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Systems and safety: MR hardware and fields

INTRODUCTION

Magnetic resonance imaging (MRI) has grown, from its initial development in the late 1970s to early 1980s, to become one of the most utilized diagnostic imaging modalities. In 2015 there were 103 million MR examinations performed in hospitals from a population of 1.1 billion people in 29 developed countries. A total of 33 000 scanners were in use in 36 countries serving a combined population of 1.7 billion [1].

The two greatest advantages of MRI are its superior soft tissue discrimination compared to X-rays or CT, and a lack of exposure to ionizing radiation. MRI uses a combination of magnetic fields of varying frequencies: radiofrequency in the megahertz (MHz) region; audio or "very low frequencies" (VLF) up to tens of kilohertz (kHz); and a static field (zero hertz). None of these possesses sufficiently localized concentrations of electromagnetic (EM) energy to damage atoms, molecules, or cells (Figure 1.1). The risk of cancer induction from magnetic field exposures encountered in MRI is quite possibly zero – unlike X-rays, CT, mammography, or the radioactive tracers used in nuclear medicine. This makes MRI very attractive for serial examinations, for scans of children whose tissues are more sensitive to the ionizing radiation used in alternative modalities, or for research studies on groups of healthy volunteers.

So, is MRI safe?

Obviously not, or there would be no need for this book. Whilst later chapters will show that MRI is relatively benign from a biological point of view, the *practice* of MRI may involve significant risk to the patient and to others present during the examination. The MRI examination room is potentially the most hazardous environment within the radiology department because of the possibility of catastrophic and fatal accidents where practice is poor or where safety protocols are not fully observed or understood.

Nowhere is this better illustrated than in the tragic case of a six-year old boy who in 2001 was struck by an oxygen tank which had flown into the scanner, later dying from his injuries. This prompted a root and branches review of MR safety practice within the USA by the American College of Radiology [2] leading to a series of recommendations. It is concerning, that even today, not all these recommendations are routinely followed in every institution. In a 10-year review of MRI-related incidents reported to the US Food and Drug Authority (FDA) 59% were thermal (excessive heating, burns), 11% mechanical (cuts, fractures, slips, falls, crush and lifting injuries), 9% from projectiles, and 6% acoustic (hearing loss) [3] (Figure 1.2).

Essentials of MRI Safety, First Edition. Donald W. McRobbie.

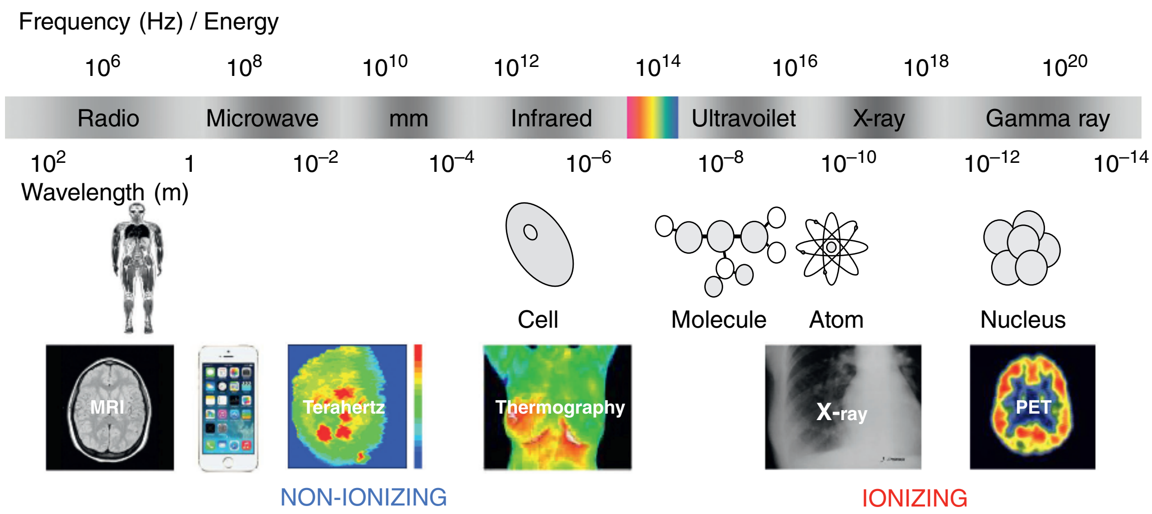

Figure 1.1 The electromagnetic spectrum showing frequency and wavelength of radiations, relative scale, and applications.

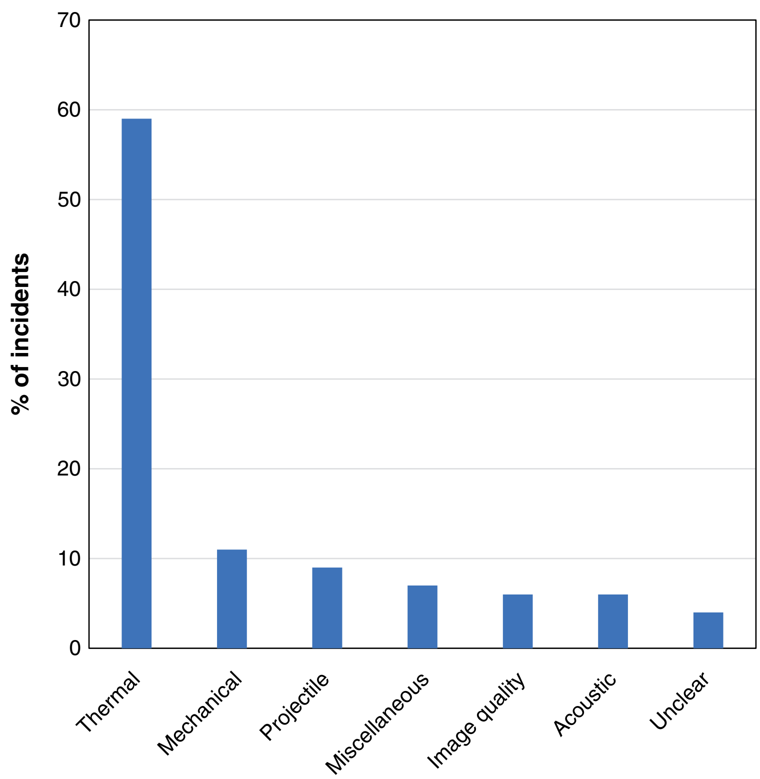

Figure 1.2 MRI adverse events reported to the FDA. Data from [3].

A significant source of risk from MRI arises when patients have implants, particularly active implanted medical devices (AIMDs), such as cardiac pacemakers or neuro-stimulators. However, whereas a decade ago, custom might have been pre-cautionary – not to scan these patients, modern practice is moving towards finding ways to scan whenever there may be significant benefit to the patient. This requires that all MR practitioners have a deeper understanding of the possible interactions between the device, human tissues, and the scanner, and of MR safety in general. That is the purpose of this book, to ensure all MR practitioners have sufficient knowledge to practise safely for the benefit of their patients.

OVERVIEW OF MRI OPERATION

MRI relies upon the properties of nuclear magnetism. The nucleus of an atom consists of subatomic particles: electrically neutral neutrons and positively charged protons. In an atom the electrical charge of the protons is usually balanced by the negative charge of the surrounding electron cloud. MRI concerns the nucleus of hydrogen, mainly as it occurs in water and fat molecules.

Nuclear magnetic resonance

Hydrogen is the simplest element in the universe with the atomic number of one, meaning its nucleus possesses a single proton. The proton is said to exhibit a property known as *spin*. The consequences of spin only become observable in an externally applied magnetic field (denoted B_0) in which the proton spins *precess*, like spinning tops or gyroscopes, around the direction of B_0. In the external field the proton spins must adhere to specific energy levels or quantum basis states (Figure 1.3a). A slight imbalance between the populations of these results in a net magnetization, M_0 (Figure 1.3b). M_0 can be manipulated by applying the appropriate frequency (or energy) of electromagnetic radiation. This is the *Larmor* or *resonance frequency*:

$$f_0 = \gamma B_0 \tag{1.1}$$

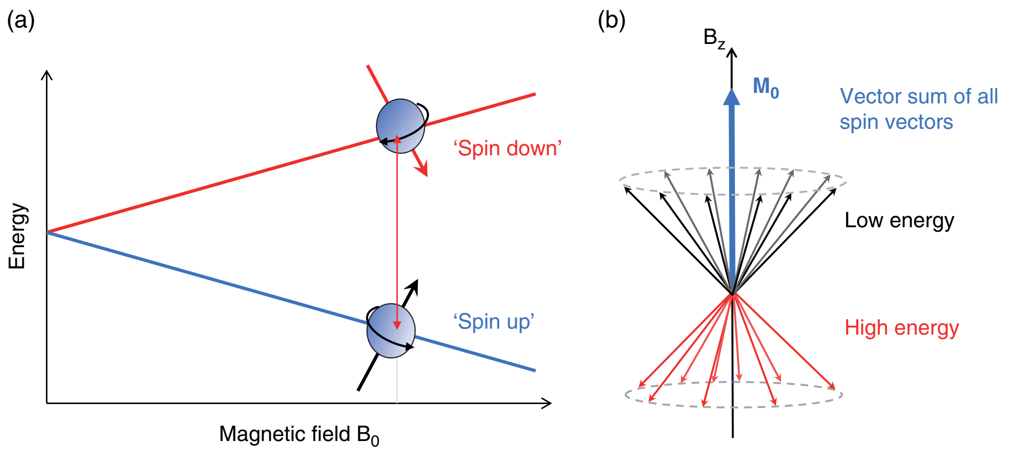

Figure 1.3 Nuclear magnetism: (a) basis state energy differences; (b) formation of macroscopic magnetization M_0 from the sum of basis state spin vectors.

