Chapter 10

Let's Talk Technical: MR Equipment

10.1 Introduction

At the start of this book we said that you don't need to understand the workings of the internal combustion engine to be able to drive a car. However, if you're curious, this chapter provides an opportunity to get down and dirty with the innards of the equipment. Most people will not need to build or service their own system, but you are likely to be involved in the exciting decision of which scanner to purchase. During this process you need to understand the technical specifications so that's what we will focus on. The basic components of an MRI scanner were introduced in Chapter 2. In this chapter we provide more technical information. A lot of the engineering detail is in the advanced boxes, leaving the most important information in the main text. You will see that:

- there are three types of magnets available for clinical MRI, capable of a range of field strengths, and each magnet type has its own advantages and disadvantages;
- magnetic field gradients are generated by a set of gradient coils and amplifiers, which are characterized by their maximum strength and slew rate, and often use active shielding to minimize eddy current artefacts;
- the radiofrequency (RF) transmit system is designed to produce a uniform excitation within the patient, using a large volume coil which may be embedded in the gradient set, and a powerful RF amplifier;
- to receive the MR signals, dedicated receive coils are used that fit closely to the patient's body, in order to maximize sensitivity and signal-to-noise ratio;
- several parts of the system require cooling systems, which may be managed with forced air or chilled water;
- the MR signal is digitized as soon as possible to minimize noise contamination in the cabling, and several computers within the MRI system are needed to synchronize scanning operations.

Figure 10.1 shows the basic architecture of a typical MR system.

10.2 Magnets

The magnet is the most expensive component of the MR system. All magnets produce a static magnetic field which is non-uniform and the homogeneity over the imaging volume needs to be optimized by a process known as shimming, whereby pieces of steel and/or electrical coils are incorporated into the system. Static shimming is performed at system installation; however, putting the patient into the system also introduces a relatively large inhomogeneity, and all systems can perform additional shimming on a per-patient or even per-scan basis (this is known as a pre-scan). A magnet will also generate a magnetic field outside of the patient aperture. Although the extent of this 'fringe field' is minimized by design, it can be necessary to use additional magnetic field shielding either for safety reasons or to avoid interference with nearby sensitive electronics. Magnets have three important properties which should be considered when purchasing a system: field strength, homogeneity over the imaging volume, and installation footprint.

10.2.1 Field Strength

Magnetic field strength (or flux density) is measured in tesla (T). The majority of clinical MR systems operate in the range of 1.5–3 T. Lower field strengths offer cost benefits, and there are several vendors offering low-cost systems in the range 0.2–0.6 T. The advantages of higher field strengths are a better SNR (see Chapter 11) and increased chemical shift, which can improve

