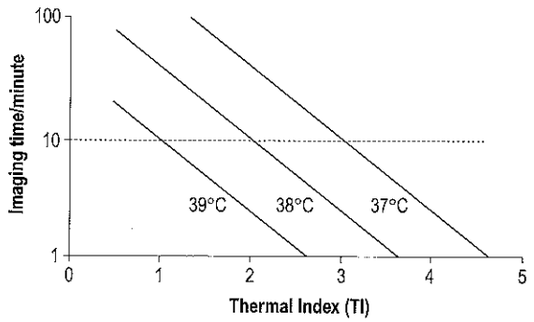

Figure 9.27 Safe imaging times at the indicated thermal index for different body temperatures.

display panel when the equipment itself can exceed an index value of 1. Index values less than 0.5 are below the threshold level for any effect and are considered safe. Above this level, the risks should be considered. The exposure time should be reduced appropriately as the TI increases above the value of 1. The approximation in Figure 9.27 shows that with a TI of 3 indicated, a normal patient may be scanned for 10 min, whereas for a patient running a temperature of 39°C, scanning for even less than 1 min will imply a non-safe situation. Thus particular care is needed when the patient is feverish and when using pulsed Doppler systems.

9.16 SUMMARY

- Ultrasound is inaudible sound (frequencies above 20 kHz).
- Ultrasound waves are produced by a transducer (a piezoceramic) that converts an electrical signal into an ultrasound beam. The transducer also acts as the receiver.
- Clinical ultrasound, having a short wavelength (frequencies from 3.5 to 15 MHz), can be formed into a narrow beam.
- Ultrasound is not electromagnetic radiation, but like light it does undergo reflection and refraction at the interface between two different media.
- Clinical ultrasound produces images from echoes reflected from internal tissue structures and interfaces.
- The echo return time is proportional to the depth of the structure, and the intensity depends on the tissue composition.
- Computer analysis of the transmitted and received signals from sector scanners enables anatomical displays in real time.
- Mechanical scanners generally produce better images, but electronic scanners are easier to handle and are more robust.
- Soft tissues that are too similar to be distinguished by planar X-ray contrast can be imaged.
- Microbubbles or nanoparticles can be used as contrast media.
- Ultrasound does not pass easily across tissue–air or tissue–bone interfaces, so lung and intracranial images are not generally practical.
- The Doppler effect is used to image the direction and velocity of blood flow.
- Signal to noise ratio and image quality improve as better instrumentation is developed.
- Most noise is electronic from the very small currents measured; additional noise comes from reverberations in the patient or in the transducer probe.
- Ultrasound, at normal diagnostic intensity levels, has no known deleterious effects and so is considered safe when indicated for fetal imaging.

Chapter 10

Magnetic resonance imaging

CHAPTER CONTENTS

Magnetic resonance imaging (MRI) uses radiowaves and magnetic fields. Although the nuclei of all atoms contain protons, only those with an odd number possess the property called nuclear magnetic resonance. Hydrogen has a single proton and thus a large magnetic moment, and it is abundant in the body, in water (free or attached to other molecules) and in fat, and so provides the best MRI signals.

The patient is placed in a magnet and a radiowave is sent through the body. The transmitter is turned off and the patient re-emits radiowaves, which are received and used for reconstruction of the image. It is the nuclei of hydrogen atoms that absorb and emit the radio-frequency (RF) energy. MRI measures the hydrogen content of individual voxels in each transverse slice of the patient and represents it as a shade of grey or colour in the corresponding image pixel on the screen.

10.1 THE SPINNING PROTON

Every proton has a positive charge and spins continually like a top around an axis called the spin vector. The circulating charge is like a small loop of current, and each proton acts like a bar magnet or dipole. Its magnetic moment $\boldsymbol{m}$ is represented by a vector joining the north and south poles, drawn as an arrow in Figure 10.1. Normally, all the individual dipoles point in a random fashion, with equal numbers in every direction. The net magnetic effect is then zero. (This ignores the tiny effect on them of the earth's magnetic field of about 50 μT.)

The patient lies prone or supine in a solenoid coil (see section 10.10 for details of equipment) carrying a direct current (DC). This produces a very uniform and strong magnetic field inside the coil, represented by a vector $\boldsymbol{B}$ pointing along the axis of the coil (and

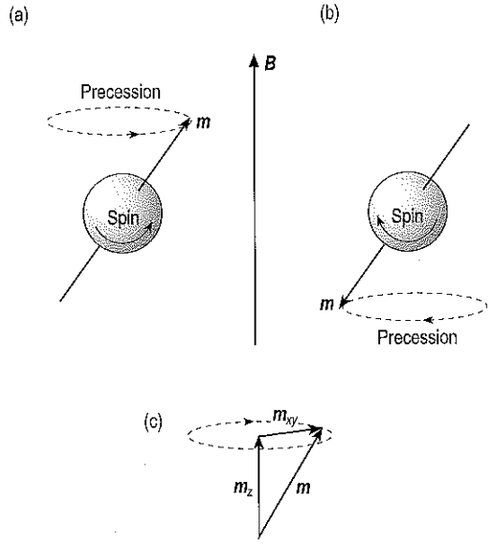

Figure 10.1 The magnetic vectors associated with a spinning proton precessing (a) parallel and (b) antiparallel to a magnetic field, *B*. (c) The transverse and longitudinal components of the magnetic vector.

the length of the patient, see Fig. 10.2). This is taken as the Z-axis; the *Y*-axis runs vertically from top to bottom and the *X*-axis horizontally across the machine. The magnetic field strength has a set value usually between 0.15 and 3 T, depending on the machine. By way of example, throughout this chapter we will take a machine with a field strength of 1 T. This is some 20 000 times greater than the earth's magnetic field.

Inside the coil, the patient becomes very slightly magnetized. The static magnetic field $\boldsymbol{B}$ causes the magnetic dipoles to turn and point along the Z-axis in one of two stable directions – either, as in Figure 10.1a, in the direction of the field (parallel or 'spin up') or, as in Figure 10.1b, in the opposite direction (antiparallel or 'spin down').

As it takes less energy to align with the magnetic field than to oppose it, slightly more dipoles point spin up than spin down. MRI depends on detecting this small difference, which is proportional to $\boldsymbol{B}$ and amounts to about three out of each million protons at 1 T.

Most of the dipoles, each with a magnetic moment $\boldsymbol{m}$, cancel each other out in pairs (parallel and antiparallel), leaving those not so paired to produce a combined, longitudinal net or bulk magnetic vector $\boldsymbol{M_z}$ in the direction of *B*.

Henceforth, the terms *spins* or *protons* will refer only to the *detectable* protons, the excess of spin-up over spin-down protons, and we will ignore the others. For example, in a cubic millimetre of water

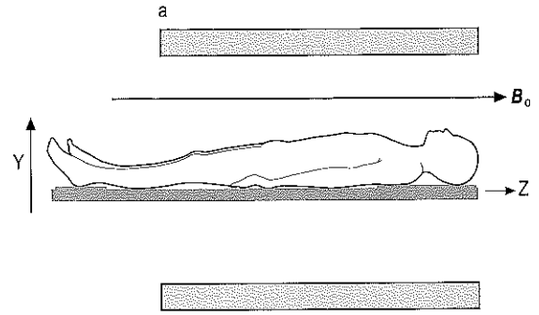

Figure 10.2 The patient on the scanner table inside the main solenoid coil, a, with magnetic field *B* in the horizontal *Z*-direction from toe to head. The *Y*-direction is vertical and the *X*-direction lateral or transverse through the patient.

there are about 7×10^{19} protons, of which only some 2×10^{14} will be detectable.

Precession

The static field also causes the spinning protons to 'wobble' in a regular manner called precession (Fig. 10.1a,b). The direction of the spin axis tilts and rotates around the direction of the magnetic field $\boldsymbol{B}$ with a fixed frequency (millions of revolutions s^{-1}), called the Larmor frequency. Figure 10.1 illustrates the difference between spin, precession and the magnetic moment $\boldsymbol{m}$ of a single proton.

This is similar to the way in which the north–south axis of the earth precesses once in 25 000 years because of the gravitational pull of the sun, and a spinning top or gyroscope precesses because of the earth's gravitational field.

The tilting of the spin axis of a precessing proton splits its magnetic vector $\boldsymbol{m}$ into a longitudinal component $\boldsymbol{m_z}$ that points in the Z-direction, and a transverse component $\boldsymbol{m_{xy}}$ that rotates in the *XY* plane (Fig. 10.1c).

Now consider all the detectable protons in a single voxel of tissue (size a few cubic millimetres, illustrated in Fig. 10.3). The $\boldsymbol{m_z}$ vectors all point in the Z-direction and add up to a combined or net longitudinal magnetism $\boldsymbol{M_z}$ (Fig. 10.3a). This cannot be measured directly, as it points in the same direction as $\boldsymbol{B}$. As the protons precess independently, their $\boldsymbol{m_{xy}}$ vectors point in all directions and cancel out. The net transverse magnetism $\boldsymbol{M_{xy}} = 0$.

The stronger the magnetic field, the faster a proton precesses. The frequency of precession (f) or Larmor frequency is proportional to the product of:

- the magnetic field strength, and
- a constant property of the nucleus called the gyromagnetic ratio γ.

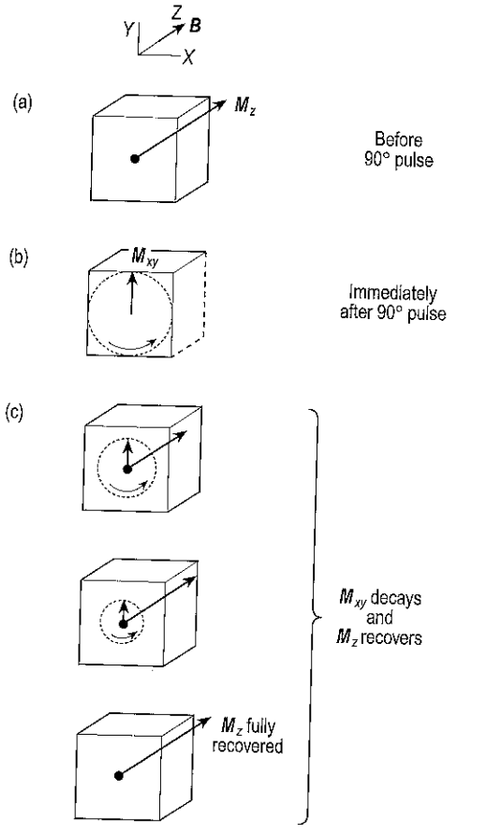

Figure 10.3 The net magnetization in a tissue voxel: a time sequence following a 90° pulse.

For hydrogen nuclei in a field of 1 T, f = 42.6 MHz. This is an RF and has a very precise value (in water, within ±0.1 Hz).

In the quantum theory, a frequency of 42.6 MHz corresponds to a quantum energy of 0.2 μeV. The energy of the antiparallel state is therefore 0.2 μeV greater than that of the parallel state. It is because of this small energy difference that slightly more spins line up parallel rather than antiparallel.

Radiofrequency coils

Surrounding and close to the patient are a set of RF coils that inject an RF pulse in a direction perpendicular to $\boldsymbol{B}$. This has two effects:

- some or all of the spin-up protons pick up energy, turn spin down, and are said to be excited. This affects $\boldsymbol{M_z}$; it may be reduced, disappear or even reverse, depending on the length and strength of the RF pulse.
- the protons are pulled into synchronism, and they now precess in step or in phase. Their $\boldsymbol{m_{xy}}$ vectors add up to a transverse magnetic vector $\boldsymbol{M_{xy}}$, which rotates in the *XY* plane (Fig. 10.3b) at the Larmor frequency.

Resonance

It is well known that an opera singer must strike precisely the right note to shatter a wine glass. Similarly, in order to affect the dipoles, the frequency of the RF generator must very accurately match the Larmor precession or resonant frequency of the dipoles. In other words, the photon energy of the radiowaves must be exactly the same as the energy difference between spin-up and spin-down protons.

180° pulse

An RF pulse of a certain total energy will give to each and every dipole exactly the energy (0.2 μeV) required to tip them through 180°. This temporarily reverses the net magnetic vector $\boldsymbol{M_z}$.

90° pulse

An RF pulse of half that total energy (i.e. half the intensity or half the duration) will tip half of the dipoles so that equal numbers point spin up and spin down, thus reducing $\boldsymbol{M_z}$ to zero. The RF pulse also causes them to move into the same phase and precess together (phase coherence).

This phase coherence of the dipoles produces a transverse magnetism $\boldsymbol{M_{xy}}$, perpendicular to $\boldsymbol{B}$, which rotates in the *XY* plane at the Larmor frequency (Fig. 10.3b). It is as if the 90° pulse has tipped the magnetic vector $\boldsymbol{M_z}$ through 90°. MRI involves sending a series of many such pulses, repeated at intervals of *TR* seconds, called the repetition time.

10.2 THE MAGNETIC RESONANCE SIGNAL

When the 90° pulse is over, the magnetic vector $\boldsymbol{M_{xy}}$ continues for a while to rotate in the transverse *XY* plane. Just like the rotating magnet in a dynamo, it induces in the RF coil an alternating (RF) voltage of a few microvolts. After amplification by an RF amplifier (receiver), tuned like a radio to the resonant frequency, the amplitude or envelope of this signal is sampled, digitized and computer analysed. Using methods of spatial encoding and signal processing described in section 10.4, the magnetic resonance (MR) signal from each individual voxel in the scan matrix can be identified to produce the pixel grey or colour level in the MR image.

Note that only $\boldsymbol{M_{xy}}$ produces an MR signal; $\boldsymbol{M_z}$ does not. But because $\boldsymbol{M_{xy}}$ is produced by tipping $\boldsymbol{M_z}$, the

signal produced by a 90° pulse depends on the value of M_z immediately before that pulse is applied.