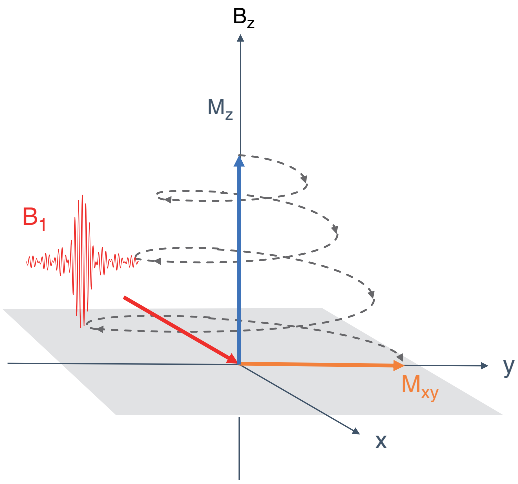

Figure 1.4 Excitation of the macroscopic magnetization **M** by the B_1 RF field.

where the subscript "0" means "at resonance". $\gamma\!\!\!^-$ ("gamma bar") is the *gyromagnetic* ratio of the hydrogen nucleus. When frequency f is expressed in MHz and B_0 in tesla, $\gamma\!\!\!^-$ has a value of approximately 42.58 MHz T^{-1}. This simple relationship underpins all of MRI.

The radiofrequency energy is applied as a magnetic field B_1 orthogonal to the direction of B_0 (Figure 1.4). Whilst B_1 is present the magnetization precesses around both B_0 and B_1 directions, tipping away from the z-axis (usually head–foot) of the scanner. B_1 is applied in a short burst as a RF pulse. The angle of deflection away from the z-axis is known as the flip angle α. For a simple rectangular shaped RF pulse this is

$$\alpha = \gamma B_1 t_p \tag{1.2}$$

where t_p is the duration of the pulse (in seconds), B_1 is the amplitude of the "excitation" pulse (in tesla), and $\gamma = 2\pi \times \gamma\!\!\!^-$ (2.68×10^8 radians s^{-1}).

Example 1.1 B_1 amplitude

What B_1 amplitude is required for a 1 ms rectangular shaped RF pulse to produce a flip angle of 90°?

Express α in radians ($=\frac{\pi}{2}$). From Equation 1.2

$$B_1 = \frac{\alpha}{\gamma\, t_p} = \frac{0.5 \times \pi}{2.68 \times 10^8 \times 0.001} = 5.9\,\mu T$$

Once excited, the magnetization recovers towards its initial equilibrium value M_0 by two independent relaxation processes: *T_1 relaxation* restores the longitudinal or z-component of magnetization towards M_0; *T_2 relaxation* causes the transverse component, the *signal*, to decay to zero. T_1 and T_2 *relaxation times* vary by tissue type and exhibit changes due to pathology, often increasing where disease or injury is present.

Image formation

Image formation is achieved by varying the value of magnetic field in the z- or B_0 direction. The field variation is applied by passing electrical pulses through one or more sets of *gradient coils*, forming the gradient pulses. The gradients, known as G_x, G_y, G_z, are designed to produce linear variations in the z-component of the magnetic field with respect to the x, y, and z axes (Figure 1.5). In terms of their function in image formation they are known as slice-select (G_{SS}), phase-encode (G_{PE}), or frequency-encode (G_{FE}).

Slice selection

By applying a narrow bandwidth B_1 pulse, shaped to include a limited range of frequencies, simultaneously with the slice select gradient G_{ss}, the excitation region is restricted to a narrow slice of the patient's anatomy with a width or thickness:

$$sw = \frac{\Delta f}{\gamma G_{ss}} \tag{1.3}$$

The slice thickness can be controlled by changing the amplitude of the slice-select gradient or by changing the bandwidth of the RF pulse. The slice orientation can be selected by using different gradient coils (or a combination of coils for oblique views). By changing the RF frequency images may be acquired as a series of 2D multiple slices at different locations (Figure 1.6).

In-plane localization

The localization of the MR signal within a slice is usually achieved by two processes: phase-encoding (PE) and frequency-encoding (FE), each using gradient pulses along orthogonal directions. These pulses encode the MR signal in terms of spatial frequencies. Image acquisition requires multiple repetitions of the basic block of a *pulse sequence* using a different amplitude

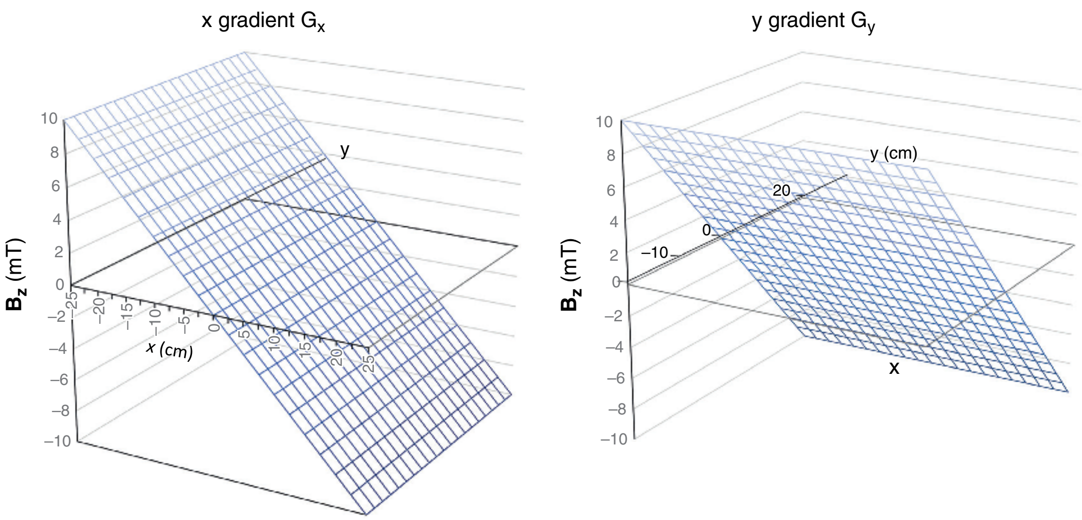

Figure 1.5 B_z from magnetic field gradients G_x and G_y.

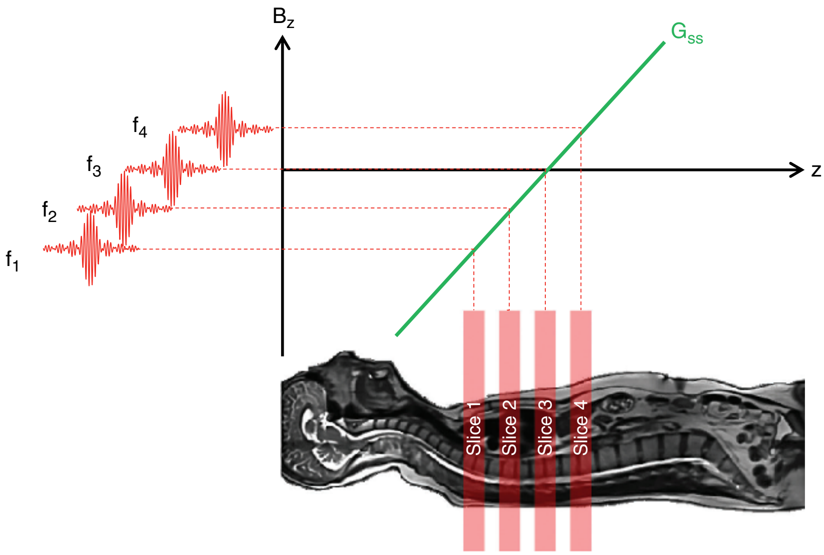

Figure 1.6 Multiple slice imaging. Changing the frequency of each RF pulse whilst a gradient G_{ss} is applied selects a different slice position.

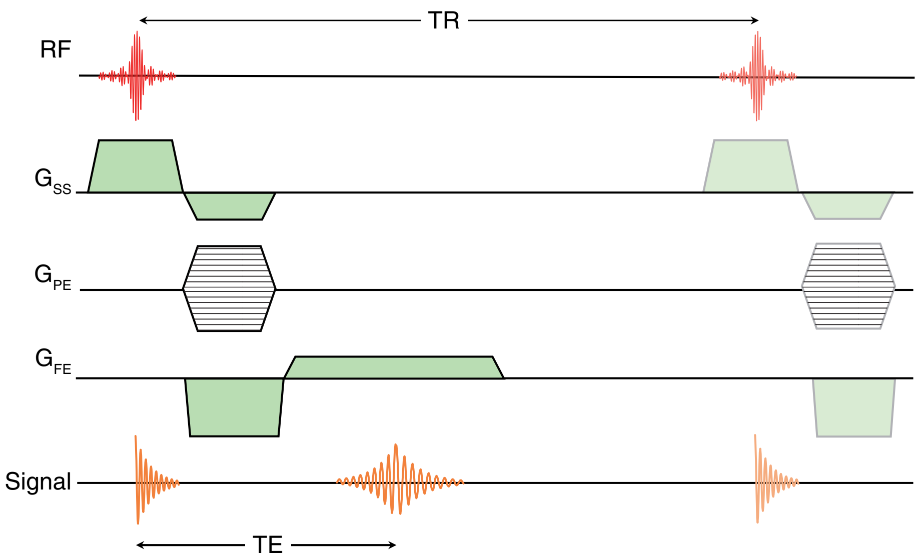

Figure 1.7 Simple 2D pulse gradient echo (GRE) sequence showing pulse amplitudes and timings of the components.

of PE pulse each time (Figure 1.7). TR is the time interval between successive repetitions. Image reconstruction is achieved by the mathematical operation of a two-dimensional (2D) Fourier transform.

The application of a second set of PE gradients in conjunction with the selection of a thicker slab of tissue makes three-dimensional (3D) imaging possible (Figure 1.8).