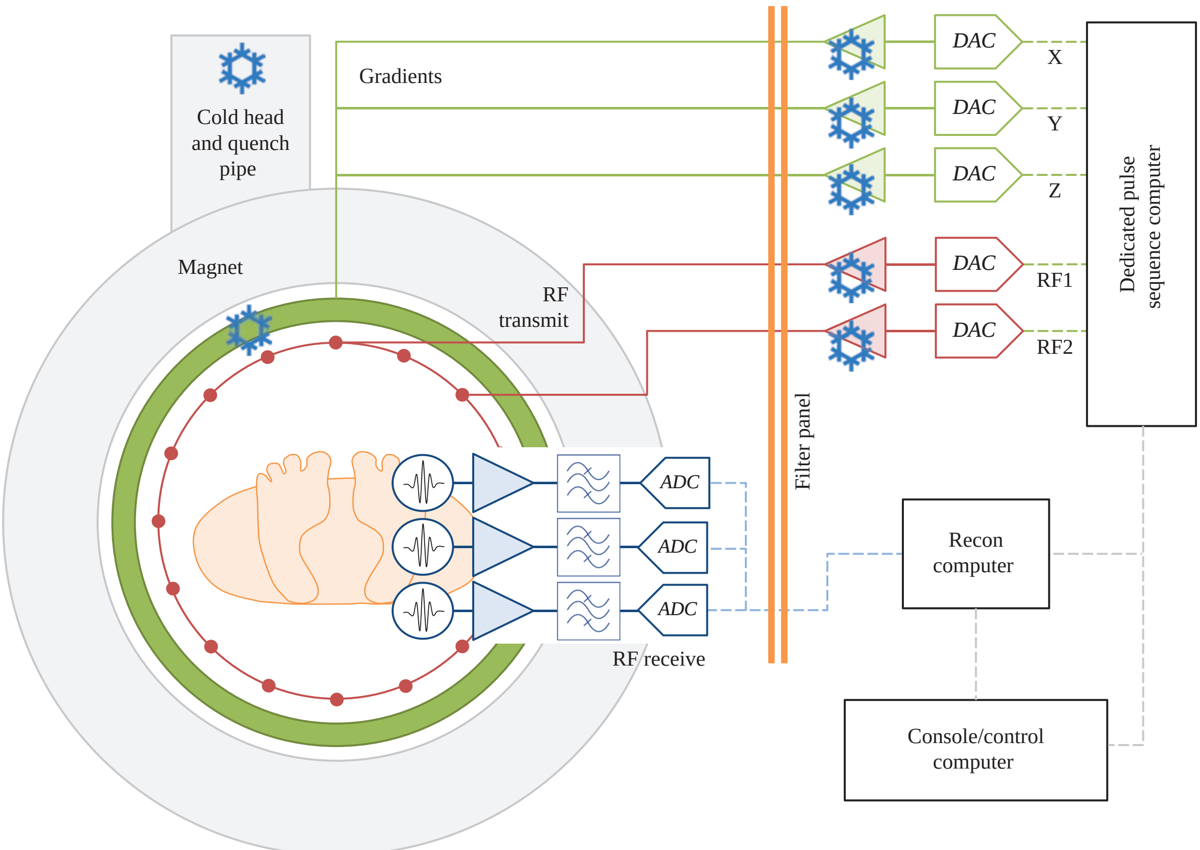

Figure 10.1 Basic components and architecture of an MR system. Solid lines indicate analogue signals, dotted lines are digital. The snowflake symbol indicates components which require specialist cooling (either by chilled water or forced air cooling).

spectral fat suppression and spectroscopy. The improvement in SNR with increasing field strength may be traded for increased spatial resolution, or decreased imaging time. The field strength available is strongly dependent on the type of magnet: superconducting, electromagnets, or permanent.

Superconducting magnets are by far the most common, having the typical 'tunnel' shape and allowing field strengths of 1.5–3 T. These magnets are made from special metal alloys, which have zero electrical resistance at temperatures close to absolute zero (–273.16 °C, 0 K). The magnet construction includes a large volume of liquid helium (4 K) to keep the wires at their superconducting temperature. Once an electric current is running in a loop of superconducting wire, it will continue to circulate indefinitely without resistive losses, provided the coils remain below their critical temperature. As well as the highest field strengths, superconducting magnets also offer the best homogeneity and stability over time.

Super Cool Magnets

Since you are most likely to meet a superconducting magnet in your local MRI unit, let's look at some of the design details. The superconducting wire is a niobium–titanium (NbTi) alloy, and is made up of many NbTi filaments embedded in a copper matrix, coated with an insulator. The copper matrix makes it easier to handle the delicate superconducting windings, and protects the superconducting windings against a quench. The NbTi filaments have a superconducting transition temperature of 7.7 K and become superconducting when immersed in liquid helium (He), which has a boiling point of 4.2 K. A typical superconducting magnet consists of a cryostat, i.e. a large chamber of liquid helium, in which the magnet coils are immersed, surrounded by a cold shield and a vacuum chamber (Figure 10.3a).