The peak signal is proportional to, and the pixel brightness depends on:

- proton or spin density (PD, number of protons per cubic millimetre) of the voxel
- the gyromagnetic ratio of the nucleus
- the static field strength B, because placing the patient in a stronger magnetic field increases the preponderance of protons that are initially spin up over those that are spin down.

Only mobile protons give signals; those in large molecules or effectively immobilized in bone do not. The greater part of the signal is due to body water, whether free or bound to molecules. Air, in sinuses for example, having no hydrogen, produces no signal and always appears black in the image. Fat has a higher PD than other soft tissues. Grey matter has a somewhat greater PD than white matter. Tissues do not, however, vary greatly in their proton densities.

Free induction decay

The MR signal is greatest immediately after the brief 90° pulse has been switched off. Thereafter, the dipoles are free to return, some earlier than others, to their original orientation. As indicated in Figure 10.3c, M_z regrows or 'recovers' while M_{xy} decreases or 'decays', and, accordingly, the strength of the MR signal induced in the receiver coil also decays, although its frequency remains the same.

At any instant of time, M_z and M_{xy} combine to produce a *sum* vector M. The 90° pulse has tipped the sum vector through 90° into the XY transverse plane. Then, during relaxation, as M_z increases and M_{xy} decreases, the sum vector spirals (beehive fashion, Fig. 10.4) back from the transverse plane to the longitudinal Z-direction. This is caused by two concurrent and quite independent methods of energy loss or 'relaxation': spin–lattice and spin–spin relaxation.

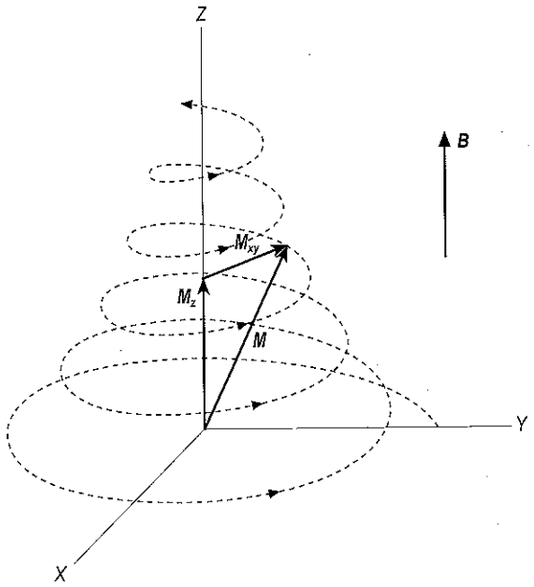

Figure 10.4 Behaviour of the sum magnetic vector M during free induction decay.

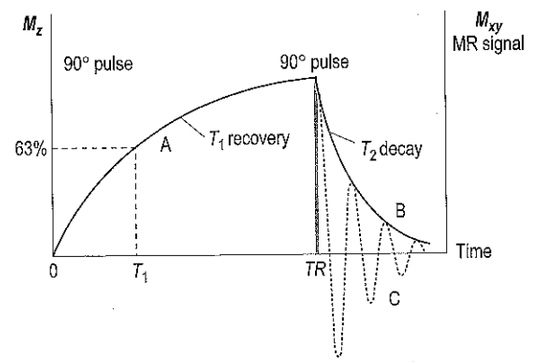

Figure 10.5 T_1 recovery of M_z and T_2 decay of M_{xy}. MR, magnetic resonance.

Spin–lattice relaxation or T_1 recovery

The dipoles are continuously jostled by the thermal motion of the rest of the molecule or nearby molecules. The excited protons give up their energy to the molecular lattice. One by one, the dipoles tip back parallel to the Z-axis, and M_z slowly reappears (Fig. 10.3c). This is also called longitudinal relaxation.

In Figure 10.5, the left-hand axis refers to curve A, which shows how, after M_z has been destroyed at time 0 by the first 90° pulse, it increases again relatively slowly and does so exponentially with a time constant T_1 that depends on the type of tissue.

T_1 is the time for M_z to recover to 63% of its maximum value. After three time constants, recovery is 95% complete:

Time	0	T_1	$2T_1$	$3T_1$
Recovery (%)	0	63	87	95

T_1 has a value of so many hundreds of milliseconds (Table 10.1). If a stronger magnetic field B is used, the protons precess faster and T_1 of tissue lengthens, and M_z reappears more slowly.

Table 10.1 Typical relaxation times of tissues in a field of 1 T

Material	T_1 (ms)	T_2 (ms)
Fat	250	80
Liver	400	40
Kidney	550	60
Spleen	400	60
White matter	650	90
Grey matter	800	100
Cerebrospinal fluid	2000	150
Water	3000	3000
Bone, teeth	Very long	Very short

Causes of spin–lattice relaxation

- Jostling by large molecules that are slow moving and near to the resonant frequency is most effective at removing energy from excited dipoles. Fat (large molecules with low inherent energy) can absorb energy easily and so has a relatively short T_1, as does water bound to the surface of proteins.
- Jostling by small, lightweight molecules with little inertia is rapid and so relatively ineffective at removing energy from the excited dipoles. Consequently, free water, urine, amniotic fluid, cerebrospinal fluid (CSF) and other solutions of salts have a long T_1. The greater the proportion of free water in a tissue, the longer is T_1.
- The atoms in solids and rigid macromolecules are relatively fixed, and they are the least effective at removing energy. Compact bone, teeth, calculi and metallic clips have a very long T_1.

Spin–spin relaxation or T_2 decay

Energy transfer between nuclei produces a loss of phase coherence, resulting in an exponential decay of the transverse magnetic vector, dependent on tissue type.

In a pulse sequence, M_z has not fully recovered when, TR seconds after the first pulse, a second 90° pulse converts the available M_z into M_{xy}. In Figure 10.5, the right-hand axis refers to curve B, which shows how M_{xy} then decays relatively rapidly, for the following reason.

Immediately after the 90° pulse, the dipoles are still all precessing in phase and their m_{xy} vectors simply add up. The large net magnetic vector M_{xy} induces a large MR signal. The dipoles then progressively dephase, as some rotate faster or slower than others. As a result, the net strength of the rotating magnetic vector M_{xy} decreases, and so does the induced signal (dotted curve C in Fig. 10.5).

Much of this dephasing effect is due to field inhomogeneities from machine factors external to the patient, but the clinically important part is related to tissue structure and is called spin–spin or transverse relaxation, or T_2 decay.

As depicted in Figure 10.5, M_{xy} decreases or decays exponentially (curve B), and so does the induced signal (curve C), both with a time constant T_2.

T_2 is the time for the MR signal to fall to 37% of its maximum value. T_2 has a value of so many tens of milliseconds (Table 10.1). After three time constants, only 5% of it remains:

Time	0	$1T_2$	$2T_2$	$3T_2$
Signal (%)	100	37	14	5

Causes of spin–spin relaxation The dephasing occurs because a spinning proton experiences a tiny additional magnetic field (around 1 μT) produced by each neighbouring proton. Individual protons are affected slightly differently, and the magnetic field B therefore varies a little from place to place and from time to time on the submicroscopic scale. So does the rate of precession; some precess faster and some slower, and energy passes from one proton to another, or spin to spin.

The local variation of magnetic field is greatest in solids and rigid macromolecules in which the atoms are relatively fixed. The dipoles in compact bone, tendons, teeth, calculi and metallic clips dephase quickly. They have a very short T_2 and do not produce a lasting signal.

The effect is least in free water, urine, amniotic fluid, CSF and other solutions of salts. The lighter molecules are in rapid thermal motion, which smoothes out the local field and results in a long T_2. Broadly speaking, the greater the proportion of free water in tissue, the longer is T_2. Spleen has a longer relaxation time than liver, and renal medulla longer than the cortex.

Water bound to the surface of proteins and other large molecules, which move more slowly, has a shorter T_2 than free water, and so does the hydrogen in fat.

Tissue characteristics

T_2 is always shorter than T_1. T_2 is more or less unaffected by, but T_1 of tissue increases with, magnetic field strength, i.e. with resonant frequency. There are no *precise* values of T_1 or T_2 for specific tissues. Figures cover a wide range, and only representative, rounded values are given in Table 10.1. Abnormal tissue tends to have a higher PD, T_1 and T_2 than normal tissue, because of increased water content or vascularity.

Because T_1 and T_2 are properties of the tissues that show a greater variation than PD itself, they are the properties actually used in forming the MR image. Comparing the range of T_1 and T_2 values for brain tissue in Table 10.1 with the limited range of computed tomography (CT) numbers (which depend on the very small differences in X-ray absorption) shows the superior soft tissue contrast resolution of MRI compared with CT.

10.3 SPIN–ECHO SEQUENCE

In practice, the MR or free induction decay (FID) signal is rarely measured because it decays so very rapidly – with a time constant T_2^* of a few milliseconds, much shorter than T_2. This happens because the static field $\boldsymbol{B}$ is not perfectly uniform:

- mainly because of the magnetic field gradient (see section 10.4) deliberately produced across the voxel
- because of unavoidable imperfections in the engineering of the magnet
- because the introduction of the patient unavoidably distorts the static field (due to magnetic susceptibility, see section 10.8).

These systemic effects unfortunately add to the effect of spin–spin interactions in the tissue in causing some dipoles to precess faster than others after a 90° pulse, with consequent dephasing.

To remove the effects associated with the static field but leaving the tissue characteristic T_2 effect, a *spin–echo (SE) pulse sequence* is used. Figure 10.6 depicts one cycle of this sequence, which is repeated hundreds of times in producing one MR image frame. It should be studied in conjunction with the sequence of events (a–d) depicted in Figure 10.7.

In the SE sequence, each 90° pulse is followed, t seconds later, by a 180° pulse. The signal is measured after a further and equal time interval (so that the echo time, TE, is $2t$).

- *Step a.* Immediately after the 90° pulse, the dipoles are all precessing exactly in phase (Fig. 10.7a). $\boldsymbol{M}_{xy}$ is a maximum, and so is the FID signal, but it is not measured at this stage, because it decays so rapidly.
- *Step b.* The $\boldsymbol{m}_{xy}$ vectors begin to dephase, the faster precessing ones (leaders) getting ahead of the slower ones (laggers in Fig. 10.7b). $\boldsymbol{M}_{xy}$ and the FID signal decay with time constant T_2^* (extreme left, Fig. 10.6).
- *Step c.* After time t, the 180° pulse is applied and tips all the dipoles from spin up to spin down. This

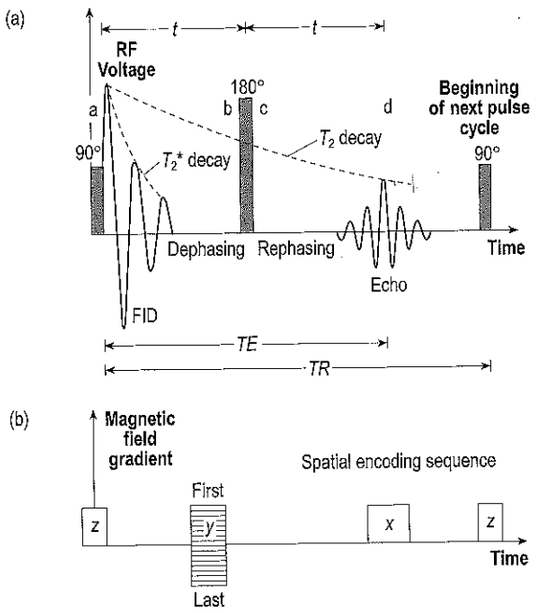

Figure 10.6 One cycle of the spin–echo sequence of (a) radiofrequency pulses and signal and (b) magnetic field gradient switching. FID, free induction decay; RF, radiofrequency; *TE*, echo time; *TR*, repetition time.

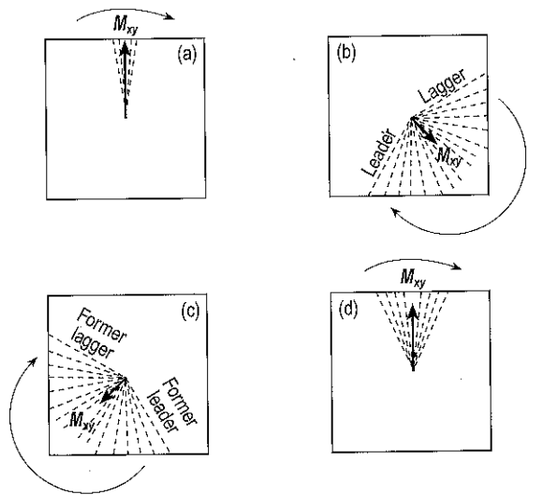

Figure 10.7 Dephasing and rephasing of the m_{xy} vectors of individual protons in a voxel in a spin–echo sequence: (a) dephasing starts immediately after a 90° pulse, (b) just before the 180° pulse; (c) just after the 180° pulse, rephasing begins; and (d) at time *TE* (echo time), only residual dephasing remains.

turns the individual $\boldsymbol{m}_{xy}$ vectors through 180° in the X-direction (Fig. 10.7c). Laggers become leaders, and vice versa. $\boldsymbol{M}_{xy}$ and the signal are still small. As the $\boldsymbol{m}_{xy}$ vectors continue to rotate in the XY plane, they now rephase. The faster ones catch up with the slower ones, and $\boldsymbol{M}_{xy}$ and the MR signal regrow.

- *Step d.* After a further time t, they are again momentarily in phase (Fig. 10.7d) and $\boldsymbol{M}_{xy}$ and the MR signal are at their peak. Thereafter, they grow out of phase again and $\boldsymbol{M}_{xy}$ and the MR signal decay.

The 180° pulse is often called the rephasing or refocusing pulse. It reverses and eliminates the dephasing effect of systemic magnetic field inhomogeneities. This leaves only the residual dephasing (Fig. 10.7d) due to the random effects of spin–spin interaction, T_2.

The MR signal reappears as an echo of the initial signal (see Fig. 10.6) and is essentially two FID signals back to back. When measured at time $TE = 2t$, it will have been reduced in amplitude by T_2 decay. The longer is TE, the smaller the MR signal.