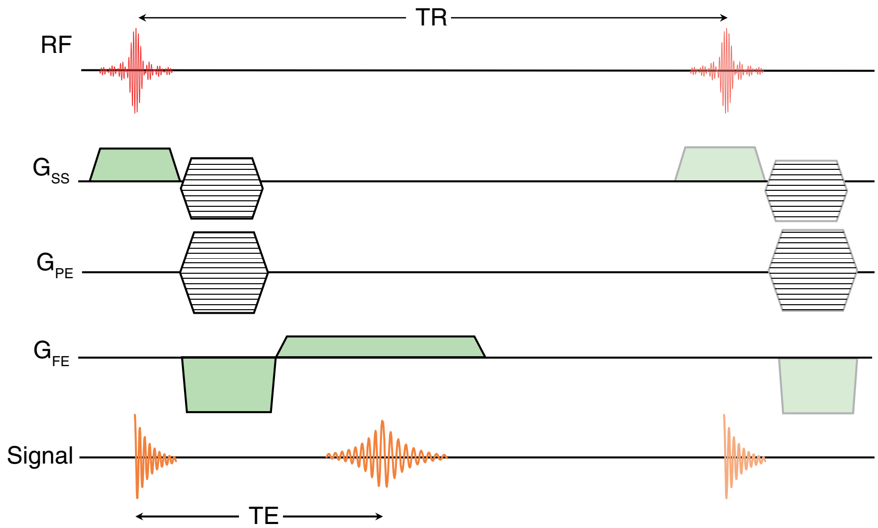

Figure 1.8 3D imaging sequence with a second phase-encode gradient in the slab direction.

Pulse sequences

A clinical scanner has many different pulse sequences available, broadly designated as either *gradient echo* (GRE) or *spin echo* (SE). In GRE (e.g. Figure 1.7) each signal collected arises from a low flip angle RF pulse, typically less than 40°. T_1-weighted images are generated using so-called spoiled gradient echo. Rewound GRE uses slightly higher flip angles (>40°), producing bright-fluid images, popular in cardiac MRI. GRE images are shown in Figure 1.9.

The SE sequence was initially developed for its ability to recover signal loss arising from B_0 inhomogeneities. These occur across the field-of-view (FOV) as dephasing, or "fanning out", of transverse components of the magnetization (Figure 1.10) following a 90° pulse. By applying a 180° pulse orthogonal to the 90° pulse (and also to B_0), the fan of magnetization vectors is twisted around in such a way that those with a phase lag advance in phase and vice-versa. At time TE, equal to twice the interval between the 90° and 180° pulses, the magnetization rephases, giving a strong echo whose magnitude depends upon the tissue T_2.

Spin echo can be further enhanced by using multiple 180° pulses to form a series or train of echoes, each of which can have different phase-encoding applied. In Turbo or Fast Spin Echo (TSE/FSE) the overall scan time is reduced by the echo train length (ETL) or Turbo-factor (TF). Typical ETL/TFs are in the range 3–20, although single shot acquisitions with 128–256 echoes are also possible.

The addition of a preparation 180° pulse prior to the 90° inverts the magnetization to lie along the -z axis. As each tissue recovers towards $+M_0$, there is a time at which its magnetization passes through zero. An image formed at this point, will not contain signal from that tissue. Short TI Inversion Recovery (STIR) removes fat from the image, whilst FLuid Attenuated Inversion Recovery (FLAIR) removes the cerebrospinal fluid (CSF). Typical SE images are shown in Figure 1.11.

Parallel imaging

In parallel imaging a multi-element RF receive array coil is used to provide additional spatial information, and to reduce the number of lines of signal required to form an image. Parallel imaging reduces the number of TR periods of an acquisition by an amount known as the reduction factor R, SENSE factor or iPAT factor. The use of parallel imaging reduces the patient's overall RF exposure.

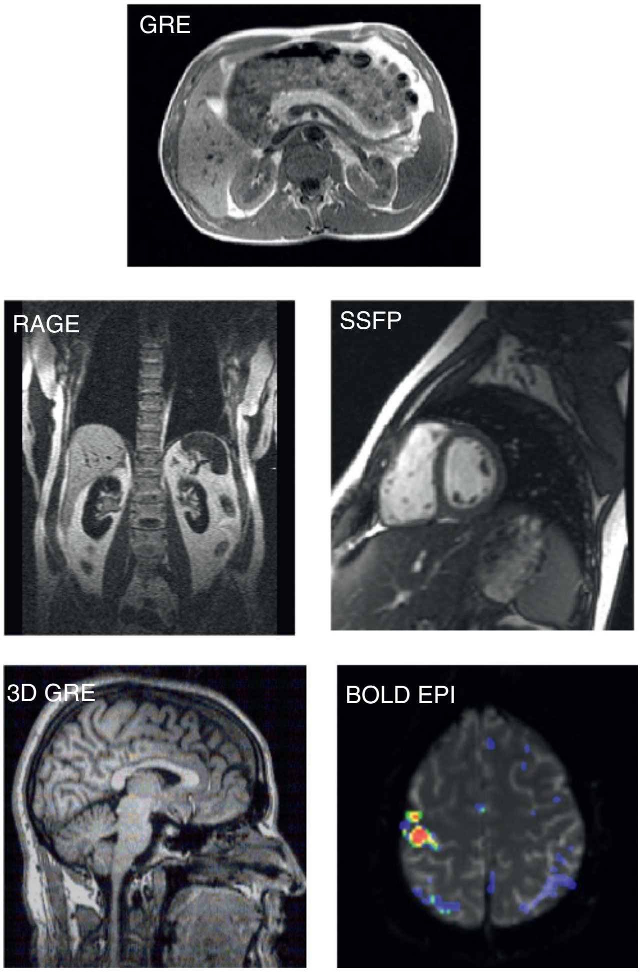

Figure 1.9 Gradient echo images: Gradient echo (GRE) abdomen; Rapid Acquired Gradient Echo (RAGE); 3-D Gradient Echo (3D GRE); Steady State Free Precession (SSFP) heart; BOLD-EPI with brain activation map overlay. Source: Flinders Medical Centre, Adelaide, Australia; Charing Cross Hospital, London, UK. Reproduced with permission.

Overview of MRI applications

Since its adoption in the late 1980s the scope of MRI's clinical applications has grown, and continues to grow. Brain, spine, and musculoskeletal imaging were the first major applications.

MRI's ability to differentiate between grey and white matter in the brain led to its deployment in neuroradiology, particularly for white matter disease and brain tumors. The development of diffusion-weighted imaging (DWI) gives MRI the ability to detect acute stroke and chronic infarct. Functional MRI (fMRI) is a popular tool in neuroscience research which utilizes the Blood Oxygenation Level Dependent (BOLD) effect to map neural activation. White matter connectivity can be investigated using diffusion tensor imaging (DTI) or high angular diffusion imaging (HARDI) and tractography. In musculoskeletal MRI soft tissue components, muscle, bone marrow, fat, and cartilage are all visible. Whilst tendon, ligament, and cortical bone are inferred by their absence of signal, edema resulting from injury is highly conspicuous.

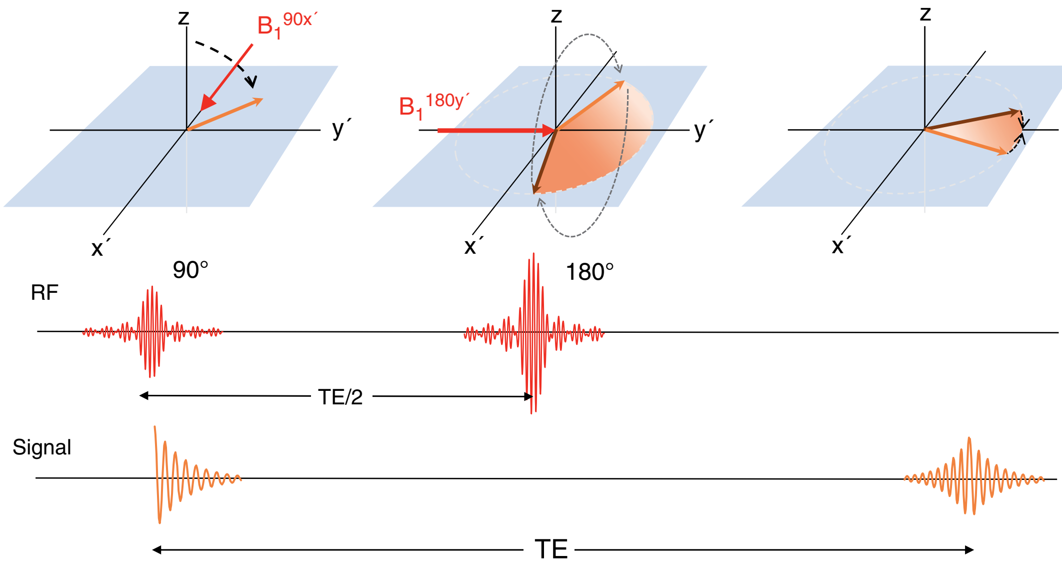

Figure 1.10 Spin echo formation: following a 90° pulse aligned with the x′ axis, magnetization in the x′y′ plane dephases; a 180° pulse aligned with the y′-axis inverts the phase of the magnetization to form a spin echo at time TE. The prime (′) indicates a frame of reference rotating at the Larmor frequency.

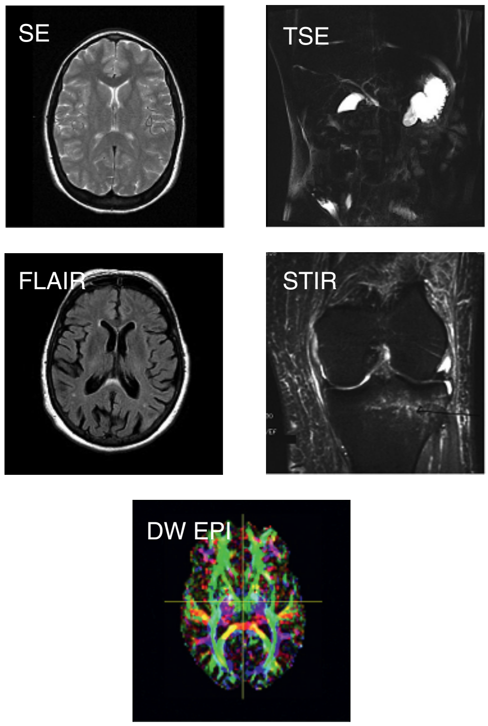

Figure 1.11 Spin echo images: (a) Spin echo (SE) T_2-weighted brain; single shot Turbo (fast) Spin Echo (TSE) MR cholangio-pancreatogram; Fluid Attenuated Inversion Recovery (FLAIR) brain; Short TI Inversion Recovery (STIR) knee; Diffusion EPI (Echo Planar Imaging) showing white matter directional anatomy. Source: Flinders Medical Centre, Adelaide, Australia. Reproduced with permission.