The helium continuously evaporates, and the helium gas is cycled through a refrigerating system to keep it below 20 K. So-called 'zero boil off'

(a) (b) (c)

Figure 10.2 Commercial examples of different magnet types in MR systems. (**a**) 1.5 T Philips Ingenia, superconducting with horizontal main field B_0; (**b**) 0.6 T Fonar upright positional system, electromagnet with horizontal B_0 (courtesy of Medserena Upright MRI Centre, London); (**c**) 0.23 T Siemens Magnetom C, permanent magnet with vertical B_0.

magnets also use cryogenic coolers to convert the gas back to liquid helium. A heater within the helium vessel is used to maintain a certain pressure in the gas space above the liquid surface – this limits the amount of evaporation. When heat is generated, e.g. by the gradients, the pressure increases and helps to limit the amount of evaporation. A pressure control circuit is used to maintain the pressure level within limits known as the 'thermal margin', and to allow escape of He gas in case of malfunction. The cryocooler or 'cold head' uses controlled gas expansion to keep the heat shield at 40 K within the magnet cryostat.

An external power supply is used when the magnet is first energized or 'ramped' up to the required field strength. A superconducting switch is used to short-circuit the magnet once the desired magnetic field has been established. Superconducting magnets have evolved significantly since MRI was introduced. The first generation of systems had two cryogenic baths, the outer one containing liquid nitrogen (Figure 10.3b) which, although very cheap, made the system cumbersome to refill. The second generation (Figure 10.3c), from the early 1990s until the early 2000s, used more efficient cold-head refrigeration and allowed for only the helium cryostat. These systems still consumed a lot of helium, which is expensive. The current state of the art is known as 'zero boil off' (Figure 10.3d), which actually means extremely low boil off: a well-maintained system will not need a refill during its lifetime. Systems in the future may use cryogen-free magnets (Figure 10.3e), which contain helium gas at around 40 K instead of liquid. Those magnets rely on extremely efficient cold-head units and have the disadvantage that if the cold-head malfunctions, the pressure and temperature can quickly rise until the magnet windings go 'normal'. However, there are also significant advantages, the obvious one being that there is no need for thousands of litres of expensive liquid helium, and there is no risk of a catastrophic quench.

Quench

If any part of the superconducting windings gets too hot, above the critical temperature of around 7 K, the wire becomes resistive and the stored electrical energy will be dissipated as heat. This rapidly warms up neighbouring parts of the windings, which will in turn dissipate more heat and propagate the effect throughout the magnet. This is known as a 'quench' and results in the collapse of the magnetic field together with very rapid boiling off of the helium. The MR cryostat typically holds around 2000 litres of liquid helium, which expands 750 times when it becomes a gas. This very large volume of He gas needs to escape from the cryostat, so magnet designers incorporate bursting-disks that blow out under high pressure. Exhaust or quench pipes then vent the gas outside the imaging room to prevent asphyxia and cold burns. Oxygen-level monitors are sometimes installed in the magnet room to alert the users to a dangerous depletion of oxygen should any He gas leak from the quench pipe. Spontaneous