Tissue contrast

The MR image maps three intrinsic properties (PD, T_1 and T_2) of tissue, and is controlled by two parameters set by the operator: TE (or time to echo) and TR (or time to repeat). These are selected to weight the contrast in the image.

The MR signal arising from a voxel and the brightness of the pixel depends on:

- How many protons there are in the voxel. The greater the PD, the larger the signal and the brighter the pixel.
- How far $\boldsymbol{M}_z$ has recovered from the previous 90° pulse when the next 90° pulse tips it, i.e. the length of T_1 compared with TR.

  Figure 10.8 compares two tissues that differ only in T_1, their PDs being the same. It shows that the shorter T_1 (or the longer TR), the greater the $\boldsymbol{M}_z$ available to be tipped and the larger the MR signal, the brighter the pixel, and the better the signal to noise ratio (SNR). It also shows that the TR can be selected to give a maximum contrast between particular tissues.
- How far $\boldsymbol{M}_{xy}$ has decayed when the echo is formed, i.e. the length of T_2 compared to TE.

  Figure 10.9 compares two tissues that differ only in T_2 (their T_1 and PD being the same). It shows that the longer T_2 or the shorter TE, the larger the MR signal, the brighter the pixel, and the better the SNR. It also shows that the TE can be selected to give a maximum contrast between particular tissues.

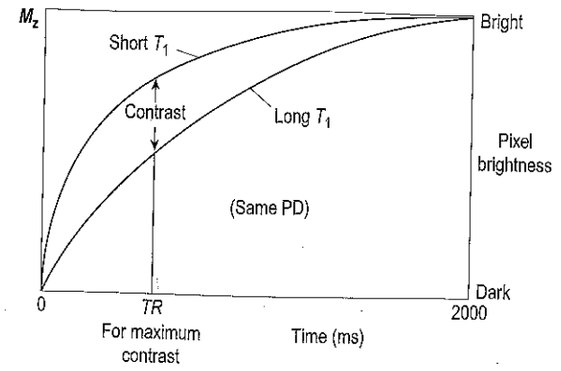

Figure 10.8 T_1 contrast.
PD, proton density; *TR*, repetition time.

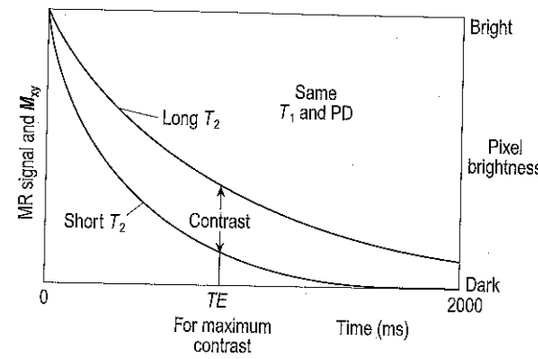

Figure 10.9 T_2 contrast.
MR, magnetic resonance; PD, proton density; *TE*, echo time.

Weighted images

The TE and TR are chosen so that pixel brightness depends on one of three combinations of PD, T_1 and T_2.

T_1-weighted image Figure 10.8 shows that maximum contrast (the difference between the curves) between tissues of different T_1 is produced by a fairly short TR.

A TR of 300–800 ms is used – about the same as the average T_1 of the tissues of interest. A short TE (15 ms) is also used, as this reduces the effect of T_2 on contrast. (It cannot be much shorter, as the system has to apply the 180° RF pulse and a series of gradient pulses before the MR signal can be measured.)

Image contrast is then principally due to the T_1 recovery properties of the tissues. The shorter is T_1, the stronger the signal and the brighter the pixel. *Fat is bright*, as is fatty bone marrow, while *water and CSF are dark*.

T_2-weighted image Figure 10.9 shows that the longer is TE, the greater the contrast between tissues of

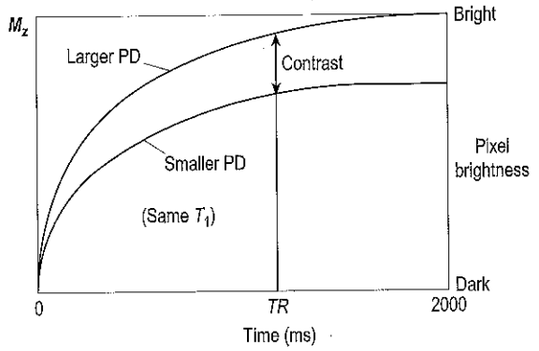

Figure 10.10 Proton density (PD) contrast. *TR*, repetition time.

different T_2. However, it must not be so long that the signal is so small as to be obscured by background noise.

A relatively long *TE* of 90–140 ms is used – about the same as the average T_2 of the tissues of interest. A long *TR* (1000–2000 ms) is also used, as this reduces the effect of T_1 on contrast, although unfortunately it increases the imaging time.

Image contrast is then principally caused by differences in the T_2 decay properties of the tissues. The longer is T_2, the stronger the signal and the brighter the pixel. *Water and CSF appear brighter than fat.*

Proton density–weighted image Figure 10.10 compares two tissues that differ only in PD, their T_1 being the same. It shows that the longer is *TR*, the greater the contrast between tissues of different PDs.

A long *TR* (1000–3000 ms) is used – about $3T_1$ – and this reduces the effect of T_1 on contrast. A short as possible *TE* (15 ms) is used, as this reduces the effect of T_2. Generally speaking, PD weighting produces greater signal strength and less noise.

Image contrast is then principally caused by differences in the proton densities of the tissues. The greater the PD, the stronger the signal and the brighter the image. CSF, fat and indeed most tissues, having a high PD (number of hydrogen protons), appear bright.

Summary A large signal and a bright pixel result from tissues having a large PD and a long T_2. A small signal and a dark pixel result from a long T_1 and (as will be seen later) arterial blood flow. *In all images*, air and cortical bone, having no hydrogen, appear black.

The *TR* controls the amount of T_1 weighting, and *TE* the amount of T_2 weighting in the image, as in Table 10.2.

In practice, matters may not be so clear-cut. It is not possible to rid any image of some T_2 weighting, and relative weighting may differ from tissue to tissue across an image.

Table 10.2 T_1-, T_2- and proton density–weighting effects

Weighting	Properties	Result
T_1-weighted	*TR* and *TE* both short	Short T_1 = bright
T_2-weighted	*TE* and *TR* both long	Long T_2 = bright
PD-weighted	Short *TE* and long *TR*	High PD = bright

PD, proton density; *TE*, echo time; *TR*, repetition time.

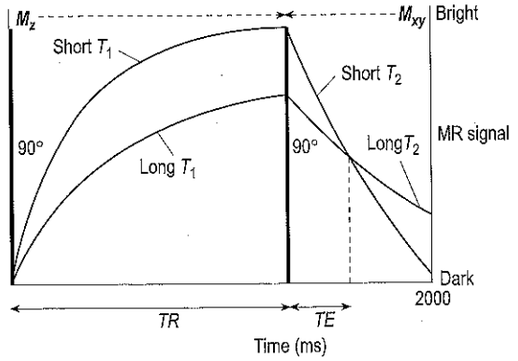

Figure 10.11 Contrast: combined effects of T_1 and T_2. MR, magnetic resonance; *TE*, echo time; *TR*, repetition time.

White and grey matter In a PD-weighted image, grey matter, with its somewhat higher PD, appears brighter than white matter. In a T_2-weighted image, grey matter, with its longer T_2 and higher PD, is brighter than white matter. In a T_1-weighted image, white matter is brighter than grey matter but its shorter T_1 is somewhat counteracted by its lower PD.

The opposing effects of T_1 and T_2

T_1 and T_2 are mutually antagonistic and have opposing effects on image brightness. Generally speaking, tissues with a long T_1 also have a long T_2, and those with a short T_1 have a short T_2. This is why images cannot be weighted for both T_1 and T_2.

Remembering once again that *the signal produced by a 90° pulse depends on the value of M_z immediately before the pulse is applied*, Figures 10.8 and 10.9 can be combined into Figure 10.11. This shows that, with an injudicious choice of *TE* and *TR*, tissues with quite different relaxation times can produce equal signals. Showing no contrast, they will be indistinguishable. With a shorter *TE* than this, the image tends to be T_1-weighted, and with a longer *TE* it tends to be T_2-weighted.

Table 10.3 Factors affecting magnetic resonance signal and contrast

Fixed parameter	Machine settings	Tissue characteristics
Gyromagnetic ratio of nucleus	*TR*	T_1
	TE	T_2
	–	Proton density
Static magnetic field	*TI* (see section 10.5)	Flow (see section 10.6)
	Tip angle (see section 10.5)	Contrast medium (see section 10.6)

TE, echo time; *TR*, repetition time.

Other factors affecting the magnetic resonance signal and contrast

Table 10.3 shows that there is a more complex state of affairs than in X-ray imaging.

10.4 SPATIAL ENCODING

To produce an image, it is necessary to collect and analyse the signals coming from the patient in terms of their amplitude, frequency and phase, and then to Fourier transform them to produce the individual pixels or voxels in the image. Magnetic field gradients are used to localize the MR signal, which itself is encoded in terms of spatial frequencies using phase-encoding and frequency-encoding gradients. The three basic processes involved are slice selection, phase encoding and frequency encoding. Note that every spatial frequency must be collected and stored before the data can be Fourier transformed to produce the image.

Slice selection

As in CT, an MR image is made up of a series of parallel slices that are imaged in turn. Figure 10.12 represents a sagittal section through the patient with a transverse slice (shaded).

Simultaneously with the 90° RF pulse, DC is sent for a short while through a pair of gradient coils, which are additional to the main RF coil. This current produces a controlled magnetic field gradient, along the *Z*-axis, within the static magnetic field $\boldsymbol{B}_0$. The total $\boldsymbol{B}$ is diminished at (say) the head end and augmented at the toe end (Fig. 10.12b) while remaining the same at the isocentre. In this case, it varies from head to toe with a constant gradient of a few millitesla per metre (mT m^{-1}).

Accordingly, protons at the head end precess more slowly than those in the middle, and those at the foot do so faster. There is therefore a corresponding gradient, in the *Z*-direction, of the resonant frequency of the protons, and so the frequency can be used to localize the signal. The protons in a selected slice are all precessing with a narrow range of frequencies (Fig. 10.12c).

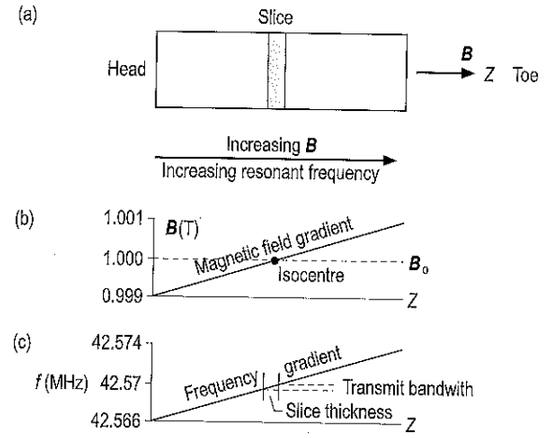

Figure 10.12 Transverse slice selection with a *Z*-field gradient: (a) sagittal cross-section of the patient, (b) magnetic field gradient, and (c) transmit frequency gradient.

The RF transmitter is tuned to generate an RF pulse that contains a small range of frequencies (a narrow bandwidth). Only protons in a certain thin slice of the patient will be excited by it. The magnetic vectors of only those protons will tip and, in due course, produce an MR signal. Note that the slice select gradient field is switched on during the application of the RF pulse. Different slices are selected in turn by simply altering the central frequency of the RF pulse, without having to move the patient.

Slice thickness

The slice thickness may be reduced by either increasing the gradient of the magnetic field or decreasing the RF (or transmit) bandwidth. A thinner slice produces better anatomical detail, the partial volume effect being less, but it takes longer to excite.

A typical slice thickness is 2–10 mm. The RF pulse inevitably contains a certain amount of electromagnetic energy of frequencies slightly higher or lower than the intended bandwidth, thus mildly exciting tissues either side of the desired slice. To prevent this 'cross-talk' affecting the image slice, a gap (say 10% of the slice thickness) may be left between slices, although this is not necessary when the slices are interleaved (see section 10.5).

The slice orientation depends on the physical gradient axis. As described here, using the *Z*-axis gives a transverse slice. However, using the *X*-axis would give a sagittal slice, and the *Y*-axis a coronal slice. Any orientation is actually possible by combining the gradients.

In-slice localization

Having selected the slice, the objects within the slice have to be localized. This is achieved by applying *phase encoding* and *frequency encoding* each orthogonal to the slice, the MR signal being measured during the latter part of the frequency encoding gradient.

The complex MR signal is sampled and computer analysed into a spectrum of component frequencies using a mathematical process called Fourier analysis. The data from every signal in a selected slice are stored in what is known as K space. This space is simply a spatial frequency domain in the computer where the signal spatial frequencies and their origin are stored. The spatial frequencies correspond to the variations in image brightness as the encoding gradients are applied. K space does not correspond to the image but rather has axes corresponding to frequency and phase. The number of lines filled in K space matches the number of encodings in the sequence (e.g. 128, 256 or 512). The central part of K space contains data from shallow encoding gradients, low spatial frequencies and hence less details but stronger signals. The upper and lower parts are filled with data from the steeper gradients, high spatial frequency so better detail but low signal intensity, as indicated in Figure 10.13. Note that K space has to be completely filled with the data from the imaging sequence before the signals can be analysed and processed into the image.

Phase encoding

Immediately after the protons in the slice have been excited by the 90° pulse, but before they are inverted by the 180° pulse, DC is passed for a few milliseconds through a second set of gradient coils. This produces a magnetic field gradient, for example in the *Y*-direction (Fig. 10.14a,b), from the front to the back of the patient.

For that brief period of time, some of the precessing dipoles and M_{xy} vectors speed up and some slow down. Those in voxels near the top of the column (say) precess more slowly and lag behind those in the middle, while those near the bottom precess faster and get ahead.