MRI has a major role in oncology through tumor imaging, often using DWI for diagnosis, staging, and treatment assessment. Applications include breast, bowel, prostate, liver, pancreas, female pelvis, in addition to brain and spine. MR *spectroscopy* (MRS) provides in-vivo bio-chemical information on tumor and tissue metabolism. Through sensitizing the MR signal to flow or by the injection of a gadolinium-based contrast agent (GBCA) angiographic images can be obtained to investigate vascular disease. The development of rapid GRE sequences has facilitated cardiac MR for heart morphology and function studies. GRE using the in-phase–out-of-phase technique is applied in liver and abdominal imaging of adenoma and cirrhosis, whilst TSE/FSE can indicate cysts, hepatocellular carcinoma and metastases. Single-shot TSE/FSE is used for MR cholangio–pancreatography (MRCP) in the biliary system. Some of these are illustrated in Figures 1.8 and 1.10.

MRI HARDWARE

The heart of the MRI system is the magnet, creating the static B_0 field that produces the tissue magnetization. The B_1 pulse that manipulates the magnetization is generated by a RF transmit coil (Tx) or coils fed by high-powered broadband RF amplifiers. The pulse shapes are generated digitally and converted to analog waveforms prior to amplification. The gradient coils provide the pulses used for spatial encoding of the signal. Finally, the MR signal is detected by receiver coils (Rx). These are usually positioned close to the anatomy of interest to maximize the signal received. Most receive coils are *array* or matrix coils formed from numerous minimally interacting smaller *elements*. The advantages of array coil technology include improved signal-to-noise ratio (SNR) and the ability to utilize parallel imaging techniques. Once detected, the MR signal is demodulated, i.e. the RF "carrier" component (at f_0) is removed, as the spatial information is stored in the VLF signal region. With "direct digital" systems, demodulation and digitization are applied directly to the amplified RF signal, close to or at the coil. The ensuing signals are transmitted digitally or optically and stored for reconstruction. Figure 1.12 shows a schematic of a typical MR system. The gradient and RF amplifier systems, control and signal processing systems, and cooling equipment are situated in the equipment or technical room, external to the MR examination or magnet room. The console and host computer are located in the control room (Figure 1.13).

Magnet system

The magnet is the largest single component of the system. Most systems use superconducting magnets. Superconductivity is a quantum mechanical property whereby, below a critical temperature T_c, an electrical conductor loses its electrical resistance, enabling large electrical currents to be sustained in perpetuity without a driving voltage from a power supply. As long as the windings are kept sufficiently cold, the current and hence the magnet's field persists indefinitely. Liquid helium with a boiling point of −269 °C (4.3 K or kelvin) is used for cooling. Safety consequences of this are considered in Chapter 12.

Superconductivity

Superconductivity was discovered by Dutch scientist Heike Kamerlingh Onnes in 1911, but it took until 1957 for Bardeen, Cooper, and Schrieffer to formulate a quantum mechanical theory (BCS theory) to account for the phenomenon. The electrical resistance of a non-superconducting metal, such as copper, depends upon its temperature, decreasing with lower temperatures, but possessing a finite resistance even at absolute zero (−273.15 °C). Below T_c the electrons in a

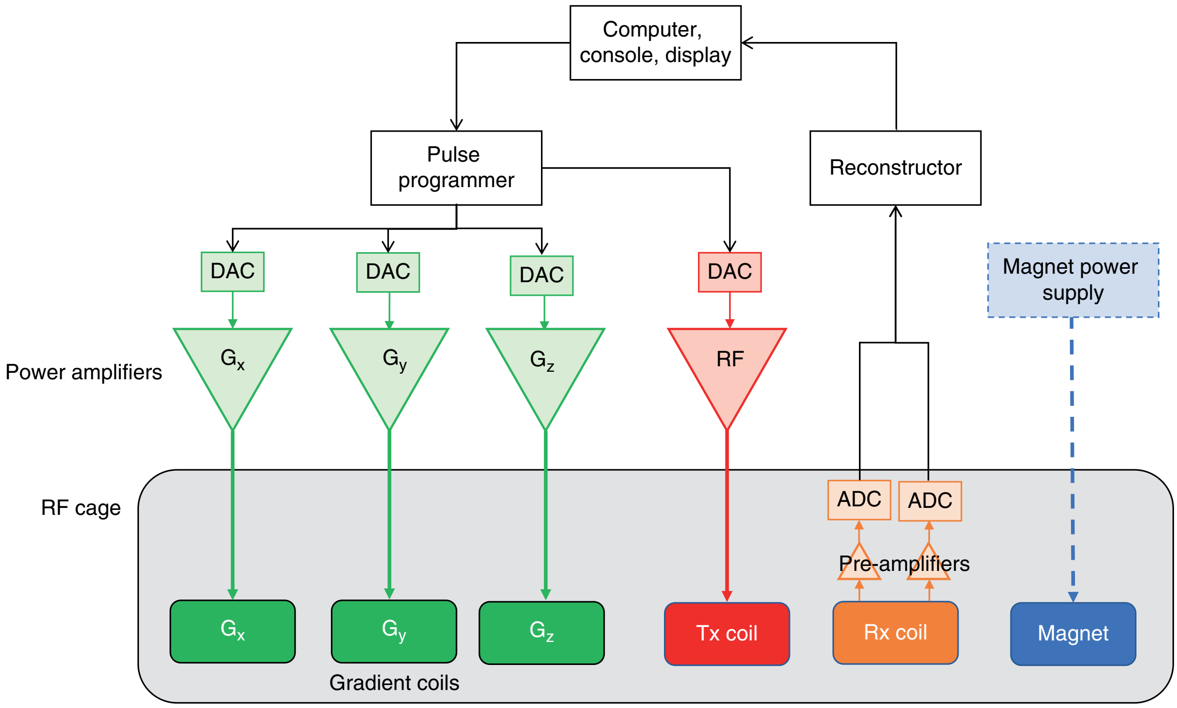

Figure 1.12 Schematic of principal MRI system components: DAC is digital-to-analog convertor; ADC is analog-to-digital convertor; Tx transmit; Rx receive; black lines denote digital signal; colored lines denote analogue signal. The magnet power supply is only required for the initial ramp-up.

superconductor pair up into "Cooper pairs" acting as a superfluid resulting in zero resistance (Figure 1.14a).

Niobium titanium (Nb-Ti) alloy used in MR magnets is a type 2 superconductor.[1] It has a second, higher critical temperature at which some magnetic flux may exist within the material. Superconductivity behaves as a thermodynamic phase with a relationship between temperature, field, and current density (Figure 1.14b). There is a critical field B_c and current density J_c above which the superconductive state cannot exist. This puts an ultimate limit on the field strength that can be achieved. Nb-Ti has a T_c of 9.5 K and B_c of 15 T. Niobium-tin (Nb-Sn) alloy can sustain higher fields.

High temperature superconductors can have T_c above 90K and can be cooled using liquid nitrogen (N) with a boiling point of 77 K (–196 °C) or with cooled helium gas. These have been used to produce 0.5 T MRI magnets, but not operating in a persistent current mode. Research is ongoing with the prospect of simplifying the cooling system and reduced dependence upon helium. Helium is a by-product of natural gas extraction, a limited resource. Nitrogen can be produced from the atmosphere.

Superconducting MR magnets

Superconducting magnets are capable of generating magnetic flux densities, colloquially referred to as "magnetic field strength", up to 7 T (tesla) in currently available systems. 1.5 and 3 T

[1] In type 1 superconductors the magnetic field inside the material is zero due to the Meissner Effect.

Figure 1.13 MRI operator's console in the control room with observation window into the magnet room. Source: James Steidl/iStockphoto.

systems are most common. B_0 is orientated horizontally along the scanner bore or patient aperture, by convention denoted as the z-axis.

Figure 1.15 shows a schematic of a modern MRI magnet in cross-section. There are two sets of windings: the field coils and the shield coils. The shield coils are wound in opposition to the field coils in order to reduce the extent of the *fringe field*. The windings are held in a large vessel or *cryostat* and bathed in liquid helium. Surrounding this is a vacuum to prevent conduction and convection heating, and layers of highly reflective sheeting to prevent radiative heating.

During operation some helium will evaporate or "boil off". In older magnets this was wasted as exhaust, but modern magnets have a refrigeration system, the *cold head* or *cryo-cooler* which re-condenses the gas as liquid. Such "zero boil-off" systems generally do not require helium replenishment. If electrical power is lost, the reliquification will not occur, but the magnet can stay cold for several days. This allows new systems to be transported cold.