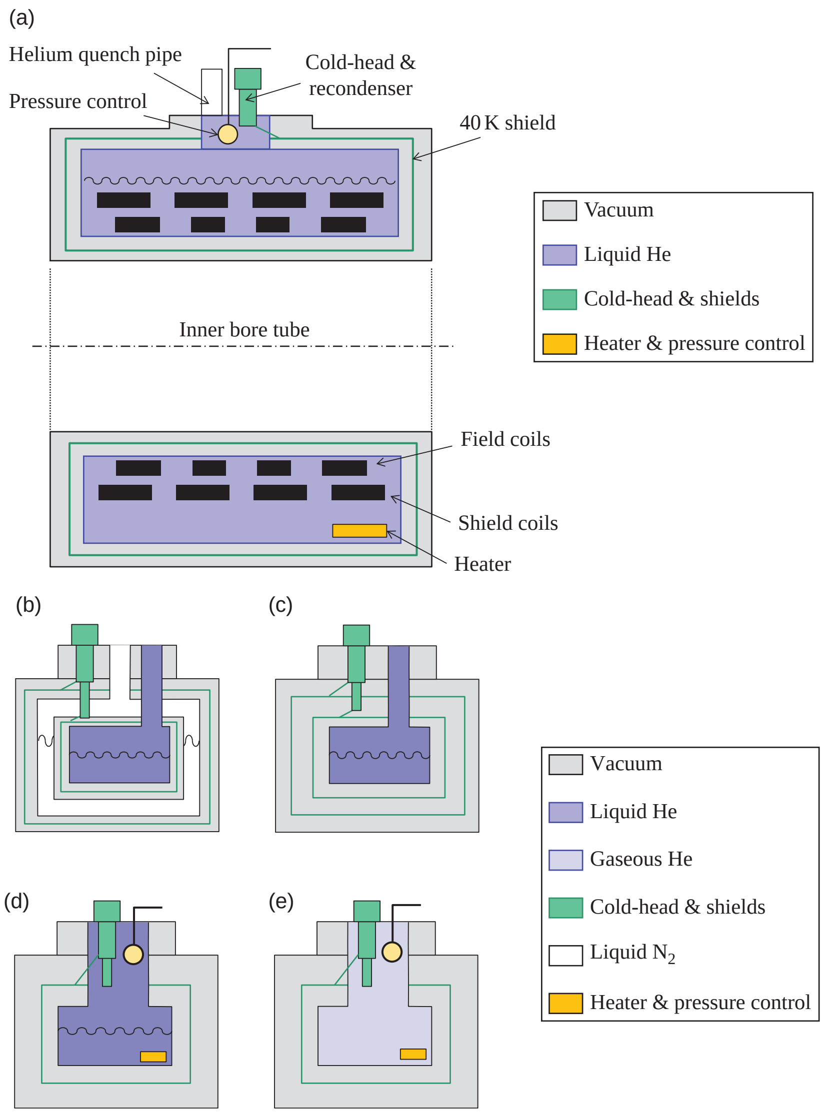

Figure 10.3 (a) Cross-section through a ‘zero boil off’ superconducting magnet. (b–e) Four generations of superconducting magnet designs (see Box ‘Super cool magnets’ for details). Adapted with permission from an original by Paul Harvey, Philips Healthcare.

quenches are rare occurrences, but there may be occasions when it is necessary to ‘ramp down’ the magnet through a controlled quench. In this situation the electrical energy is deposited in a dummy load to avoid damaging the magnet. In either case re-energizing the magnet can be a costly procedure.

Electromagnets use regular resistive coils wound around iron pole pieces. When an electric current flows through the coils the iron becomes a magnet. Magnetic fields of around 0.6 T can be achieved, putting these systems in the low-field range. MR systems based on electromagnets tend to be heavier than superconducting systems, but have lower cost since they don’t need the special wire or the expensive helium necessary to maintain the low temperature.

Permanent magnets can support fields up to 0.3 T and offer the advantage of very low running costs, since they need neither helium nor electrical power to maintain the field. They are heavier than other magnets and may require extra floor strength. The field is often orientated vertically, and it should be remembered that these systems can never be ‘switched off’.