When the gradient pulse is over, they all precess again at the same rate, and they again all emit the same frequency signal. However, the *phase differences* remain, and these are dependent on the position. Those near the top are still behind those near the bottom. There is a phase gradient in the MR signals coming from different pixels for the same type of tissue along the selected vertical line (Fig. 10.14c); the nuclei are phase-encoded. With a steep gradient, the spins will be evenly distributed in every direction and the total signal will be zero. With no gradient, the spins are all in phase and the signal will be a maximum. This is referred to as zero spatial frequency and the data are located on the central line of K space.

In order to map, for example, a 512 × 512 matrix, there must be this number of possible spatial frequencies. The phase-encoding pulse must be repeated with the gradient increased a little after each excitation, thus stepping up the phase shifts. By comparing the pattern of increasing phase angles *f*, it is possible to decipher the separate signals across the field of view (FOV). The FOV increases if the phase-encoding step size is made smaller. The pixel size equals the FOV divided by the number of phase-encoding gradient steps used.

Frequency encoding

At the same time as the gradients for phase encoding are applied, DC is passed through the third set of

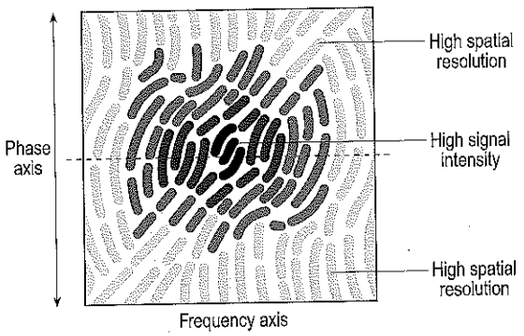

Figure 10.13 K space: each data point in the matrix is a spatial frequency component of the signal.

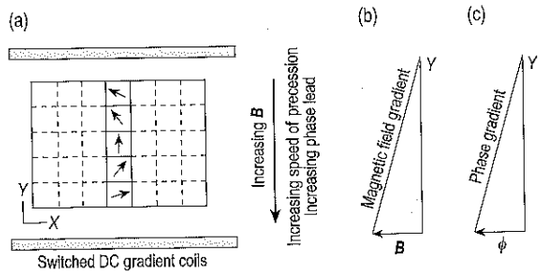

Figure 10.14 Phase encoding using a *Y*-field gradient: (a) transverse cross-section of the patient, (b) magnetic field gradient, and (c) phase gradient for a given tissue type. DC, direct current.

gradient coils to produce a magnetic field gradient, also orthogonal to the slice selection gradient, for example from side to side in the *X*-direction (Fig. 10.15).

Protons in each vertical column of Figure 10.15 experience the same magnetic field, precess with the same frequency, and emit MR signals of the same frequency. But those in the left-hand columns (say) precess more slowly than those in the middle, and those on the right precess faster. There is a corresponding frequency gradient from left to right in the MR signals emitted (Fig. 10.15c). This is a bit like the frequency gradient along the keyboard of a piano: the pitch of the note reveals (or 'codes for') the position of the key that was struck. The MR signal produced by exciting the slice therefore consists of a range of RF frequencies either side of the frequency of the applied pulse.

Field of view The receiver is tuned to accept only a certain range of frequencies, called the receive bandwidth, coming from a corresponding FOV (Fig. 10.15c). The FOV may be increased by either making the field gradient less steep or increasing the receive bandwidth. The voxel width equals the FOV divided by the number of components into which the frequency spectrum has been sampled.

The MR signal emitted by the whole slice therefore comprises a mixture or spectrum of phases as well as of frequencies. At the same time as the computer is analysing the signal for frequency, it is analysing it for phase. This Fourier analysis is analogous to the way the human ear picks out individual instruments from an orchestra. However, because phase angles repeat themselves every 360°, it is impossible to be confident about assigning a particular phase to just one point. Typically, there are several phase cycles across the whole FOV, so the phase angle of, say, 10° is repeated at 370°, 730°, and so on.

Measuring a series of phase angles over 512 repetitions is really the same as measuring a single signal at 512 points all in one go. Figure 10.16 shows how the two methods can be seen as equivalent. By using a Fourier transform along the phase-encoding direction, it is possible to measure the frequencies in this direction, which correspond to position along the phase-encoding axis. The *TR* is sometimes described as pseudotime, and by analogy the FT of the phase-encoded signals would be called a pseudofrequency. In this sense, it is possible to think of phase encoding as simply frequency encoding over a very long time scale.

Imaging time The time needed to acquire an image is obtained by multiplying the number of signal averages or excitations N_{ex}, the number of phase-encoding steps, and the pulse repetition time *TR*, and may total several minutes. Increasing the number of excitations

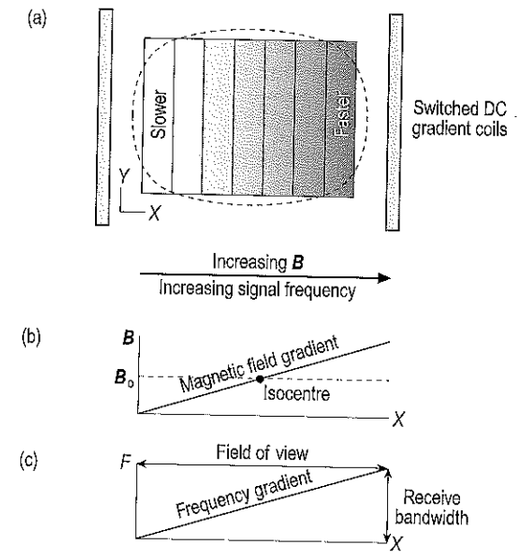

Figure 10.15 Frequency encoding using an *X*-field gradient: (a) transverse section of the patient, (b) magnetic field gradient, and (c) receive frequency gradient.

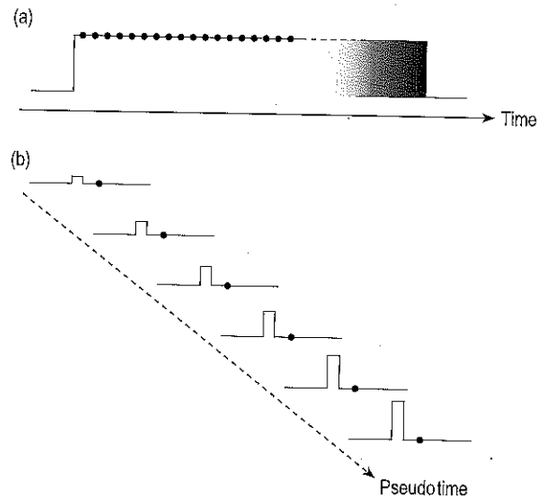

Figure 10.16 Equivalence of frequency and phase encoding, using the concept of pseudotime. Each dot represents an acquired data point. (a) With frequency encoding, all the data points are acquired at microseconds intervals – gradient on all the time; (b) with phase encoding, each data point follows the application of a separate, increasing sized gradient in pseudotime steps equal to repetition time (*TR*) ms. (Courtesy of E.A. Moore.)

reduces noise at the expense of increased imaging time. For example, four repetitions ('excitations') of the foregoing SE sequence that are averaged will improve the SNR by a factor $\sqrt{4} = 2$.

Once all the signals are collected, a fast Fourier transform converts the K-space distribution into an image of the patient. This actually produces a complex image having real and imaginary parts, which are usually combined as a complex magnitude image. Note that a phase image can also be calculated, if required.

If any tissues move during the repetitions, they will be misregistered, i.e. signals will be attributed to the wrong pixels in the phase-encoding direction of K space. Such motion artefacts, described in section 10.8, when seen on an MR image make it clear which is the direction of phase encoding *vis-à-vis* frequency encoding. (Similarly, any chemical shift artefacts, described in section 10.8, reveal the direction of frequency encoding.)

Summary of gradients

The gradient fields are superimposed in pulses on to the static field. One gradient defines the slice and is applied when the RF pulse is turned on. A second orthogonal gradient is used for phase encoding and is applied briefly between the 90° and 180° RF pulses. The third orthogonal gradient is used for frequency encoding, during which the signal is measured. By combining frequency encoding and phase encoding, all the spatial frequencies within a slice can be collected unambiguously as needed to produce the two-dimensional image. Figure 10.6 shows how the field *gradients* are applied during the SE sequence. In practice, the gradient sequences will usually be more complex than described here.

The gradients are used to control the slice thickness and the FOV. The steeper is the slice selection gradient, the thinner the slice. The steeper are the frequency and phase-encoding gradients, the smaller the FOV. The steeper a gradient, the greater the power consumed by the gradient coils.

There is a significant difference between phase and frequency encoding. All the spatial information required in the frequency-encoding direction is obtained in 10–20 ms by sampling a single echo, and increasing the matrix makes no difference to the total scan time (although it may reduce the number of slices possible within the *TR*). In the phase-encoding direction, the spatial information is not complete until all the gradient steps (e.g. 512) have been completed. This can lead to the appearance of motion artefacts (see section 10.8).

10.5 OTHER PULSE SEQUENCE AND IMAGING TECHNIQUES

So far, we have considered imaging in the transverse plane using a single SE sequence. The following gradient sequences are those most often encountered, although they may have different acronyms depending on the MR scanner manufacturer.

Multislice techniques

Most of *TR* is 'wasted' if the scanner has to wait up to 2 s before repeating pulses on a given slice. This time can be used to deliver a succession of 90° and 180° RF pulses, each of a different frequency, exciting a series of up to 32 separate slices before repeating the first slice (Fig. 10.17). The shorter that *TE* is compared with *TR*, the more slices can be interleaved in this way. If *TR* = 1000 ms and *TE* = 60 ms, in principle 1000/60 = 16 slices, but in practice about 13 slices can be excited.

By acquiring the slices out of sequence, the need for gaps due to cross-talk when slices are acquired sequentially is avoided.

Multiecho techniques

This is another way of making use of the long *TR*. Following each 90° pulse, two or more successive 180° pulses produce successive echoes with increasing *TE*. Their peak amplitudes decrease with the time constant T_2. The first echo may produce a PD-weighted image, while successive echoes produce images that are increasingly T_2-weighted (and increasingly noisy). A dual echo sequence can be written as 90, 180, 180 and produces two images per slice. The first echo has a short *TE* (20 ms) and long *TR* (2000 ms), which produces a PD-weighted image. The second echo has a longer *TE* (80 ms) and produces a T_2-weighted image. There may also be time during a single *TR* to interleave a number of slices.

Fast (or turbo) spin–echo

The 90°, 180°, 180°, ... sequence can be modified by phase encoding each of the 4–16 echoes (echo spacing 20 ms apart, say) with a different phase-encoding gradient. This reduces by a factor of 4–16 the time

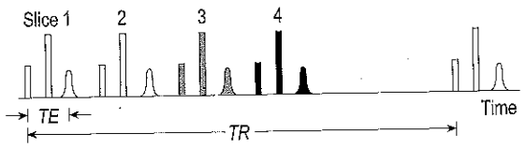

Figure 10.17 Multislice imaging.
TE, echo time; *TR*, repetition time.

(i.e. the number of *TR* intervals) needed to acquire a complete image. However, fewer slices can be interleaved. Fat produces an extremely high (bright) signal on fast SE (FSE or TSE) T_2-weighted images, because the rapid succession of 180° pulses reduces the spin–spin interactions, thus increasing T_2. Fat suppression techniques are sometimes needed. Conversely, muscle is often darker than in single SE T_2-weighted images, as the succession of pulses increases the transfer of magnetization and results in saturation.

As scan time is reduced, matrix size can be increased to improve spatial resolution (smaller voxels). Note that FSE is incompatible with respiratory compensation techniques, but the use of powerful gradients can enable an image to be obtained in a single breath hold.

Inversion recovery

To accentuate T_1 weighting, an initial 180° pulse is used (Fig. 10.18a). This tips the spins antiparallel to the *Z*-axis and inverts M_z. The spins progressively return to parallel, due to spin–lattice relaxation. M_z recovers, passing through zero and reversing direction after a time of $0.69 \times T_1$.

After a variable time (*TI* is the time to inversion, e.g. 500 ms), a 90° pulse is applied that tilts the available M_z. The M_{xy} vector so produced rotates in the transverse *XY* plane (Fig. 10.3), producing an MR signal (FID). A second 180° pulse is then used to develop an echo signal. The whole (180°, 90°, 180°) cycle is repeated after *TR* (Fig. 10.18a). Typical parameters might be *TR* = 1000 ms and *TE* = 20 ms.

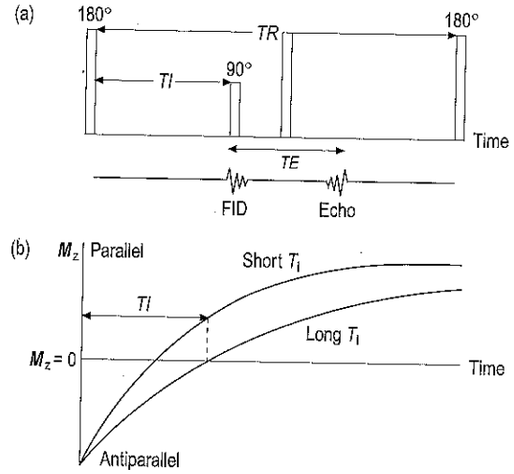

Figure 10.18 (a) Inversion recovery sequence. (b) Recovery of tissue having different T_1 values.
FID, free induction decay; *TE*, echo time; *TI*, time to inversion; *TR*, repetition time.

Consider two tissues of different T_1. If $TI = 0.69 \times$ the longer T_1, that tissue gives no signal, as there is no available M_z to convert, whereas the tissue with the shorter T_1 does. The image is T_1-weighted, and tissues with the longer T_1 are suppressed. The longer *TI* is or the shorter T_1 is, the greater the MR signal produced. *TE* controls the T_2 decay and must be short for T_1 weighting.

The *TI* is used as a T_1 contrast control. *TR* is about $3T_1$ to ensure nearly total recovery between pulses. This technique is time-consuming, especially at higher field strengths, which makes T_1 and so *TI* long but gives good grey–white matter discrimination.