To generate the field initially, a power supply is required. The process of energizing the magnet or "ramping up" involves a gradual increase of electrical current. A superconducting switch or shunt is maintained in a non-superconductive state by a small heater whilst the current grows. At the desired current the heater is turned off and the switch becomes superconducting, completing the electrical

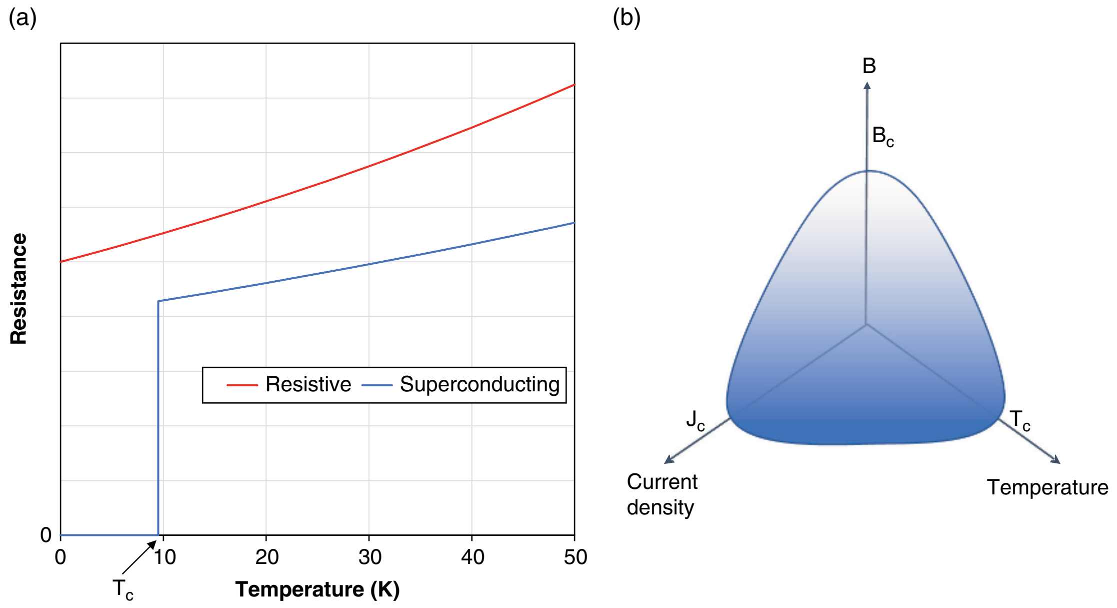

Figure 1.14 Superconductivity: (a) resistance v temperature for a superconductor and a non-superconductor; (b) Superconducting phase diagram: each of temperature, current density and magnetic field must be below a critical value T_c, J_c, B_c to maintain the superconductive state.

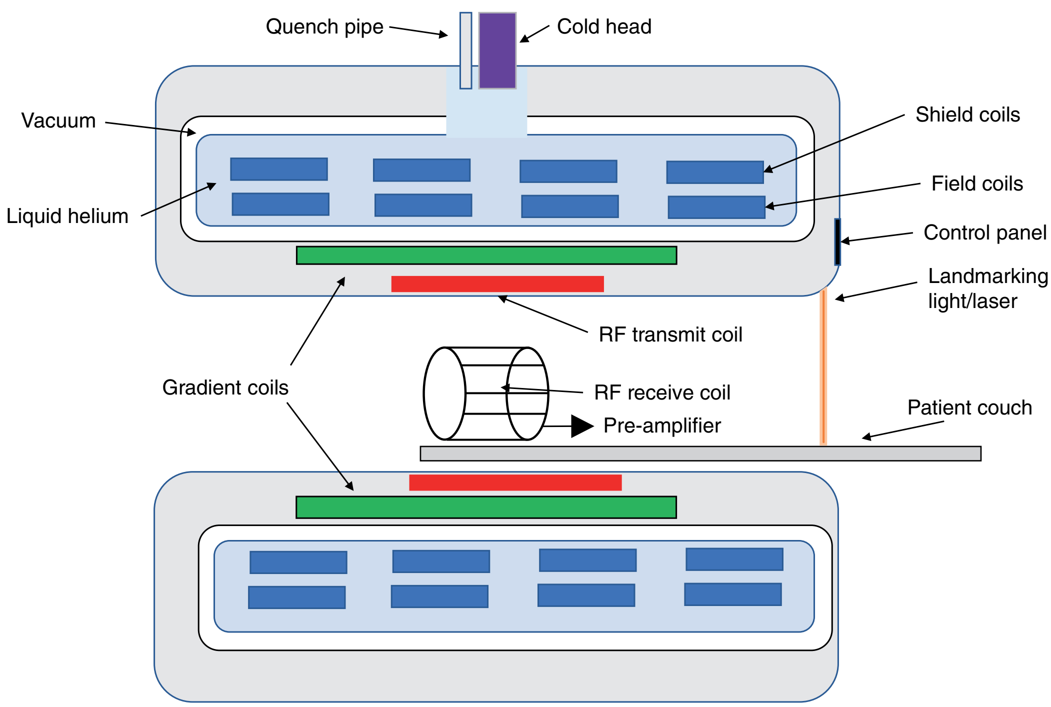

Figure 1.15 Schematic of a self-shielded superconducting MRI system.

circuit. The power supply can then be disconnected and removed from site. The reverse process, ramping down, can be used to reduce or remove the field when required, e.g. for a major hardware upgrade or after a non-injurious ferromagnetic incident to remove the offending object.

The Nb-Ti wires are 50–150 μm in diameter, embedded in a copper matrix. This provides additional mechanical strength – they are subject to significant magnetic forces – and provides a means for conducting excess heat and current in the event of a magnet failure or *quench* to prevent damage to the more delicate Nb-Ti filaments. In the superconducting state, the copper matrix acts like an insulator, providing isolation between the Nb-Ti strands.

Short larger-bore magnets

A recent industry trend has been to reduce the length of the magnet, typical to around 1.6 m and to increase the diameter of the bore from 60 to 70 cm to afford better patient comfort and to accommodate larger patients. This has implications for safety as it affects the fringe field (see Fringe field spatial gradient, page 19).

Other magnets

Other configurations of MRI systems are also available, although less common. Resistive magnets producing fields up to 0.4 T are sometimes configured as open or C-arm systems, affording better access to the patient and a less claustrophobic experience. Resistive magnets have one safety-related advantage: the field can be routinely switched off.

Permanent magnets are used in low field niche scanners for extremity imaging or in “upright systems”. These employ various rare earth materials such as neodymium-iron-boron (Nd-Fe-B). Their magnetic field is always present.

Imaging gradients subsystem

Magnetic field gradients G_x, G_y, and G_z used to spatially select or encode the MR signal during acquisition are generated by three sets of gradient coils. The field generated is *always* along z. Gradient coils usually require water cooling as they have typically hundreds of amperes (A) of electricity pulsed through them. Specialist hybrid amplifiers and power supplies are used to generate these strong pulses. A consequence of gradient pulsing is the generation of acoustic noise (Chapter 7).

Gradient pulses usually have a trapezoidal waveform (Figure 1.16). The ability of the gradient system to switch rapidly, known as the *slew rate* (SR), is defined as the maximum amplitude divided by the *rise time* required to achieve that amplitude:

$$SR = \frac{\text{Maximum amplitude}}{\text{minimum time required}} \tag{1.4}$$

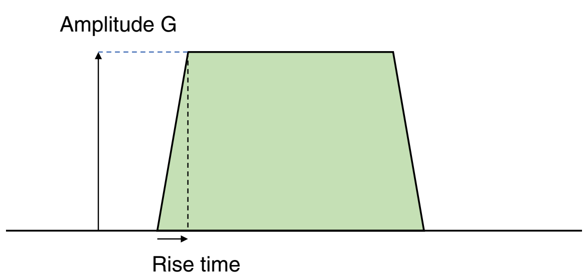

Figure 1.16 Trapezoidal gradient pulse.

Typical slew rates are 100–200 T m^{-1} s^{-1} (tesla per meter per second).

Example 1.2 Gradient performance

What is the (theoretical) maximum field produced by a 40 mTm^{-1} gradient system with a slew rate of 200 T $m^{-1}s^{-1}$? What is the minimum rise time?

Assuming that the gradient is linear over a 50 cm FOV, the maximum amplitude at the edge, 25 cm from the iso-centre is

$$G_{max} = 40 \times 0.25 = 10\ mT.$$

The minimum rise time is

$$t_{min} = \frac{40 \times 10^{-3}}{200} = 0.2\ ms.$$

Radiofrequency subsystem

The radiofrequency system comprises two subsystems: transmit and receive. RF transmit is more important for MR safety.

RF transmission

In most instances a body RF transmit coil is used (Figure 1.17). This typically has a "birdcage" design, operating in *quadrature* to produce a *circularly-polarized* magnetic field B_1. Some coils may operate in transmit and receive mode (T/R) denoted symbolically as in Figure 1.18. Examples are dedicated T/R head and knee coils. Tx coils usually have a cylindrical geometry, entirely encompassing the anatomical region to produce a uniform B_1 so that everything in the FOV experiences the same flip angle.

The coils operate in a resonant mode as tuned circuits, resulting in current amplification to achieve greater B_1 at the Larmor frequency. They are driven by powerful RF amplifiers, rated at tens of kilowatts (kW). An important aspect of RF generation is impedance matching, usually to 50 Ω (ohms), to ensure the maximum power transfer from the amplifier to the coil. B_1 is of the order of micro-tesla (μT) peak amplitude.

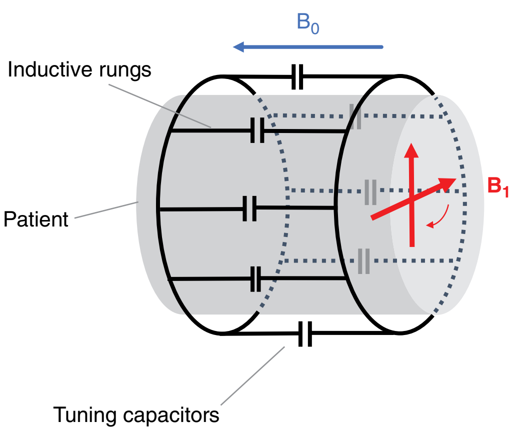

Figure 1.17 Transmit 8-rung 'birdcage' coil to produce a circularly polarised (rotating) B_{1+} field orthogonal to B_0.

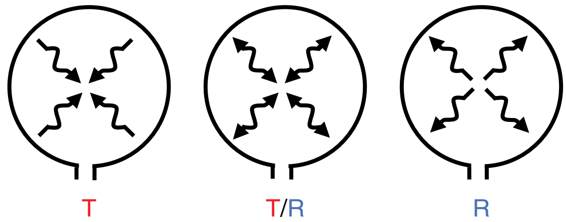

Figure 1.18 IEC 60601-2-33 [4] compliant coil labelling: left- transmit only; middle- transmit-receive; right- receive only.