10.2.2 Homogeneity

The homogeneity of a magnetic field is usually expressed in parts per million within a given spherical volume. The size of this volume is often given as the **D**iameter of a **S**pherical **V**olume (DSV). For example, a 1.5 T magnet with a maximum variation of 7.5 μT (0.0000075 T) over a 40 cm DSV has an inhomogeneity given by

$$\Delta B_0\ (\text{ppm}) = \frac{\Delta B_0}{B_0} = \left(\frac{0.0000075}{1.5}\right) \times 10^6 = 5\ \text{ppm}$$

Modern superconducting scanners have excellent homogeneity, typically much less than 1 ppm over a 40 cm DSV. There is not much point trying to improve this in the fixed shim, since the human body causes inhomogeneity of 1–5 ppm, depending on which part of the body is in the centre of the system. Manufacturers' technical sheets often quote inhomogeneity over a range of different DSVs. Care is required in their interpretation, as these figures may represent only the mean or average inhomogeneity, typically called the **R**oot **M**ean **S**quare (RMS) value, and not the maximum (peak-to-peak) value, which may be considerably more.

Fixed shimming, performed at installation, improves the magnet homogeneity and corrects for any distortions caused by any nearby ferromagnetic structures. Shimming can be done passively or actively. Passive shimming involves adding small iron plates into special rails in the magnet bore, while active shimming is done with up to 18 in-built superconducting coils in addition to the main magnet coils.

10.2.3 B_0 Mapping

Mapping of the static magnetic field (B_0) is important for certain shimming applications, known as image-based shimming, and can be used in EPI imaging for removing distortion. The simplest method of spatially mapping the native system homogeneity is to use a phase difference technique in a large, ideally spherical, non-conducting uniform phantom. Two images are acquired with different echo times (TE), most efficiently as a dual-echo gradient-echo scan. Due to the B_0 non-uniformity there will be a spatial variation in the Larmor frequency. The phase accumulation between the two different TEs is proportional to the local frequency (see Figure 10.4). Keeping the images in complex form, the phase difference can be calculated from

$$\Delta\phi(x,y) = \angle\left[S_1(x,y) \cdot S_2^*(x,y)\right]$$

where S_1 and S_2 are the images at TE_1 and TE_2 respectively, the superscript * denotes the complex conjugate, and $\angle$ represents taking the angle of the complex data. The difference in echo times $\Delta\text{TE} = \text{TE}_2 - \text{TE}_1$, and so we can estimate the local resonant frequency $\Delta\omega$, meaning the frequency difference from the nominal Larmor frequency, as follows:

$$\Delta\omega(x,y) = \frac{\Delta\phi(x,y)}{\Delta\text{TE}}$$

giving $\Delta\omega(x,y)$ in radians/second. Of course, this phase difference can exceed $\pm\pi$, in which case it is necessary to apply some form of phase unwrapping algorithm to the phase difference image. To minimize this, the ΔTE should be kept short.

Practical Considerations: Phase Images

Although most scanners allow you to generate phase images, they are not always directly usable for calculating a phase difference. If you have difficulty, it is generally easier to calculate the phase from the real (R) and imaginary (I) images, using the relationship

$$\phi_n(x,y) = \tan^{-1}\left(\frac{I_n(x,y)}{R_n(x,y)}\right)$$

Care should also be taken when using phased array receive coils for B_0 mapping, as the complex data may lose phase information during the coil combination reconstruction.

10.2.4 Installation Footprint

The amount of space needed to site the system has a direct influence on the cost of an installation. The magnet room is the largest of the spaces, and most hospitals choose to make it large enough to completely contain the magnetic fringe field, to the 0.5 mT (5 G) line. It is important to realize that the fringe field extends in all three directions, so ceiling height is also a consideration, as is the space under the floor of the magnet room.

Magnetic field shielding reduces the extent of the fringe field. Most modern magnets are actively shielded. This involves continuing the superconducting coil windings, carrying current in the opposite direction and positioned outside the inner main