Short-TI inversion recovery sequence: for fat suppression

In SE sequences, the very bright signal produced by fat may obscure contrasts in other tissues. There are several ways of dealing with this. One is to remove the signal from fat by using an inversion recovery sequence, with its initial 180° pulse, followed rapidly by a 90° pulse.

The 180° pulse tips both fat and water protons antiparallel, but they recover more quickly in fat, with its short T_1, than in water. This is shown in Figure 10.19, which may be compared with Figure 10.18. After a certain time *TI* (about 125 ms), half the spins in fat have reverted to parallel, and its $M_z = 0$. Few of the spins in water have so reverted, and it still has some M_z. A 90° pulse at this instant produces a signal from water and other tissues but none from fat.

Tip angle

There are several ways of reducing scan times. One method is to reduce *TR* to 200 ms or even 20 ms. This would give M_z little time to recover and result in rather small signals from the usual 90° RF pulse. Instead, an RF pulse of shorter duration (smaller tip angle or flip angle) is used, which inverts only a small fraction of the dipoles.

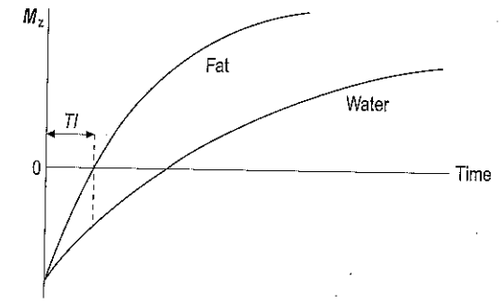

Figure 10.19 Short-*TI* inversion recovery.

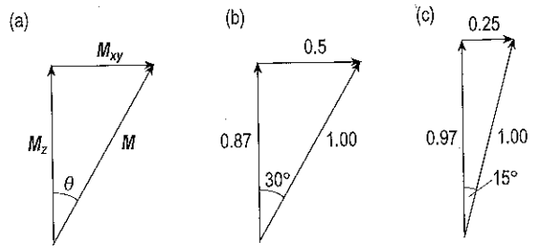

Figure 10.20 (a) Meaning of tip angle. Relative values of M_z and M_{xy} at tip angles (b) 30° and (c) 15°.

Table 10.4 Effect of tip angle

Weighting	Properties	Result
T_1-weighted	TR and TE both short	Tip angle >70°
T_2-weighted	TR and TE both long	Tip angle 5–10°
PD-weighted	Short TE and long TR	Tip angle 5–10°

PD, proton density; *TE*, echo time; *TR*, repetition time.

Figure 10.20a (which may be compared with Fig. 10.4) shows how M_z, M_{xy} and the tip angle θ are related through a right-angled triangle. Figure 10.20b shows that a 30° pulse produces only half of the usual M_{xy} and the consequent MR signal. However, it leaves 87% of M_z, which does not take long to recover fully. This ensures a good signal following the next RF pulse. Similarly (Fig. 10.20c), a 15° pulse produces 25% of the usual M_{xy} but leaves 97% of M_z.

The stronger the initial RF pulse and the longer its duration, the greater the tip angle. The greater the tip angle, the greater the T_1 weighting (Table 10.4).

The *optimum tip angle*, which gives the greatest signal, is a balance between leaving sufficient M_z and producing sufficient M_{xy}. The shorter TR is, compared with T_1, the smaller is the optimum tip angle:

TR/T_1	3	1	0.14	0.03
θ (°)	87	68	30	15

Gradient (recalled) echo (GRE)

With *TE* typically 15 ms, there is time within the short *TR* for very few slices to be imaged. If *TR* is as short as 20 ms, the image must be acquired one slice at a time. Some 15 separate slices can be taken in 30 s, a sufficiently short time for patients to hold their breath but not rapid enough to produce real-time images (see *Echo planar imaging* and *Parallel imaging*).

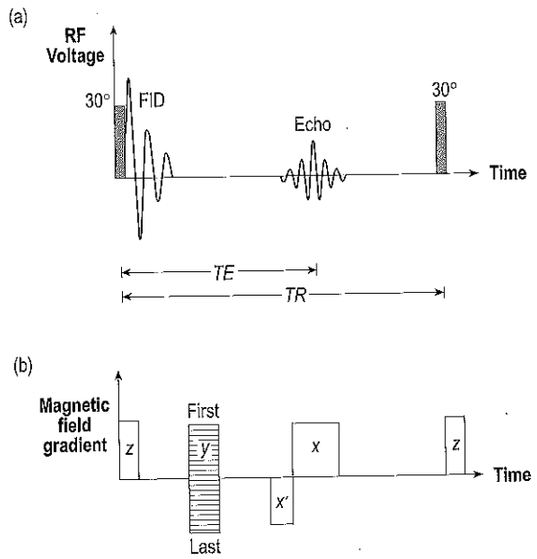

Figure 10.21 Gradient echo. (a) Radiofrequency (RF) pulses and signal and (b) magnetic field gradients.
FID, free induction decay; *TE*, echo time; *TR*, repetition time.

With such short *TE*, there is no time for a 180° pulse as used in SE sequences. Instead, rephasing is achieved using *a gradient echo* (Fig. 10.21). The gradient field is reversed to refocus the out-of-phase spins. This compensates for the dephasing produced by the change of magnetic field across the voxel produced by the gradient field (see section 10.3). Unlike SE, however, it does not eliminate the effect of inhomogeneities in the static field, and so the image is T_2*-weighted; nor does it compensate for magnetic susceptibility effects (see section 10.8).

The usual frequency-encoding gradient pulse X is preceded by a reverse pulse X′ (of half the duration). During the first, negative, part of the gradient pulse, the dipoles at one side of each voxel are made to precess faster than and get ahead of those at the other. Then the gradient current reverses and the latter begin to catch up with the former, coming into phase again to produce the echo signal. The peak of the MR signal appears at the middle of the *X*-gradient pulse (Fig. 10.21b).

To summarize: compared with the SE sequence, in a GRE sequence the RF pulse is of reduced strength and tips the magnetic vector through a smaller angle than 90°, thus allowing a short *TR*. The negative gradient pulse X′ dephases the spins, and the positive gradient pulse X, which is twice as long, rephases them.

An important characteristic of GRE scans is that *moving blood appears bright* (see *Magnetic resonance angiography*).

Echo planar imaging

A 50-ms 'snapshot' may be produced by an extremely fast form of GRE called echo planar imaging (EPI). Following a standard (90°, 180°) SE sequence and slice selection gradient, the polarity of a frequency-encoding gradient is continually reversed, as fast as possible, each time inducing a gradient echo. The phase-encoding gradient is also switched on and off, briefly, just before each echo, thus encoding each of the echoes with a different phase-encoding gradient. Multiple echoes can be collected before M_{xy} has decayed too far, to give a complete image in the 50 ms following the SE 90–180° pulses. PD or T_2 weighting is obtained by using short or long effective *TE*. T_1 weighting is possible if an inverting pulse is applied before the excitation pulse to produce saturation. The whole brain can be imaged in about 2–3 s. However, resolution, echo strength and the signal to noise are all compromised.

The strong field of a superconducting magnet is necessary, and artefacts may be a problem. As very high gradients and very fast switching are needed, this will induce small, unwanted currents in nearby metallic structures, and these may cause blurring and artefacts in the image (see section 10.8). They can be reduced by active shielding of the gradient coils (see section 10.10).

This ultrafast technique can be used for functional imaging, real-time cardiac imaging and perfusion or diffusion imaging.

Imaging in other planes

Any desired image plane can be selected without moving the patient.

To image in the coronal plane, the *Y*-gradient is used for slice selecting and the other two for frequency and phase encoding. To image in the sagittal plane, the *X*-gradient is used for slice selecting and the other two for frequency and phase encoding. To image in the coronal oblique plane, slice selection is applied to the *X*- and *Y*-gradients simultaneously.

Generally speaking, the phase-encoding gradient is best applied along the shorter dimension of the patient's anatomy.

Three-dimensional Fourier imaging (volume imaging)

This is an alternative method of spatial encoding. A shallow Z-gradient is used to select a thick slice, thick enough to include the whole volume to be imaged. As usual, frequency encoding is used along one axis, but phase encoding is used along both of the others. The data-processing requirements are increased accordingly; so is the scan time, but this may be mitigated by using GRE with a short *TE*. A three-dimensional Fourier transform is used to decode the information. Because the slice selection direction is phase encoded, it is also subject to motion artefacts and to phase wraparound artefacts, just like the normal in-plane phase encoding. The three-dimensional data can be 'reformatted' to produce images of a series of very thin contiguous slices (no gaps) in any orientation, with no cross-talk.

Parallel imaging

Most MR techniques need high temporal and spatial resolution. Before parallel imaging techniques were available, multiple slice techniques and increasing the gradient strength were the only way to do this, although the latter is expensive and has unwanted side effects (peripheral nerve stimulation).

Simultaneous acquisition of spatial harmonics (*SMASH*) and *sensitivity encoding* (*SENSE*) each use an array of RF detection coils to perform some of the phase encoding usually done by magnetic field gradients. Conventional spatial encoding with gradients is a serial process, whereas SMASH and SENSE are partially parallel imaging techniques. Each coil in an array is connected to a separate RF receiver, resulting in parallel streams of data, each of which produces a separate image. SMASH and SENSE use the information from both the localized sensitivities of the individual coils and their independent signals. Thus images are obtained with fewer phase encode steps, the limiting factor in MRI. Most existing fast-imaging sequences such as EPI can be used with SMASH and SENSE in half the normal image acquisition time.

If each RF coil *transmits* as well as receives the signal, better separation of the signals is obtained. The pulse duration is reduced, enabling short *TE*–short *TR* sequences, optimizing the SNR with reduced scan time in transmit SENSE.

This field is developing constantly and is enabling techniques such as radial or spiral scanning to become available.

10.6 SPECIALIZED IMAGING TECHNIQUES

Magnetic resonance angiography

The effect on the MR image of the flow of blood depends on many factors, including its velocity, flow profile, and direction relative to the slice (it is greatest for flow perpendicular to the slice), as well as the

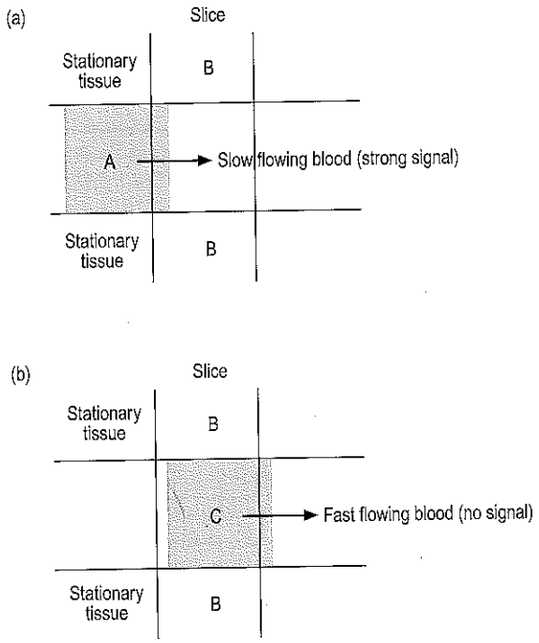

Figure 10.22 Appearance of blood during spin–echo: (a) slow-flowing and (b) fast-flowing.

pulse sequence and its parameters. In multislice imaging, the appearance depends on whether the slices are acquired in the same or opposite direction to the flow. A few simple principles can be identified regarding the SE sequence.

- *Vessels containing slow-flowing blood (e.g. in a vein) may appear bright (flow enhancement).* Previously unexcited blood (A in Fig. 10.22a) that enters the slice during a 90° pulse is more affected by the pulse and produces a stronger echo than stationary tissue (B) in which M_z has not yet fully recovered from the previous 90° pulse. The effect is most noticeable when the pulses are repeated rapidly, with a short TR.
- *Vessels containing fast-flowing blood (e.g. in the aorta) appear dark or void.* Some of the blood (C in Fig. 10.22b) excited by the 90° pulse has already left the slice before the 180° pulse occurs, and so produces no echo signal. Conventional T_1 weighting with presaturation pulses produces 'black blood' flow, and so any signals from a vessel indicate stagnant flow or an occlusion.
- *Turbulent flow produces a rapid loss of coherence, thus reducing M_{xy}, and usually appears dark* – for example, turbulence downstream of a stricture in a vessel.

In a GRE scan, *flowing blood and CSF usually appear bright*, for the following reason. The RF pulses, which are rapidly repeated, excite spins only in the selected slice, but the gradient pulse (unlike the 180° pulse in SE, Fig. 10.22) rephases all spins whether in the slice or outside it. Even if (as in the second principle above) some of the excited blood has left the slice, the gradient pulse will rephase it, and it will still produce a signal. Flowing blood does not therefore appear black. On the contrary (as in the first principle above), the in-flowing blood will give a larger signal and appear brighter than stationary tissues, which have previously been repeatedly excited as long as *TR* is well below the T_1 recovery of the stationary tissues.

To image only the blood vessels in GRE, moving blood is recognized either by its increased brightness in 'time of flight angiography', or by the phase change caused by movement along the magnetic field gradient ('phase contrast angiography'). Stationary tissue shows no net phase change. No contrast medium is required because of the large difference in the MR signals from flowing blood and tissue. In two-dimensional angiography, a series of images, stacked in the direction of the vessel, is produced, analysed for maximum intensity related to each voxel; this is projected on to a single image plane to produce the angiogram. The same construction technique is used for CT angiography (Ch. 7.4.2). For good three-dimensional (volume) angiography with high signal to noise, high-velocity flow is needed (e.g. intracranial flow).

Perfusion imaging

Perfusion imaging uses a paramagnetic contrast agent to measure the rate at which blood is delivered to the capillary bed of tissues and thus metabolic activity. There are two main methods – either using a bolus of contrast agent or arterial spin labelling – usually needing EPI techniques.