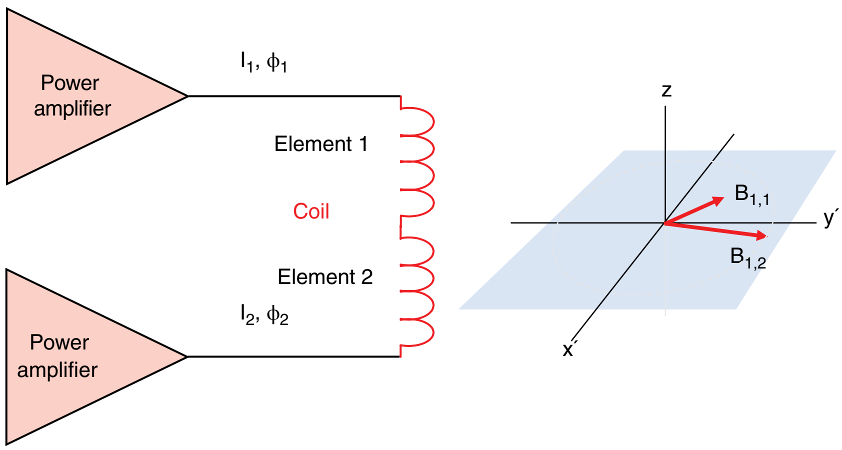

Figure 1.19 Parallel transmit: two (or more) independent RF power amplifiers drive elements of the transmit coil.

In 3 T systems, operating at 128 MHz, the B_1-field in tissue is often quite non-uniform. In this instance *parallel transmit* systems can help. These utilize multi-element Tx coils powered by independent amplifiers capable of changing both the amplitude and phase (relative direction) of the RF pulses (Figure 1.19).

RF reception

The purpose of the RF receiver coils is to detect the tiny (micro-volt) MR signals. A parallel tuned circuit is used (Figure 1.20) to magnify the voltage prior to pre-amplification and further processing. The receive coil requires protection circuitry to prevent the large transmit pulses from coupling into the coil. A simple means of achieving this is to use crossed diodes and a detuning capacitor. During the large Tx pulses the diodes conduct and so the total capacitance becomes the sum of both capacitors and the circuit is off-resonance. During signal detection the diodes do not conduct, and C_d is "invisible." A fault in this circuitry can lead to large induced currents in the coil and potential heating or burns for the patient.

ELECTROMAGNETIC FIELDS

The extent and magnitude of the fields involved in MRI are summarized in Table 1.1 and Figure 1.21.

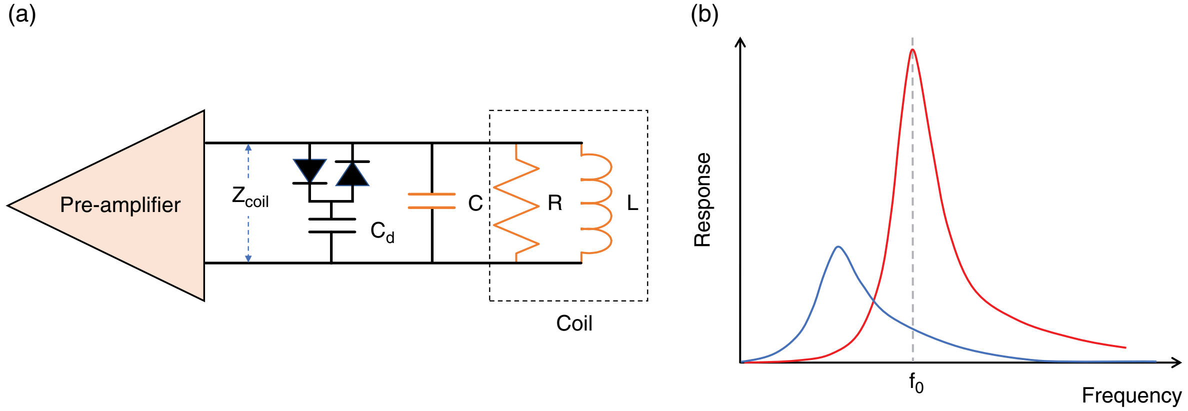

Figure 1.20 Receive coil and pre-amplifier: (a) the coil has inductance L and resistance R; C_d is a detuning capacitor; (b) response of the coil during signal reception (red line) and RF transmission (blue line).

Table 1.1 Magnetic fields in MRI.

Field	Amplitude	Frequency / Slew rate	Pulse duration
Static field B_0	0.2-7 T	0 Hz	Always present
Static fringe field spatial gradient dB/dz	0-25 T m^{-1}	0 Hz	Always present
Imaging gradients G_x, G_y, G_z	0-80 mT m^{-1}	0-10 kHz 0-200 T m^{-1} s^{-1}	0-10 ms
RF transmit field B_1	0-50 μT	8-300 MHz	0-10 ms

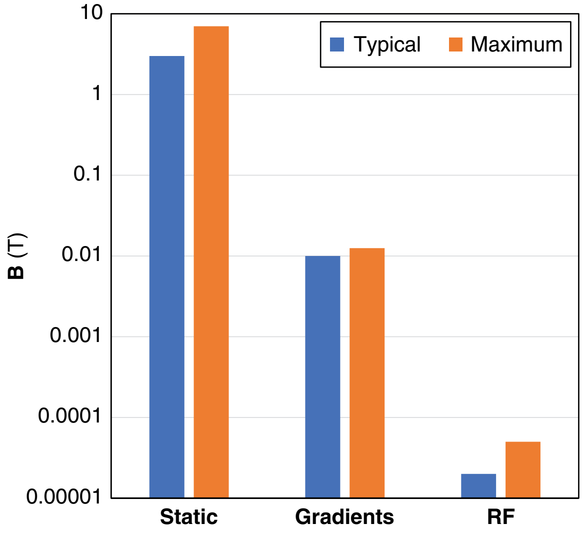

Figure 1.21 Relative magnitude of magnetic fields used in MRI.

Static field

Definition of magnetic flux density and the tesla

Whilst MR practitioners commonly refer to their magnets in terms of "magnetic field strength", this nomenclature is scientifically incorrect. The proper term is *magnetic flux density*, denoted as **B**. **B** is a vector field with components in each direction B_x, B_y and B_z. MRI is only sensitive to B_z and that is what we refer to colloquially as the "field." Magnetic flux density has the SI (International System) unit of the tesla (T). An older unit is the gauss (G). One tesla equals 10 000 G.

The scientific definition of the tesla is in terms of force. Referring to Figure 1.22, one tesla is the amount of magnetic flux density which exerts a force of one newton (N) on a charged particle of charge one coulomb (C) moving at right angles to the field direction with a velocity of one meter per second (m s^{-1}). It's not an easy definition, but the fact that it is defined in terms of force is highly apt for MR safety!

MYTHBUSTER:

The unit of "magnetic field strength" is not the tesla, but is amperes per meter. B is the *magnetic flux density*.

So, what is magnetic field strength in actuality? It is given the symbol **H** and has units of amperes per meter (A m^{-1}). It is defined in terms of a cylindrical electromagnet, just like our scanner – the current in the windings generates an **H**-field. In free space

$$B = \mu_0 H \tag{1.5}$$

μ_0 is the magnetic *permeability* in a vacuum, equal to $4\pi \times 10^{-7}$ henrys per meter (H m^{-1}).

One way of visualizing magnetic fields is through magnetic field lines. If you have ever done the experiment of introducing iron filings to the proximity of a simple bar magnet you may have observed the pattern shown in Figure 1.23a. These illustrate the magnetic "lines of force". A small compass needle positioned anywhere will align with these. We can think of the magnetic flux density as being the intensity of grouping of these lines: the more closely grouped together, the stronger the B-field.

B_0 fringe field

The most uniform and dense grouping of lines of force for an MR magnet (Figure 1.23b) occurs within the bore. As we move away from the bore the lines diverge and consequently the B-field decreases. We call this region the *fringe field*. Your scanner manufacturer provides field maps showing fringe field contours at 0.5, 1, 3, 5, 10, 20, 40, and 200 mT [4] (Figure 1.24). These are important for MRI suite design (Chapter 12). A modern MRI system utilizes self-shielding in order to reduce the spatial extent of the fringe field (Figure 1.25).

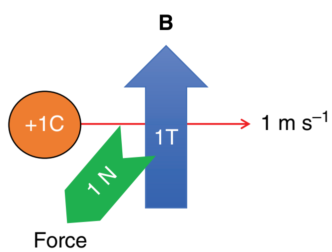

Figure 1.22 Definition of the SI unit tesla.

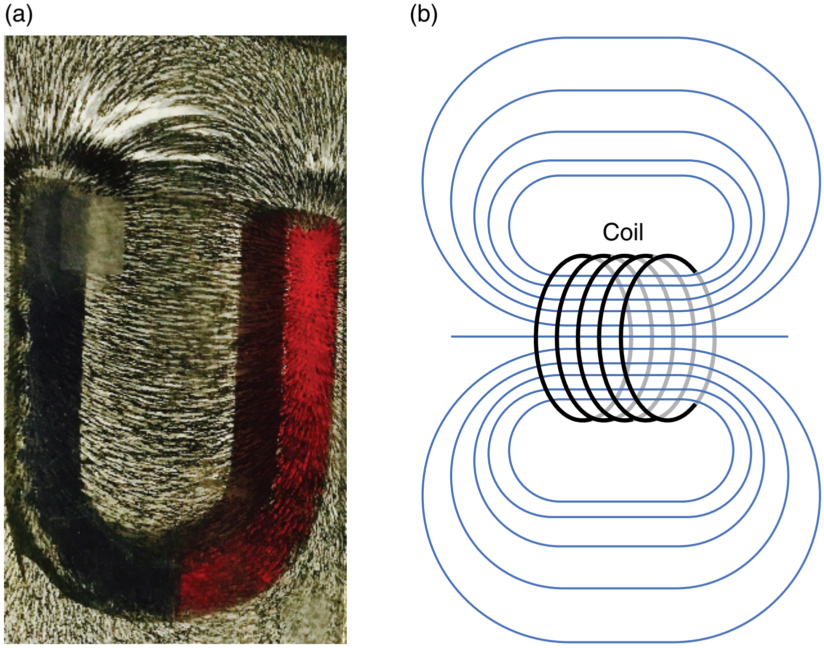

Figure 1.23 Magnetic field lines of force: (a) seen in the pattern of iron filings around a permanent magnet; (b) from an electromagnet.