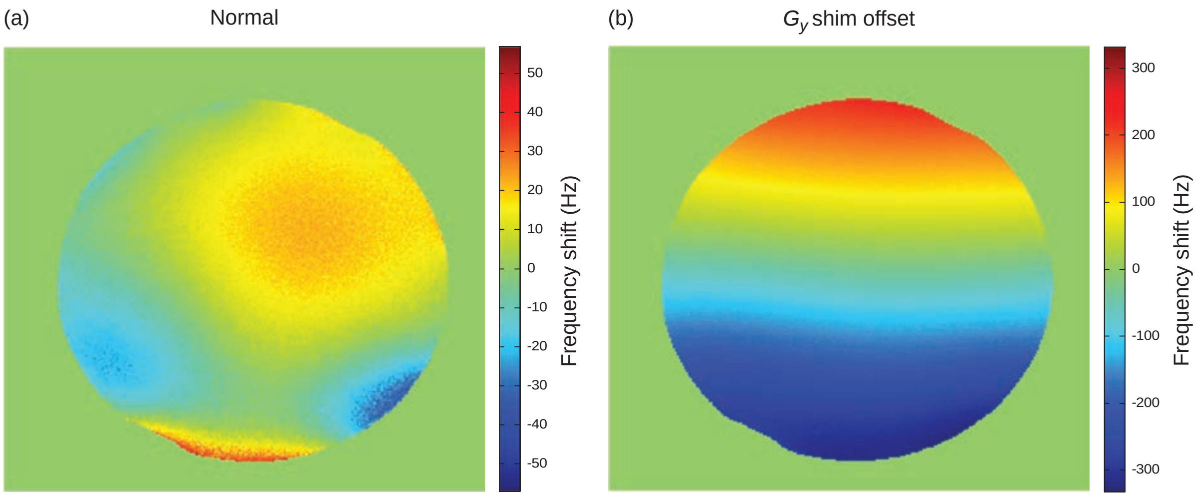

Figure 10.4 B_0 map in a 3 T system (**a**) using the default shim values and (**b**) with a large offset deliberately imposed on the *y* shim (up–down).

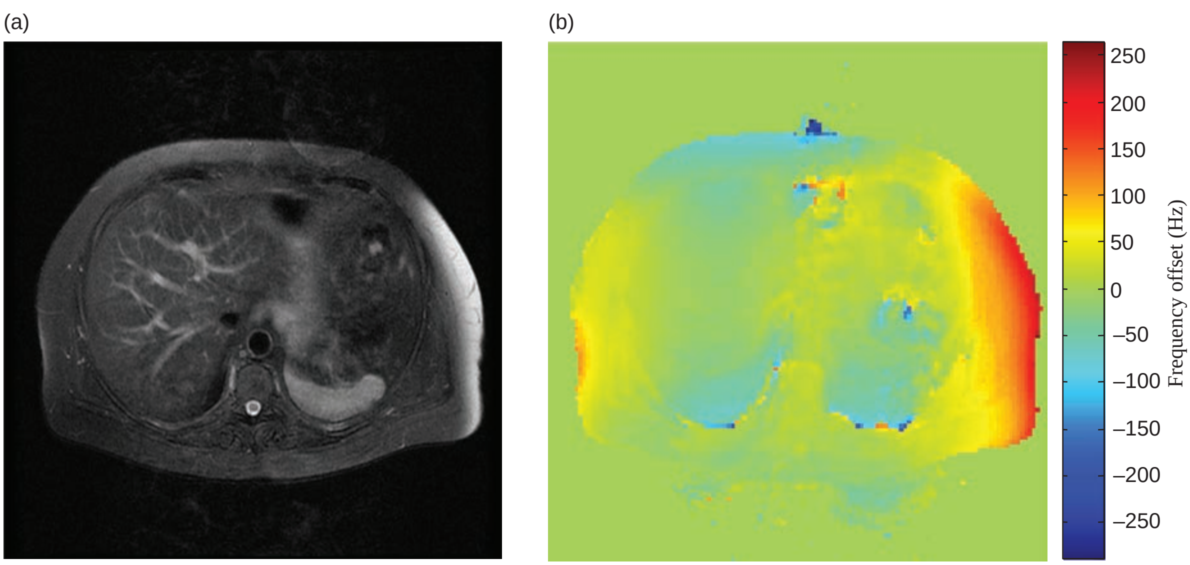

Figure 10.5 (**a**) T_2-weighted FSE scan with poor fat suppression on the right. (**b**) The B_0 map at the same slice location shows a large frequency shift in this region explaining the failure of the fat suppression. This was the result of a poor shimming prescan.