Gadolinium (as the ion Gd^{3+}) is a highly suitable material to use as, having seven unpaired electrons, it is strongly paramagnetic. Being very toxic, it is chelated with diethylenetriaminepenta-acetic acid (DTPA), which is water-soluble and to which it remains bound until excreted. However, it is contraindicated in patients with renal dysfunction.

Used as an intravenous contrast medium, Gd-DTPA is not itself visible on the MR image. Tumbling at around the Larmor frequency, the paramagnetic molecules shorten both T_1 and T_2 of the hydrogen nuclei in their vicinity as a magnetic susceptibility effect. As the effect on T_1 is greater than on T_2, it is called a positive contrast agent. The area of uptake is made brighter in a T_1-weighted image. T_1 weighting is therefore generally used after gadolinium injection.

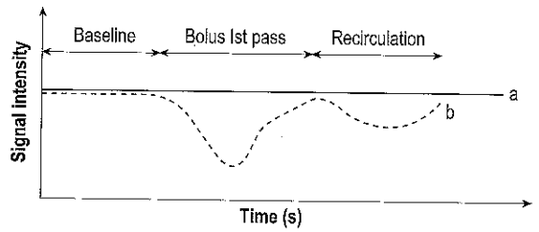

Figure 10.23 Signal intensity during bolus transit through different tissues: a, no change in signal, no perfusion; b, strong change, normal perfusion.

However, during the first pass of the agent, the gadolinium remains in the blood vessels and causes a large susceptibility effect in perfused tissues. T_2* weighting with a gradient echo is sensitive to these changes. Because the transit of the bolus through the tissue lasts only a few seconds, fast imaging techniques must be used. Analysis of the signal decay gives information on blood volume and a measurement of perfusion in terms of millilitres of blood per 100 g tissue per minute, the tracer concentration being roughly proportional to the relaxation rate in normally perfused tissues (Fig. 10.23).

As Gd-DTPA is water-soluble, it may produce increased contrast between pathological and normal tissue. It does not cross the normal blood–brain barrier and so is used to reveal breakdown of the barrier. The protons in water are more affected than those in fat. Water may appear equally bright as fat, and fat suppression techniques may have to be used.

The greater the concentration of Gd-DTPA, the shorter the relaxation times. Because T_1 and T_2 shortening have opposite effects, the concentration of Gd-DTPA must not be so great that the T_2 effect cancels out the T_1 effect.

In *arterial spin labelling*, two images are generally obtained, one in which the blood water magnetization is different, obtained by spin inversion techniques (*spin-labelled* image), from that of tissue water, and the *control* image, in which they are in the same state (no inversion is applied). Subtracting the two images gives the perfusion image from which the blood flow can be measured.

Applying the off-resonance RF pulse for inversion can cause magnetization transfer between free and bound water, resulting in a decrease of the expected contrast. To mitigate the effect in the subtracted image, an off-resonance RF pulse is usually applied before acquisition of the control image so that the magnetization transfer effects cancel.

Diffusion imaging

Diffusion-weighted imaging requires combining EPI or fast GRE sequences with two large gradient pulses applied after each RF excitation. The gradient pulses effectively cancel if spins do not move (are static), while moving spins undergo a phase shift. Tissue water with normal random thermal motion (diffusion) attenuates the signal due to the destructive interference of all the phase-dispersed spins, while a high signal appears from tissues with restricted movement or diffusion. Subtle changes in the restriction of movement are reflected in the signal attenuation that is directly related to the apparent (or effective) diffusion coefficient multiplied by a sensitivity weighting factor that depends on the time and the amplitude of the diffusion gradient.

Diffusion gradients must be strong and are applied in *X*-, *Y*- and *Z*-directions to obtain the diffusion-weighted image. Although the images themselves provide visual information on, for example, tissues damaged because of oedema, much more can be obtained only by post-processing of the data to provide parameters that enable quantitative assessment of tissue integrity and connectivity, for example.

Functional imaging (functional MRI)

Functional MRI techniques acquire images of the brain during an activity or stimulus and compares them with at-rest images. The physiological effect that produces the greatest change in MR signal between stimulus and rest is the blood oxygenation level. Haemoglobin contains iron and transports oxygen bound to the iron in the vascular system and so into the tissues. Oxyhaemoglobin is diamagnetic (magnetic properties are weakly opposed to the main field), while deoxyhaemoglobin is paramagnetic and produces magnetic field inhomogeneities in neighbouring tissues, increasing T_2*. At rest, tissues have roughly equal amounts of oxy- and deoxyhaemoglobin. When metabolic activity is increased, more oxygen is extracted from the capillaries, increasing blood flow and causing a change in deoxyhaemoglobin and thus the MR signal. The effects are very short-lived and so need very rapid sequences such as EPI or fast GRE. Typically, low-resolution volume images are acquired with long *TE* (about 50 ms) every few seconds while the physiological stimulus is applied. The areas of the subtracted images (stimulus minus rest) that show increased signal intensity correspond to the brain area activated by the stimulus. The subtracted images can then be overlaid on to a normal high-resolution image of the area to provide the functional map.

Magnetic resonance spectroscopic imaging

Magnetic resonance spectroscopy provides a frequency spectrum fingerprint of the tissue based on its molecular and chemical composition. Peak intensities and

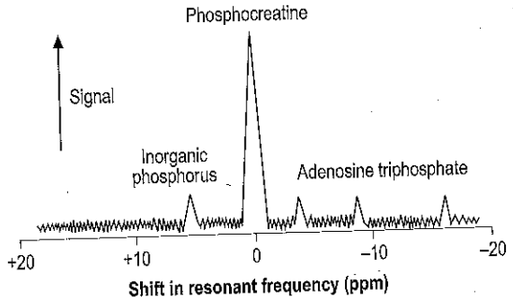

Figure 10.24 Magnetic resonance spectroscopy.

position indicate how an atom is bonded to a molecule. Most clinical spectroscopy studies hydrogen, but other MR-active nuclei are possible.

Nuclei with even numbers of protons and even numbers of neutrons (even Z and even A) cannot show MR ($\gamma = 0$). Those with an odd number of protons or an odd number of neutrons can do so. Different nuclides have different gyromagnetic ratios.

For MRI to be feasible, a nuclide must have a high gyromagnetic ratio, the isotope must be abundant in the element, and the element must be abundant in the human body. Of the four most abundant elements in the human body, ^{1}H is, on all three counts, the easiest to image. ^{16}O, ^{14}N and ^{12}C possess no nuclear magnetism. ^{13}C has an odd number of neutrons but accounts for only 1% of carbon atoms. Naturally abundant ^{31}P has a lower gyromagnetic ratio (17.2 MHz at 1 T) than ^{1}H, but it can be imaged using a sufficiently strong magnetic field and appropriate surface coils. Although the MR signal is several orders of magnitude less than for ^{1}H and so the images have lower spatial resolution (see below), the metabolism of phosphorus is of significant interest as an indicator of energy metabolism and is also used as a monitor of therapy outcome.

Because of chemical shift, phosphorus nuclei have different resonant frequencies when bound in inorganic salts, adenosine triphosphate, phospho monoester, phospho diester and phosphocreatine. Using a broadband RF pulse, all these can be made to resonate. The MR signals from a defined volume of tissue can be analysed as a frequency spectrum (Fig. 10.24), and each of the metabolites can be imaged separately. Sequential imaging allows the study of their metabolism in vivo.

A high magnetic field (2 T or more) is needed to give sufficient signal strength and sufficiently good *spectral* resolution. The field must be uniform to better than 1 ppm. As spectroscopy depends on frequency, only phase-encoding gradients can be used in imaging. Accordingly, to reduce imaging time, much larger pixels in a coarser matrix of 1 cm pixels must be used.

It is not practicable to produce MR images in vivo with other MR active nuclides (such as ^{19}F or ^{13}C) and other metabolites, although they are routinely assessed in vitro, as their low SNR implies very long scan times.

Dixon method for chemical shift imaging

Spinning protons are affected to some extent by the magnetic fields of atomic electrons circulating in nearby atoms. The resonant frequency of a proton is therefore affected by its chemical environment. Measurement of this 'chemical shift' can give information about molecular structure.

The valence electrons in the H–O bond in water produce a slightly smaller magnetic field in the region of the proton than do those in the H–C bond in lipids. As a result, the resonant frequency of the proton is about 3 ppm greater in fat than in water.

The difference in resonant frequencies can be exploited to produce separate images of water and fat:

- After the 90° pulse in a conventional MR scan, the dipoles in water and fat precess at slightly different rates and are continually going in and out of phase with each other every few milliseconds. A (water plus fat) image is obtained by setting the TE when they are exactly in phase, and a (water minus fat) image is obtained by slightly delaying the TE until they are exactly out of phase. Adding these two images produces a water-only image, and subtracting them gives a fat-only image.

10.7 MAGNETIC RESONANCE IMAGE QUALITY

Image quality depends on many parameters, such as the magnetic field strength, the pulse sequence and its timing, number of excitations, slice thickness, slice separation, dimensions of the image matrix and FOV, inherent or injected contrast, and the use of surface coils. In this section, their effect on the signal, noise, image contrast, spatial resolution and scan times are considered. Section 10.8 looks specifically at image artefacts.

Signal to noise ratio

Noise is a random variation in the MR signal, occurring at all frequencies and all the time, due to the following.

- *Patient*. Noise is principally due to the presence of the patient. Random thermal movement of the hydrogen atoms in the tissues induces in the receiver coils currents having a wide range of RFs, called white noise.

- *Scanner*. Electronic noise comes from the statistical fluctuations in the numbers of electrons in the currents flowing in the electronic circuits.
- *Environment*. Some noise can come from RF interference, either outside the scanner room or inside from ancillary equipment.

All noise reduces and obscures contrasts between tissues. It appears worst in the areas of low PD and low signal. The SNR can be improved by any of the following measures, although each needs to be optimized for the image required:

- *Increasing the signal* by:
  - — increasing voxel size by increasing the FOV or slice thickness or by decreasing the number of phase-encoding steps, although at the expense of spatial resolution
  - — decreasing TE
  - — increasing TR or the tip angle.

  SE sequences generally give a bigger signal than gradient echo (GRE). The signal strength would also be increased by using a machine with a higher field strength.
- *Reducing the noise* by:
  - — increasing N_{ex} (the number of excitations)
  - — reducing the bandwidth of the receiver so that it picks up less of the spectrum of noise frequencies, although unfortunately this increases the chemical shift and motion artefacts, described below
  - — reducing cross-talk by having larger gaps between slices or by using multislice techniques
  - — reducing the volume of tissue from which noise is picked up by using well-positioned surface coils of good design such as a phased array with multiple small coils.

  Three-dimensional imaging can give a better SNR than two-dimensional multislice imaging, but at the expense of increased imaging times.

Contrast

Contrast is the difference in the SNR between adjacent tissues and is controlled similarly to the SNR, as above. It can, however, be enhanced, either by using MR techniques to weight PD, T_1 and T_2, depending on the tissue to be imaged (as in section 10.3), or by using a contrast agent.

- *Magnetization transfer contrast* uses off-resonant frequency RF pulses to transfer magnetization to free protons to suppress the signal from protons bound to macromolecules.
- *Fat suppression* can be achieved using a short-TI inversion recovery (STIR) sequence to enhance the contrast between lesions and adjacent fatty tissue.
- T_2 *weighting* specifically increases the contrast between normal and abnormal tissue, the latter being brighter, as it contains water and is more evident than on T_1- or PD-weighted images.
- *Paramagnetic contrast media* such as gadolinium shortens T_1 of nearby hydrogen nuclei in adjacent tissues and thus enhances the inherent contrast. Manganese (as the ion Mn^{3+}), with five unpaired electrons, and iron (as the ion Fe^{2+}), with four, have also been used as paramagnetic or positive contrast agents but are not in such wide clinical use.
- *Superparamagnetic contrast media*. Minute (30 nm) particles of the iron oxide Fe_3O_4 with an inert coating are too small to be ferromagnetic but, being very easily magnetized, they are referred to as superparamagnetic. So too is dysprosium (as the ion Dy^{3+}) DTPA, with five unpaired electrons. Used as contrast agents, they produce local magnetic field gradients that are sufficiently large to shorten T_2^* and T_2. Areas of uptake appear black. They are also called bulk susceptibility or negative contrast agents.
- *Hyperpolarized gas as a new contrast agent*. Hyperpolarized (by laser) xenon (^{129}Xe) is starting to be used for imaging and spectroscopy. Xenon dissolves in blood and shows a large chemical shift. It can be used at very low fields, giving good SNR. It has good potential for MRI of the lungs and for low-field angiography. It is also capable of transferring polarization to ^{13}C for future spectroscopic imaging.

Spatial resolution

Spatial resolution depends on the pixel size, which in turn depends on the matrix and the FOV chosen. A representative value would be 1 mm. Using a larger matrix, reducing the FOV and using a local coil reduce the pixel size and improve resolution. Using a thinner slice also improves resolution, on account of the partial volume effect. Three-dimensional data (volume) acquisition, described above, has high resolution and better SNR than a two-dimensional (multislice) scan of the same voxel size.

Scan time

Scan time is a function of TR, the phase matrix (number of encodings), and the number of excitations (N_{ex}). Short scan times obviously help to reduce motion

artefacts and thus improve image quality, but a compromise is needed.

- *Reducing the* TR:
  - — reduces SNR
  - — reduces the number of slices per acquisition
  - — increases T_1 weighting (tissues more likely to be saturated).
- *Reducing the phase matrix*:
  - — reduces resolution.
- *Reducing the* N_{ex}:
  - — reduces SNR
  - — increases motion artefact.

10.8 ARTEFACTS

Aliasing

Aliasing is a sign that the FOV is too small. It occurs if that part of the patient which is excited by the RF overlaps the chosen FOV. Then signals picked up by the body coil from tissues outside the FOV are falsely allocated to pixels within the matrix. This can produce image wrap-round *in the phase-encoding direction*. Part of the image is shifted bodily to the opposite side from its true anatomy.