Fringe field spatial gradient

As we move further from the bore of the magnet, the lines of force diverge, and the fringe field decreases (Figure 1.23b). The amount it decreases with distance is known as the *fringe field spatial gradient*, specified in T m^{-1}. The fringe field spatial gradient is responsible for the attractive force on ferromagnetic objects. Your manufacturer is required to provide you with information about the fringe field gradient. Figure 1.25 shows how the B_0 field and its spatial gradient dB/dz vary along the z-axis. The fringe field is compressed for the shielded magnet but produces a stronger spatial gradient close to the bore entrance. This is highly significant for projectile safety.

MYTHBUSTER:

The fringe spatial field gradient is always present as long as the main static B_0 field exists. It should not be confused with the imaging gradients.

The imaging gradients

Gradient amplitude is measured in mT m^{-1} (milli-tesla per meter). When a gradient pulse is applied, e.g. along the x-axis, the total B experienced at a point x is

$$B(x) = B_0 + x\,G_x \quad (1.6)$$

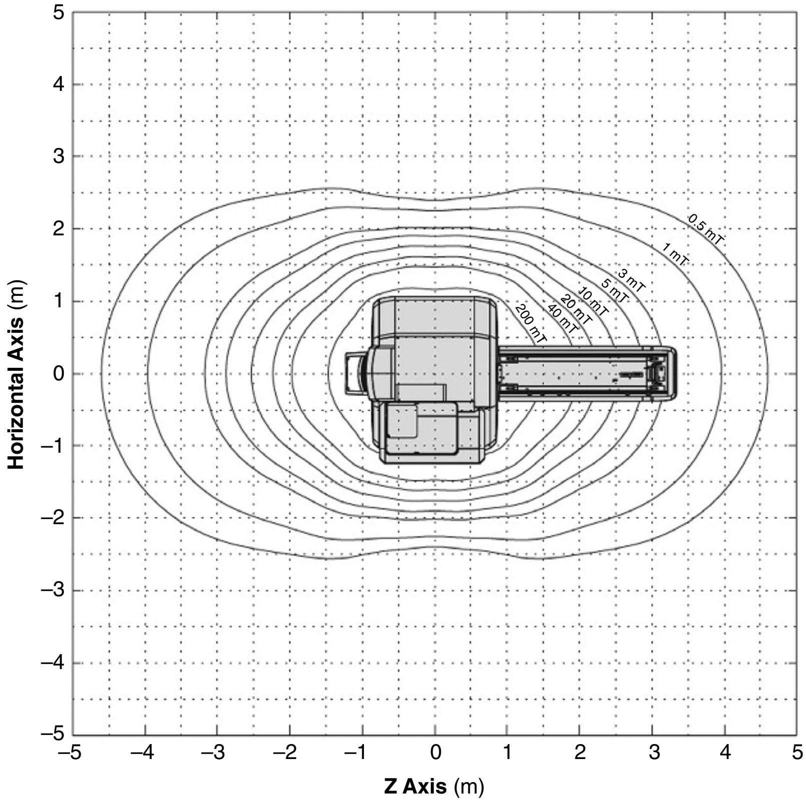

Figure 1.24 Fringe field contours at 0.5, 1, 3, 5, 10, 20, 40 and 200 mT for a 3 T MR magnet. Reproduced with permission of Siemens Healthineers.

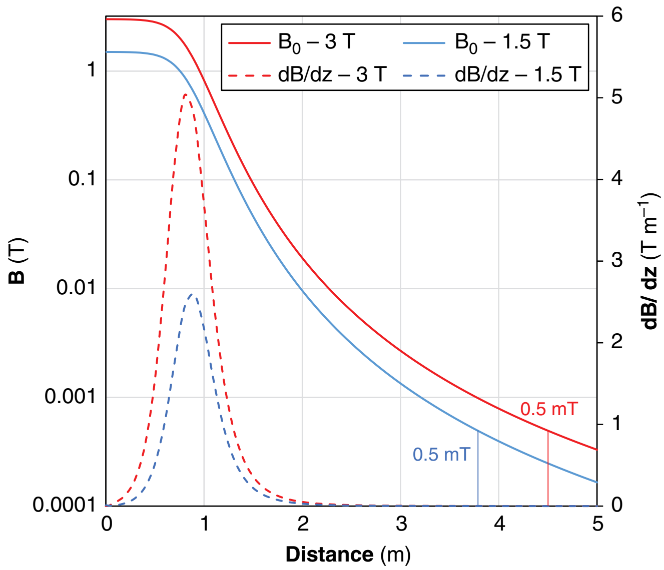

Figure 1.25 The magnitude of the B_0 fringe field (solid lines, logarithmic LH scale) and its spatial gradient dB/dz (dashed lines, linear RH scale) along the z-axis simulated for shielded 1.5 and 3 T MRI magnets. The vertical lines indicate the locations of the 0.5 mT contour. The iso-centre is located at z=0, and the bore entrance at 0.8 m.

Example 1.3 B_z from a gradient

In a 1.5 T MRI system with a gradient amplitude of 10 mT m^{-1} what is the total magnetic field at a point x = 10 cm from the isocentre?

$$B_z(x) = 1.5 + 0.1 \times 10 \times 10^{-3} = 1.501\,\text{T}$$

At a point x = –10 cm, the resultant B-field is 1.499 T.

The contribution to the overall magnetic field of the gradients is small, but we could not image without them. The strength of the field produced by the gradients decreases rapidly outside the bore of the magnet, and is negligibly small away from the magnet.

As the gradients are switched, they produce *time-varying magnetic fields*. The *rate of change* of field is given by the derivative of B with respect to time, or dB/dt (measured in T s^{-1}). For a trapezoidal gradient waveform (Figure 1.16)

$$\frac{dB}{dt} = \frac{\Delta B}{\Delta t} \tag{1.7}$$

where ΔB is the change in B produced by the gradient and Δt is the time over which the change occurs. dB/dt is important when considering acute physiological effects, such as peripheral nerve stimulation (PNS). See Chapter 4.

Example 1.4 Gradient dB/dt

In the example of Figure 1.16 if the peak gradient amplitude is 10 mT and the rise time 0.1 ms, what is the dB/dt?

$$\frac{dB}{dt} = \frac{10\,mT}{0.1\,ms} = 100\,\text{T}\,\text{s}^{-1}$$

Radiofrequency field

Figure 1.26 shows simulations of the electric and magnetic fields generated around an eight-rung birdcage transmit coil [5]. The magnetic B_1-field is highly uniform, whilst the electric field (E) is concentrated around the rungs. In air B_1 decreases rapidly beyond the limits of the transmit coil.[2] B_1 is produced as a pulse consisting of a "carrier" frequency (at the Larmor frequency) multiplied by a shape or envelope (Figure 1.27). The simple rectangular pulses of Equation 1.2 are seldom used in practice and a more general expression for flip angle is

$$\alpha = \gamma \int_0^{t_p} B_1(t)\,dt \tag{1.8}$$

This is equal to the area under the curve of the pulse envelope. Three important points arise:

1. for the same pulse shape and duration, the B_1 amplitude is proportional to the flip angle;
2. for the same pulse shape and duration, the B_1 amplitude required to produce a given flip angle is independent of B_0;
3. the peak amplitude of B_1 alone is not sufficient to characterize the RF exposure.

[2] This is not true in tissue. See Chapter 2, page 54.

MYTHBUSTER:

The amplitude of the B_1 RF excitation pulse does not depend upon the static field strength B_0.

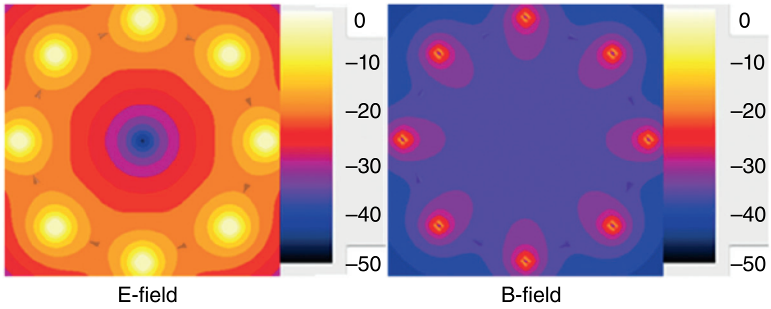

Figure 1.26 Simulated electric (L) and magnetic fields (R) from an eight-rung birdcage coil. Scale in dB. Source [5], licensee BioMed Central Ltd.

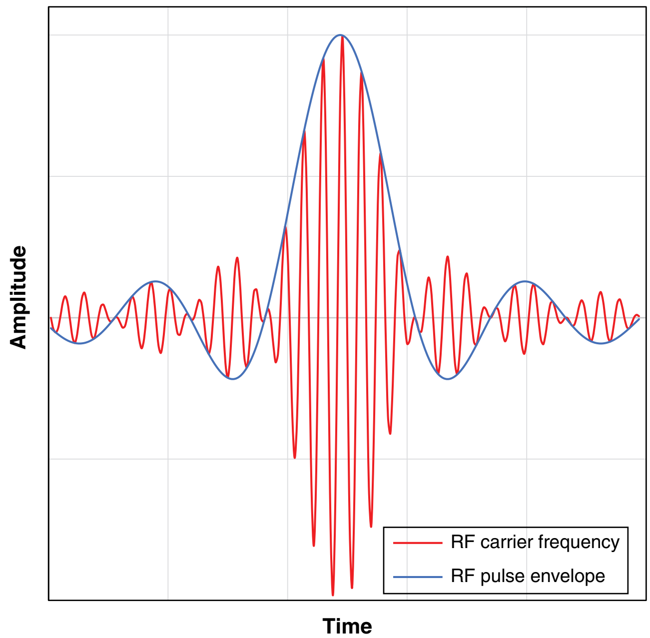

Figure 1.27 RF pulse consisting of the carrier (Larmor) frequency multiplied by a shape function or pulse envelope. The example shown is a truncated sinc (sinx/x) function.