magnet winding. This partially cancels the field outside the main magnet coils, thereby reducing the stray field strength. For all magnet types, shielding can also be 'passive', achieved by positioning iron or high-permeability steel plates close to the magnet. Active shielding makes the magnet more expensive; however, passive shielding can also be expensive, depending on how much steel is needed, and may require additional architectural reinforcement.

10.3 Gradient Subsystem

As we saw in Chapter 8, spatial localization of the MR signal requires the use of three orthogonal linear magnetic field gradients ('gradients' for short). The gradient subsystem comprises the set of gradient coils, inside the bore of the magnet, and the gradient amplifiers which drive electrical current through them. In the standard cylindrical clinical systems,

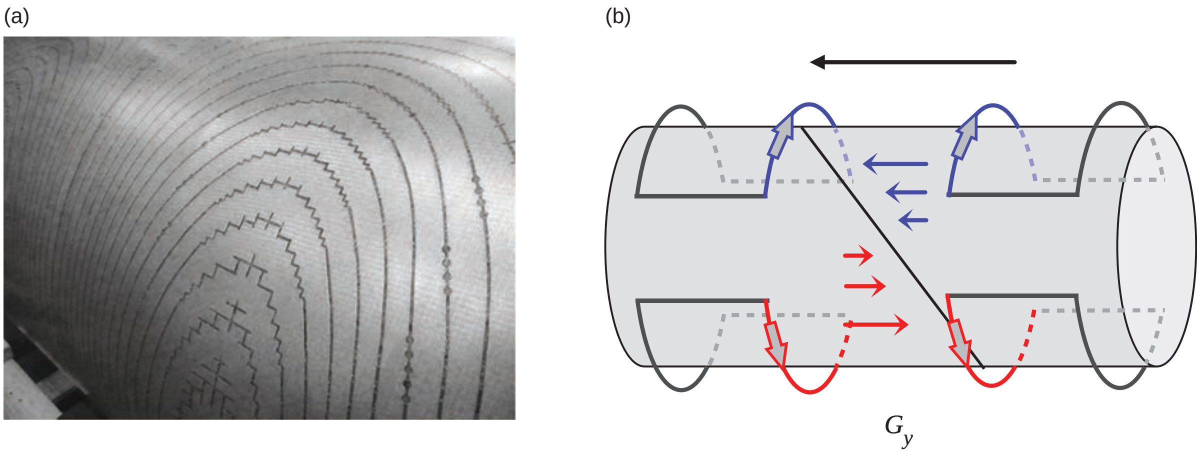

Figure 10.6 (a) Commercial gradient coil set, showing the outermost copper shield 'coil'. (b) Schematic diagram of current path to generate the *y* gradient.

the direction along the bore is termed the *z* axis, the left–right direction is termed the *x* axis and the top–bottom direction is termed the *y* axis. The null point at the centre of the gradient coils, and also the centre of the magnet, is called the isocentre.

Each gradient coil is designed to produce a field which points along the *z* axis, with its magnitude changing along the gradient direction. Mathematically, this is written as e.g. $\partial B_z/\partial x$ for the *x* gradient. The 'coils' don't look much like conventional solenoids though: they are more likely to be a sheet-like copper panel with cut-outs to create the current paths (Figure 10.6). This is known as distributed winding, or fingerprint winding, which optimizes the shielding and allows for the inductance to be minimized. If you follow the path of the current, it forms two opposing loops (Figure 10.6) which is essential to produce the desired magnetic field gradients. The coils heat up rapidly due to the large currents, and almost all commercial MR systems have a cooling system, tubes carrying chilled water through the gradient coils and amplifiers.