The electronic circuits are designed to suppress aliasing in the frequency-encoding direction. Aliasing in the phase-encoding direction can be reduced by:

- An anti-aliasing or over-sampling technique; for example, doubling the FOV in the phase-encoding direction and at the same time doubling the number of phase-encoding steps (to keep the same pixel size), halving N_{ex} to keep the same imaging time and SNR, and displaying only the central half of the image (i.e. covering the original FOV).
- Using a surface coil that more closely matches the FOV, or increasing the FOV to match the coil.

Motion artefacts

Patients may find it hard to keep still during the long imaging time. As well as causing blurring or smearing of the image, motion can be responsible for ghost images. Frequency encoding takes place so quickly and phase encoding so slowly that the effects of motion are usually apparent only *in the phase-encoding direction*. Note that the ghosts appear in this direction even if the motion is through the slice or in the frequency direction. Swapping the frequency and phase direction sometimes displaces the artefact from the area of interest. Cyclical motion of tissues due to heart motion or breathing can produce multiple images in the phase-encoding direction of, for example, the abdominal aorta or any pulsating vessel.

Cardiac triggering can be used to reduce cardiac motion artefacts. The pulse sequence is triggered by the R-wave and *TR* made equal to the R–R interval. The acquisition of data is thereby synchronized with cardiac motion. Respiratory triggering is also possible but more difficult because the motion is slower. Reliance is usually placed on a fast scan with breath holding.

Pulsatile flow in arteries and the chambers of the heart can also produce multiple ghost images (e.g. of the aorta), again in the phase-encoding direction. The faster the motion, the wider is the spacing of the ghosts. Techniques such as the *motion artefact suppression technique* removes artefacts caused by phase changes resulting from movement (e.g. of blood and CSF). It does so by modifying the field gradients and is generally described as gradient moment rephasing.

Magnetic susceptibility

When an iron core is inserted in a DC solenoid, it becomes magnetized and increases the magnetic field. A patient inserted inside a DC solenoid similarly becomes very slightly magnetized. This is an atomic and not a nuclear effect. The patient disturbs the static magnetic field very slightly, especially:

- at the interfaces between tissue (susceptible) and air (not susceptible to magnetization) in lungs, sinuses, etc.
- because of the concentration of haemoglobin that may occur following bleeding, as this is slightly ferromagnetic due to its iron content.

Such unavoidable inhomogeneities of, typically, 2 ppm in the local magnetic field lead to an increase or a decrease in the MR signal and are responsible for certain artefacts. They are likely to be more noticeable on GRE than on SE images.

Chemical shift artefact

The imaging system cannot distinguish whether changes in resonant frequency are due to chemical shift or to the frequency-encoding gradient. Both are interpreted as a displacement *in the frequency-encoding direction*. Compared with water, fat, with its slightly higher resonant frequency, is falsely allocated to locations a few pixels up the gradient. For example, kidney is displaced relative to perinephric fat (Fig. 10.25). A white band is produced where the signal from fat is superimposed on that from water, and a black band where neither produces a signal. Similar effects are seen with fat around the optic nerve and at the margins of vertebral bodies. The stronger the static field of the machine, the greater the shift in terms of pixels.

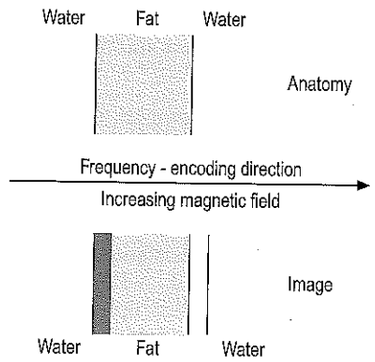

Figure 10.25 Chemical shift artefact.

It can be reduced by using a steeper gradient or a wider receiver bandwidth (although the latter lets through more noise). Alternatively, two chemical shift images can be superimposed with perfect registration between fat and water, thus eliminating chemical shift artefacts.

Other artefacts

A *central line* or 'zipper' artefact is produced across the middle of the image (usually *in the phase-encoding direction*) when RF leaks from the transmitter to the receiver. Line artefacts can also be produced by RF interference from outside, with the patient acting as an aerial (e.g. when the shielded door is ajar).

Implants of *ferrous materials* distort the local magnetic field and can distort or even black out quite a large surrounding area of the image.

Truncation or 'ringing' refers to the parallel striations that can appear at high contrast interfaces (e.g. between fat and muscle or between CSF and the spinal cord). This is similar to the artefact encountered in CT and is due to a low sampling rate. This is more likely *in the phase-encoding direction*. It can be reduced by increasing the matrix or reducing the FOV.

Methods to reduce the more common artefacts are summarized in Table 10.5.

Table 10.5 Common artefacts and solutions

Artefact	Possible solution
Aliasing	Enlarge the FOV or better match surface coils
Chemical shift	Increase frequency bandwidth, reduce FOV or superimposition
Motion	Counsel, immobilize or sedate patient, or swap phase and frequency
Magnetic susceptibility	Remove metal object or use spin–echo
Phase mismapping	Exchange phase and frequency directions, or use gating
Truncation	Increase phase encodings or reduce the FOV

FOV, field of view.

10.9 QUALITY ASSURANCE

As for other imaging modalities, MRI requires a regular quality assurance programme. The homogeneity of the magnetic field is crucial, and it can be measured directly at different positions within the magnet using a special nuclear MR probe or, indirectly, by using imaging test devices.

The same general quality assurance factors as for CT need to be considered: contrast, resolution, noise, artefacts (ghosting) and geometrical distortion. Quality assurance tests should include verification of slice thickness, image uniformity and linearity, SNR, spatial resolution and contrast (using phantoms), RF pulse parameters, and the video display characteristics. Test devices are available designed to fit into the coils to measure each of parameters listed.

Emergency equipment and safety features regarding the special hazards of MRI should also be part of standard checks. Box 10.1 gives a short checklist of quality assurance tests.

10.10 MAGNETS AND COILS

The patient is placed inside the magnet bore and is surrounded by a set of coils (the innermost coils in Fig. 10.26) connected to an RF generator (transmitter or oscillator), which sends through them a pulse of RF current lasting 1 ms or less.

There are three types of main magnet.

- A *permanent magnet* consists of two opposing flat-faced highly magnetized pole pieces (iron and alloys of aluminium, nickel and cobalt are commonly used) fixed to an iron frame. It is large and can weigh some 80 tonnes. It is expensive to buy but the cheapest to run. It requires no power but cannot be shut down and will only give low-strength, vertical fields, up to about 0.3 T. This magnet is used for claustrophobic patients, children, obese adults and interventional procedures.

Box 10.1 Quality assurance checklist

Signal
- Signal to noise ratio
- Uniformity
- Ghosting

Geometrical
- Image scaling
- Distortion
- Slice profile
- Slice width
- Slice position
- Spatial resolution

Functional
- T_1 and T_2
- Imaging speed
- Stability
- Flow

Hardware
- Magnet
- Field contours
- Gradient calibration

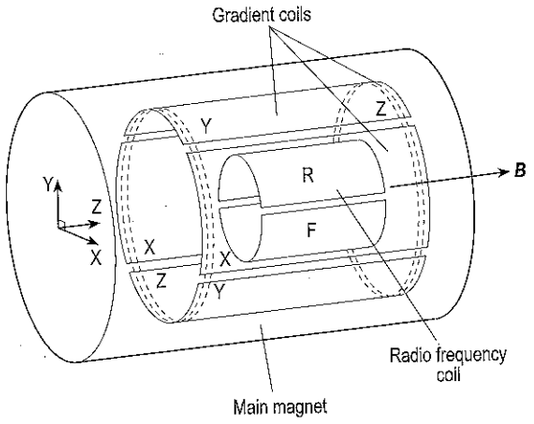

Figure 10.26 Arrangement of the coils in a magnetic resonance machine.

- A *resistive electromagnet* is a set of DC coils with copper or aluminium conductors, which consume some 50–100 kW of power. The heat produced is removed by cooling water, pumped rapidly through the hollow coils. The vertical or horizontal magnetic field is limited by heating to 0.5 T, and has significant fringe fields. It can be switched off at will, at the end of the day or in an emergency. It then takes 15–30 min to 'ramp up', i.e. re-establish the field. It is the cheapest and smallest, weighing some 2 tonnes.
- A *superconducting electromagnet* is a DC solenoid, about 1 m in diameter, with conductors made of a rather brittle niobium–titanium alloy in a copper matrix. They are supercooled by a 'cryogen': liquid helium at 4 K (−269°C). At this temperature, they have negligible resistance, and large (DC) currents can be used without overheating, producing horizontal fields up to at least 3 T but with significant fringe fields. The machine is correspondingly large and expensive and weighs some 6 tonnes. Its tunnel configuration is unsuitable for very large or claustrophobic patients. Small-bore machines up to 7 T are also in use for brain imaging.

  It takes several hours for the coil to cool down and the current to build up. The coil is then short circuited and the power removed. The current continues to flow while using virtually no power, but liquid gas is consumed to maintain the low-temperature need for negligible resistance. If and when the machine is shut down, the electromagnetic energy (some 20 kWh) stored within the superconducting coil has to be removed carefully to avoid a 'quench' (see section 10.11).

  The expensive liquid helium is contained within a fragile cryostat (vacuum, Dewar or Thermos vessels) and replenished periodically. A refrigerator system is used to reduce helium losses. Care must be taken when replenishing the cryogen. Air entering the system would solidify like a plug.

  The coolant level must be logged daily. If it falls too low, quenching occurs; the temperature rises, superconductivity is lost and the stored energy released. If the temperature rises, the liquid gas boils off rapidly and must be vented outside the building. As the superconductivity disappears, the copper matrix takes over the conduction of current.

The magnetic field

The main field must be stable, unaffected by ambient temperature, and uniform to 5 ppm over a large volume for imaging (1 ppm for spectroscopy). In the case of a permanent magnet, the field is usually aligned vertically, from front to back of the supine patient, while in a solenoid it points horizontally along the length of the patient.

The magnetic field lines form closed loops and crowd together within a solenoid but spread widely outside it as a *fringe field* (Fig. 10.27). This effect is reduced by an iron shroud weighing many tonnes or by additional large shimming coils. The fringe field of a permanent magnet is negligible, as it is concentrated within the iron yoke.

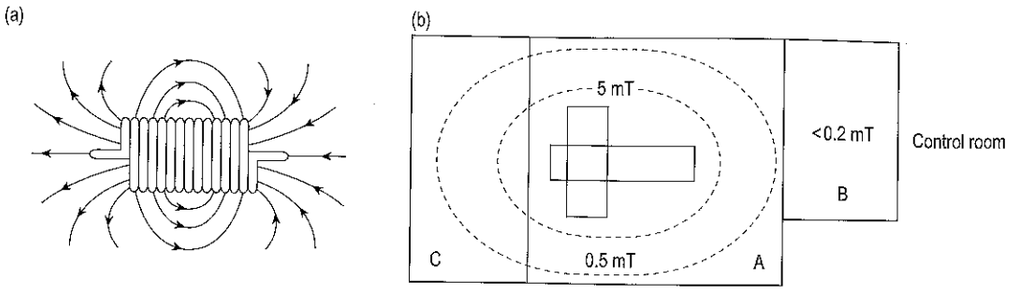

Figure 10.27 Fringe fields (a) and field contour plots (b) showing controlled areas. A, scanner room; C, equipment room; B, operator control area/room.

Optimum field strength The static magnetic field should be large enough to produce an adequate signal but not so large that it exceeds the safety guidelines (see section 10.11). Opinions differ about the optimum field strength for MRI. *Against a high field* is an increased T_1, necessitating a longer *TR* and imaging time and the greater cost of the magnet. The static field is harder to make uniform, and there is a stronger fringe field, which can affect equipment in adjacent areas. Chemical shift artefacts are increased unless a higher field gradient is also used. Motion and susceptibility artefacts are worsened. Potential hazards to the patient, including RF heating, are greater. *In favour of a high field* is a larger MR signal and improved SNR producing optimum images, and special applications become possible.

Coils

Working inwards from the outer main magnet coils (see Fig. 10.26), there are the following.

- The *shim coils* (not shown) carrying DC, which are fine tuned to make the main magnetic field as uniform as possible throughout the imaging volume.
- The three sets of *gradient coils*, carrying DC, which are varied to alter the slope of the magnetic field, typically 20 mT m^{-1}. The coils are connected to gradient amplifiers that control the rise time and maximum value of the gradients. The currents must be switched off rapidly, in 1 ms or less, which causes the coils to emit a loud bang. The direction the current is passed through the coil determines the increase or decrease in the field strength relative to the centre. The slice select or *Z*-axis gradient field is switched on during the application of the RF pulse. The steeper the gradient, the thinner the slice.

  During the few milliseconds that the MR echo signal is being received, DC is passed through a second set of gradient coils (XX in Fig. 10.26). This produces a magnetic field gradient from side to side, in the *X*-direction (Fig. 10.15b), for example for frequency encoding. Steep frequency encoding gradients are needed for small FOV.

  Immediately after the protons in the slice have been excited by the 90° pulse, but before they are inverted by the 180° pulse, DC is passed for a few milliseconds through a third set of gradient coils (YY in Fig. 10.26). This produces a magnetic field gradient from the front to the back of the patient, in the *Y*-direction (e.g. for phase encoding). Steep-phase encoding gradients produce fine-phase matrices.

- RF (transmitter/receiver) coils, which are tuned like a radio to the resonant frequency. They produce a magnetic field at right angles to the main field. To maximize the signal, the coil should be as close as possible to the part being imaged. The RF coils are of several types.
  - The standard *body coil* is usually a permanent part of the scanner. It is used to transmit the RF pulse for all types of scan and to receive the MR signal when imaging large parts of the body (e.g. the chest and abdomen). The patient should be positioned so that the coil includes the anatomy to be imaged.
  - The *head* (transmit/receiver) *coil* is part of the helmet used in brain scanning.
  - *Surface* or local (receiver) *coils* are separate coils designed to be applied as close as possible to image parts close to the surface, the lumbar spine, knee, orbit, etc., before the patient is inserted into the machine. They receive signals effectively from a depth equal to the coil radius. They allow smaller voxels and give better resolution but have a smaller FOV and less

uniformity. They are harder to use and must be positioned carefully. Being closer to the patient than body coils, they pick up a larger signal. Having a smaller FOV and limited penetration, they pick up less noise, thus improving the SNR. Intracavity (e.g. rectal) coils can also be used to give greater resolution of small structures deep in the body.