B_{1+} and $B_{1+}RMS$

The parameter B_{1+}RMS is used to characterize the average B_1 exposure. The "+" refers to the rotating component of B_1 responsible for excitation of the magnetization. An efficient coil should not generate a B_{1-}. RMS stands for root-mean-square and is a type of averaging used for time-varying

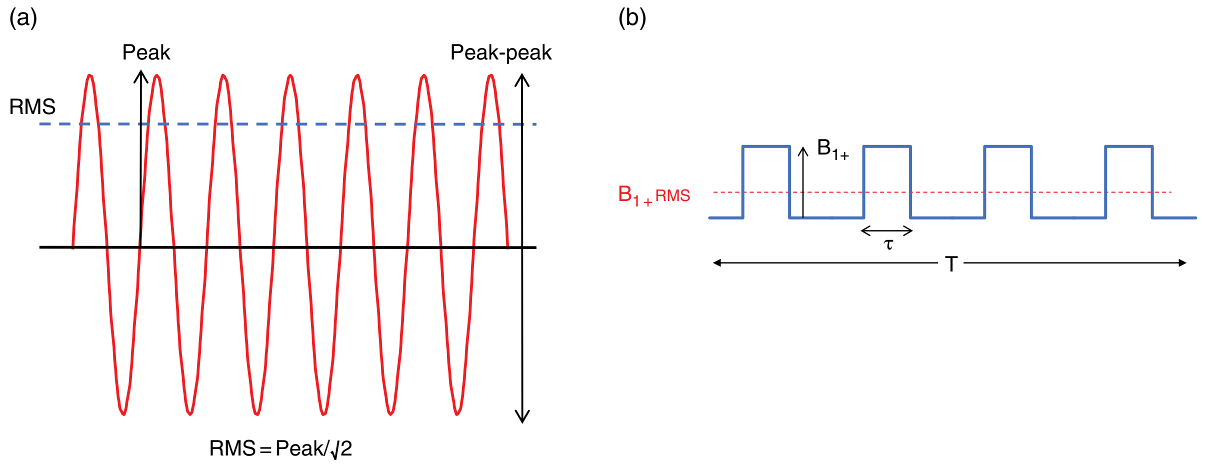

Figure 1.28 (a) the RMS value of a sinusoid is the peak amplitude divided by √2; (b) B_{1+}RMS for a train of N RF pulses of amplitude B_{1+}, duration τ within time T.

waveforms. For example, the RMS value for a sinusoidal waveform is 1/√2 or approximately 0.71 of the peak amplitude (Figure 1.28a). B_{1+}RMS is defined as

$$B_{1+RMS} = \sqrt{\frac{\int_0^T (B_{1+}(t))^2\, dt}{T}} \tag{1.9}$$

calculated over 10 second intervals (T=10 s). The easiest way to visualize this is to consider a regular train of N rectangular RF pulses (Figure 1.28b), each of amplitude B_{1+} and duration t_p. In this case

$$B_{1+RMS} = B_{1+}\sqrt{\frac{N\, t_p}{TR}} \tag{1.10}$$

Consequently B_{1+}RMS depends upon the

- flip angle
- number of RF pulses (echoes, slices, etc.)
- RF pulse shape
- TR.

B_1 is also a time-varying magnetic field. We can calculate the magnitude of dB/dt for a circulating field $B_{1+}e^{-i\omega t}$ as[3]

$$\left|\frac{dB_1}{dt}\right| = \left|\frac{d}{dt} B_{1+}e^{-i\omega t}\right| = \omega B_{1+} = 2\pi f B_{1+} \tag{1.11}$$

The rate of change is proportional both to the frequency and the amplitude. As B_{1+} is typically μT and f in MHz, RF dB/dt is of the order of a few tesla per second.

[3] Using complex notation where $i = \sqrt{-1}$, the operator $e^{i\omega t}$ signifies circular motion.

Scanning modes

The IEC standard 60601-2-33 [4] defines three modes for scanning:

- **Normal mode**: mode of operation of the MR equipment in which none of the outputs has a value that may cause physiological stress to patients.
- **First-level controlled mode**: mode of operation of the MR equipment in which one or more outputs reach a value that may cause physiological stress to patients which needs to be controlled by medical supervision.
  - Software allowing access to this mode must require specific acknowledgement by the operator that the first-level controlled mode has been entered.
- **Second-level controlled mode**: mode of operation of the MR equipment in which one or more outputs reach a value that may produce significant risk for patients, for which explicit ethical approval is required (i.e. a Human Studies protocol approved to local requirements).

"Outputs" refers to the magnitude of the magnetic fields. Clinical scanners are usually restricted to the Normal and First Level Modes.

OTHER MEDICAL DEVICES

Along with an understanding of MRI hardware and fields it is important to understand how these interact with other medical devices. A system to categorize the MRI safety (we used to say "compatibility") of other devices: implants, accessories, medical equipment, tools, fire extinguishers, gas tanks, etc. uses three labels [6]:

- ***MR Safe*** means that the device poses no risk to the patient in the MR environment. Image quality may be affected.
- ***MR Conditional*** means that the device poses no additional risk to the patient when introduced to the MR environment under specified conditions.
- ***MR Unsafe*** means that the device may not be introduced into the MR environment as it poses significant risk to the patient and/or staff.

The approved signs are shown in Figure 1.29. The "MR environment" generally means the MR examination room, or areas with a fringe field exceeding 0.5 mT, rather than just the scanner itself. The safety of implants is considered in Chapters 9–11.

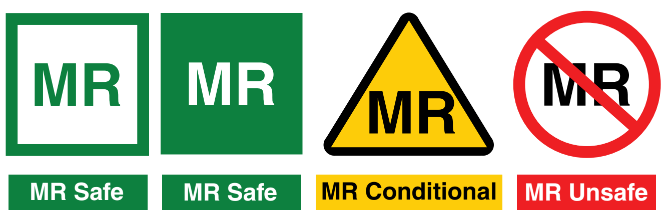

Figure 1.29 MR device labeling according to ASTM-F2503 (IEC-62570) [6]. The two MR Safe symbols are equivalent; either may be used.

CONCLUSIONS

MRI incidents can lead to injury or death. The most frequent incidents are thermal, followed by mechanical, projectile, and hearing loss. We have considered the basic elements of MRI acquisitions, the components of the scanner and the magnetic fields encountered. There is a symmetry about the magnitude and time-variance of the fields: B_0 is of the order of tesla; the imaging gradient fields a thousand times lower, typically milli-tesla; B_1 is typically one thousand times less again, in micro-tesla. At the same time the temporal variations range from zero to one hertz for movement in the static field, to kHz for the gradients, and MHz for B_1.

In the next chapter we consider the physical interactions of these fields with non-biological matter. For those wishing to become MR Safety Experts, Chapter 2 should be read in conjunction with Appendix I supplemented by reading further reading of standard electromagnetism texts.

Revision Questions

1. The Larmor frequency of a 1.5 T MRI scanner is approximately:
   A. 10 MHz
   B. 42.58 MHz
   C. 64 MHz
   D. 85 MHz
   E. 128 MHz
2. In a 1.5 T MRI scanner, if the B_{1+} amplitude required to produce a 90° flip angle is 10 μT, what B_{1+} amplitude is required in a 3T scanner if the pulse shape is unchanged?
   A. 5 μT
   B. 10 μT
   C. 20 μT
   D. 40 μT
   E. 0.01 mT
3. Which of the following is true for B_1?
   A. It is applied along the z-direction along the magnet bore
   B. It is a single sinusoid
   C. It is a radio wave
   D. It is generated by the x and y gradient coils
   E. It rotates with the magnetization precession.
4. Which of the following is untrue for the static field spatial gradient in a superconducting MRI system?
   A. It is measured in tesla per meter
   B. It is required for image acquisition
   C. It is always present
   D. It is responsible for the translational magnetic force
   E. It is reduced in extent in a self-shielded magnet.
5. If an imaging gradient system has a peak amplitude of 50 mT m^{-1} and a slew rate of 200 T m^{-1} s^{-1} what is the minimum achievable rise time for a full amplitude pulse?
   A. 10 μs
   B. 0.1 ms
   C. 0.2 ms
   D. 0.25 ms
   E. 0.4 ms

6. Which of the following is not acceptable terminology for MR safety according to ASTM-F2503?
   A. MR safe
   B. MR unsafe
   C. MR compatible
   D. MR conditional
   E. MR acceptable.

References

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2. Kanal, E., Borgstede, J.P., Barkovich, A.J. et al. (2002). American College of Radiology white paper on MR safety. *American Journal of Roentenology* 178:1335–1347.
3. Delfino, J.G., Krainak D.M., Flesher S.A. et al. (2019). MRI-related FDA adverse event reports: a 10-year review. *Medical Physics* doi: 10.1002/mp. 13768.
4. International Electrotechnical Commission (2015). *Medical Electrical Equipment – Part 2-33: Particular Requirements for the Safety of Magnetic Resonance Equipment for Medical Diagnosis. IEC 60601-2-33 3.3 edn*. Geneva: IEC.
5. Lui, Y., Shen, J., Kainz, W. et al. (2013). Numerical investigations of MRI RF field induced heating for external fixation devices. *BioMedical Engineering OnLine* 12:12 doi.org/10.1186/1475-925X-12-12.
6. ASTM F2503-13 (2015). *Standard Practice for Marking Medical Devices and Other Items for Safety in the Magnetic Resonance Environment*. West Conshohocken, PA: ASTM International.

Further reading and resources

McRobbie, D.W., Moore, E.A., Graves, M.J., et al. (2017). *MRI from Picture to Proton* 3rd edn. Chapters 3, 4, 8, 9, and 10. Cambridge, UK: Cambridge University Press.