— *Phased array coils* are multiple (four or more) receiver coils whose signals are received individually (less noise) and then combined to increase SNR but with a large FOV. All data are acquired in a single sequence. The greater the number of independent coils, the more difficult it is to keep them decoupled without losing the signal. Geometrical and electrical methods are needed for this. With a 64-channel array, MR cine at 125 frames s^{-1} has been achieved at 1.5T.

— *Transmit phased array coils* produce a current on each element. Special amplifiers are needed to define this current. Careful control of the relative amplitude and phase of each element enables them to be mutually independent. With reduced pulse duration, higher SNR, improved field homogeneity and, incidentally, reduced specific absorption rate (see section 10.11) at high field strengths, parallel MRI can be used in the excitation phase as in Transmit SENSE.

The complete magnetic resonance system

In addition to the magnet, coils and their controllers in Figure 10.24, there is a pulse control unit. The pulse sequences are selected at the operator console, and the control unit synchronizes the gradients and RF pulses for the selected parameters. The main computer has an array processor (for the Fourier transforms) and sends the data to the image processor (Fig. 10.28).

Siting of the machine should take account of:

- steel girders and reinforced concrete, which may become magnetized in the fringe field, and moving elevators and vehicles, all of which can distort the main field
- lifts and power cables, which may cause RF interference and so distort the image and also produce linear artefacts.

The walls of the room incorporate wire mesh (a Faraday cage) to screen the scanner from such external RF interference. Doors are similarly screened and also interlocked to ensure that they are properly closed during imaging.

10.11 HAZARDS AND SAFE PRACTICE

Magnetic resonance imaging does not involve ionizing radiation. As at present practised, following (in the UK) Medicines and Healthcare products Regulatory Agency (MHRA) guidelines based on the Health Protection Agency advice and implementing European directives, the threshold for the following potential hazards do not appear to be exceeded; certainly, no acute ill effects have been noted.

Static magnetic field (always present)

Voltages might be induced in flowing blood, which could cause depolarization, and, in moving heart muscle, changes may be seen in the electrocardiogram. No adverse effects are expected if fields do not exceed the following MHRA guidelines for whole body exposure of patients:

Normal mode:	less than 2.5T
Controlled mode:	between 2.5T and 4T
Research or experimental mode:	more than 4T

Pregnant patients should not be exposed above 2.5T. In the **controlled** mode, patients must have a panic button and be monitored with constant visual contact with the possibility of verbal contact. Using a pulse oximeter (fibre-optic) is also recommended. Note that the patient's electrocardiogram is affected above 0.3T but returns to normal after exposure. Ethics committee approval is required above 4T, as adverse bioeffects may occur.

Staff should not be exposed to more than 2T whole body and 5T for limbs. Over 24h, the average exposure should not exceed 0.2T. Manufacturers supply field plots around their machines to show 200mT field lines (0.2T). Table 10.6 gives the restricted exposure times.

Note that MRI is contraindicated if the patient has an implanted pacemaker. Anyone with a cardiac pacemaker should be excluded from MR areas and where stray fields are greater than 0.5mT. Consequently, this defines the controlled area for safety purposes around the scanner installation.

Time-varying gradient fields (dB/dt)

Electric fields are produced perpendicular to the gradient field, inducing eddy currents in conductive tissues and causing stimulation, for example peripheral nerve stimulation, involuntary muscular contraction, breathing difficulties and even ventricular fibrillation. Peripheral nerve stimulation has the lowest threshold, about 60 T s^{-1} for a rise time less than 1ms, but it does vary between patients.

Particular care should be taken of patients with heart disease. Other effects are flashes of light (magnetophosphenes) on the retina, vertigo, nausea and sensations of metallic taste all experienced with dB/dt for fields greater than 3T.

Symptoms associated with time-varying gradient fields do not seem to occur below 20 T s^{-1}. There is not thought to be any effect on fetal development but, as a precaution, MRI is usually not carried out during the first trimester of a pregnancy and pregnant staff may be redeployed.

Implanted devices and monitoring equipment may be affected by dB/dt-induced voltages (e.g. cochlear implants, cardiac pacemakers and electrocardiography monitors). Devices need to be specified as **MR compatible** or **MR safe** to be unaffected by the MR scanner and safe to use. This specification also indicates they do not affect the image quality. Note that **MR conditional** labels indicate equipment that may be taken into the scanner room but may be affected by the fields and may cause image distortion.

Acoustic noise associated with fast-switching magnetic fields (GRE) increases with field strength and with higher gradient amplitudes (e.g. shorter *TR*, *TE*, high resolution, thin slices). The machine limit is 140dB, although most do not exceed 120dB. However, hearing protection is required to prevent irreversible damage at 90dB. Earplugs reduce noise by 10–30dB and alleviate patient discomfort and distress. Hearing protection should be matched to the frequency spectrum of the noise produced. Note also that low-frequency sounds are transmitted to the fetus and, although not yet proven, damage to hearing development is of concern.

Radiofrequency fields

Microwave heating may occur, especially at the higher frequencies associated with strong static fields. It is usually compensated by vasodilation. The cornea, with no blood supply, and the testes, with little, may be at risk. Heating of metallic implants may also present a problem. Skin and rectal temperature rise may be monitored and should not exceed 1°C.

The *specific absorption ratio* (SAR) is the RF energy deposited per mass of tissue expressed as watts per kilogram. Restricting whole body SAR in the body to an average of 1 W kg^{-1} restricts the whole body temperature rise to 0.5°C. Table 10.7 gives recommended exposure limits. The patient's weight is needed to calculate the temperature rise and control pulse sequences to ensure safe heating levels.

The SAR is greater for large body parts than for small, for high static fields than for low, for a 180° pulse than for a 90° pulse, for SE than for GRE, and for high-conductivity tissues (brain, blood, liver and CSF) than for low-conductivity tissues (fat and bone marrow). There may be some hotspots, and some combination of imaging parameters may not be allowed.

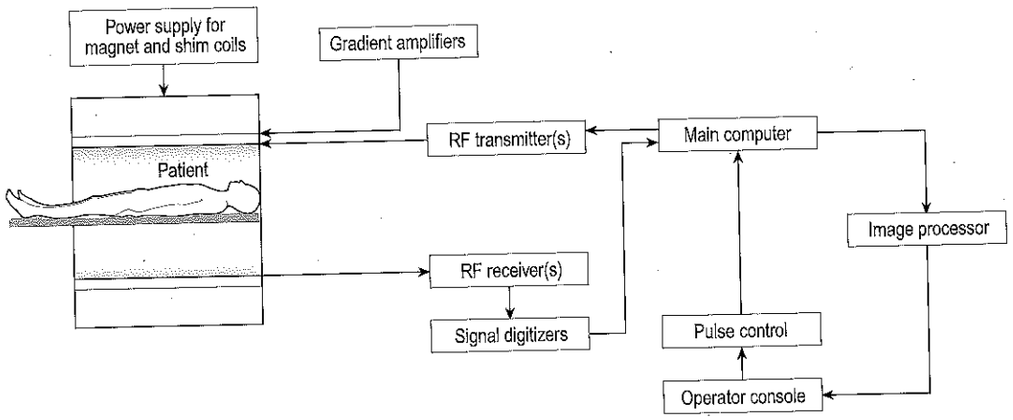

Figure 10.28 Typical system configuration. RF, radiofrequency.

Table 10.6 Restricted whole body exposure times

Exposure time (h)	Average field strength (T)
24	0.2
8	0.6
4	1.2
2	2.0
1	2.0 (limit)

Table 10.7 Recommended specific absorption ratio whole body exposure limits

Exposure time (t, min)	SAR limit (W kg^{-1})	
	<0.5°C	<1.0°C
<15	2.0	4.0
15–30	30/t	60/t
>30	1.0	2.0

Only MR-compatible monitoring equipment should be used in the imaging room, and special care taken over the disposition of leads to minimize induced currents. The majority of reported adverse incidents concern burns to the patient, often from monitoring leads and electrodes.

Other hazards

Protocols for the operation of the MR machine take account of other potential hazards, such as the following.

Mechanical attraction of ferromagnetic objects varies as the square of the magnetic field and the inverse cube of the distance (projectile effect). The fringe field, which can extend for a few metres (as in Fig. 10.27b), may convert scissors and scalpels into potentially lethal projectiles. Oxygen cylinders, patient beds, firefighting apparatus, etc. have all caused major incidents and injury. Aneurysm clips may be displaced or rotated in the tissues when the patient is inserted into the magnet. Non-magnetic, MR-compatible materials should always be used. MRI may be contraindicated in case of ferrous foreign bodies, especially near the eye. Joint and dental prostheses are firmly fixed and should present no problem.

The *fringe field* can also affect some watches, destroy data on computer disks and credit cards, distort nearby video displays, and affect photomultipliers. It is minimized by the design of the coil and the use of iron shielding. The area around the main magnet is designated as *controlled* (see Fig. 10.27) and must be carefully supervised. On account of the effect on implanted pacemakers, free access of the public is limited to areas outside the controlled area where the field is less than 0.5 mT.

Safety procedures

Patient screening Patients must be screened or interviewed before any examination, in accordance with the local imaging protocol. Safety questions must cover implants (especially pacemakers), surgical history, functional disorders, allergies (contrast agents), presence of metallic objects (internal, fixed externally or removable) and weight (for the SAR). Much advice is available on implantable devices, metallic foreign bodies and transdermal patches, tattoos, etc. Appropriate clothing should be provided, and patients should remove hairpins, jewellery and even eye make-up (as this can contain iron oxide, creating artefacts in brain images).

Requirements for safe imaging Once cleared for imaging, patient comfort, safety and confidence are crucial. Successful and safe imaging requires:

- positioning equipment leads to avoid RF burns
- using MR-compatible foam pads for comfort while ensuring that pillows and covers do not inhibit heat loss
- providing music, human contact or even light sedation, as appropriate, to minimize claustrophobia
- hearing protection against the loud percussive noise produced by the repeatedly switching gradient – essential for anaesthetized patients
- visual monitoring and, when appropriate, physiological monitoring using MR-compatible equipment.

Note that only appropriately trained and experienced anaesthetists should attend the patient, as special equipment and procedures are necessary.

Emergencies

Written procedures to cover emergencies and avoid panic reactions include the following.

Cardiac arrest The procedure comprises resuscitation to keep airways open and cardiac massage while the patient is removed from the magnet on to an MR-compatible trolley and taken quickly to the resuscitation area outside the controlled area, where the resuscitation team will take over. Or, if the scanner has a resistive magnet, it is switched off to enable prompt access by the resuscitation team. All equipment is removed before switching the magnet on again.

Fire Resistive magnets should be switched off. Permanent magnets have lower fringe fields, but firefighting equipment should be used only at a distance of 1 m or more from the bore. Superconducting magnets should be quenched only if the firemen need to enter the inner controlled area. Note that non-ferrous carbon dioxide extinguishers should be used.

Quench Before initiating a controlled quench in the event of a fire or when someone is trapped to the magnet by a metal object, the door should be fixed open to avoid a build-up of pressure in the scan room. All scan rooms contain an oxygen monitor and a gas-venting system. External venting channels should be checked periodically to ensure that they function. Accidental quenching, releasing helium cooling gas into the room, might cause suffocation or produce frostbite. The glass window between the scanner and the control room should be broken to equalize the pressure if necessary.

Table 10.8 Favourable effects of changing imaging parameters

Parameter	Effect of parameter	
	when increased	when decreased
TR	SNR increases Allows more slices	Scan time shortens T_1 contrast may increase
TE	T_2 contrast may increase	SNR increases Allows more slices
Slice thickness	SNR increases Larger volume scanned	Spatial resolution increases Partial volume effects lessen
Slice interspace	Larger volume scanned Cross-talk reduces	Less chance of pathology escaping detection
Matrix	Smaller pixels Spatial resolution improves	Larger pixels SNR increases Scan time shortens
N_{ex}	SNR increases	Scan time shortens
FOV	Image covers larger area SNR improves Aliasing artefacts less likely	Spatial resolution improves
	when body or head coil used	**when local coil used**
Coils	Gives large FOV	SNR increases Motion artefacts less likely Aliasing artefacts less likely

FOV, field of view; SNR, signal to noise ratio.

10.12 SUMMARY

- Table 10.8 lists the favourable effects of changing imaging parameters.
- MRI measures the hydrogen content of individual voxels in each transverse slice of the patient and represents it as a shade of grey or colour in the corresponding image pixel on the screen.
- The patient is placed in a strong electromagnetic field for an MRI scan.
- Hydrogen nuclei (protons) in the body align themselves parallel or antiparallel with the magnetic field.
- For each transverse image slice, a short, powerful radiosignal is sent through the patient's body, perpendicular to the main magnetic field.
- The hydrogen atoms, which have the same frequency as the radiowave, resonate with the RF wave.
- The hydrogen atoms return to their original energy state, releasing their excitation energy as an RF signal (the MR signal) when the input radiowave is turned off. The time this takes, relaxation time, depends on the type of tissue.
- The time and signals are computer-analysed and an image is reconstructed.
- The favourable effects of increasing each MRI parameter are also the drawbacks of decreasing it, and vice versa.
- Soft tissue contrast is high. The range of T_1 and T_2 values in soft tissue is even wider than the range of CT numbers.
- Bone and air do not produce artefacts.
- MRI is non-invasive, contrast media being required only for specialized techniques.
- Images can be obtained simultaneously in a number of planes at any angle.
- MHRA guidelines should be incorporated into safety protocols to avoid potential hazards to patients and staff.
- Ionizing radiation is not involved.