When considering the gradient subsystem, there are some key properties which affect performance. These include the maximum gradient strength and slew rate, the duty cycle, gradient linearity, and the capacity of the cooling system.

10.3.1 Gradient Strength and Slew Rate

Gradient pulses in conventional pulse sequences are trapezoidal in shape, with a sloping rise, followed by a flat plateau and a sloping fall (Figure 10.7). The

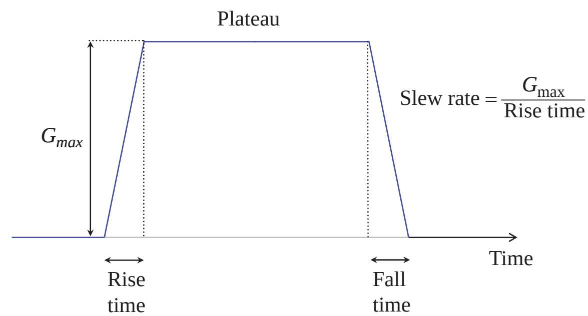

Figure 10.7 Trapezoidal gradient pulse showing definitions for G_{max} and slew rate.

strength of the gradient, or how rapidly the field changes over distance, is expressed in milli-tesla per metre (mT m^{-1}), in the range of 1–50 mT m^{-1}, depending on the purpose of the gradient pulse. If the gradient set allows very high gradient amplitudes, images can be acquired with smaller fields of view and/or thinner slice thicknesses, while maintaining short echo times or echo spacings. Large peak gradient amplitudes are also required for some specialist scanning techniques, e.g. diffusion imaging. So the maximum gradient possible, G_{max}, is an important property of the MR system.

The gradient rise time, or how rapidly the field changes with time from zero to the peak amplitude, is usually expressed in microseconds (μs), with typical values from 1000 μs down to 200 μs. The gradient slew rate is calculated by dividing the peak gradient amplitude by the rise time. Typical slew rates are in the range 20–200 T m^{-1} s^{-1}. High slew rates are

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necessary for acquiring high-quality EPI images or for very short TE and TR, e.g. for cardiac imaging, and to achieve short echo spacing in TSE. Systems with very high slew rates may require software limits to prevent peripheral nerve stimulation (see Chapter 20).

10.3.2 Gradient Duty Cycle

There are at least three definitions of duty cycle in common use. It is not easy to find out how manufacturers define their measurement of gradient duty cycle, which obviously makes it difficult to compare systems. In addition, when you look at manufacturers' data sheets, you will find that almost all of them show a gradient duty cycle of 100%. So it's a rather useless property when it comes to making a purchase decision.

Reporting for Duty: Ways to Measure Gradient Duty Cycle

The first definition is quite easy to describe: a gradient which oscillates continuously from maximum positive to maximum negative strength at the maximum slew rate has a duty cycle of 100%. Since both the gradient coils and the gradient amplifiers experience large temperature changes, the duty cycle may be limited in order to control thermal behaviour. This might reduce the slew rate, or the maximum strength, but in either case, the duty cycle would be reduced.

The second definition of duty cycle is the percentage of time for which the gradients can be run at maximum strength (without switching). Since this is a high-current situation, it is primarily limited by the temperature control of the system. Note that without a definition of the time for the measurement, the duty cycle figure is rather meaningless.

The last definition of duty cycle is also known as the RMS power, and is calculated using the formula:

$$\text{Duty cycle} = \frac{\int_{t1}^{t2} G^2 dt}{G_{max}^2} \times 100\%$$

where G is a function of time, and G_{max} is the maximum gradient reached between times t_1 and t_2. When the time-integral is over the TR period, the duty cycle reaches a steady state. However, this definition of duty cycle is obviously dependent on the pulse sequence for measuring, as well as the gradient subsystem itself. So in order to compare different systems, you need to make sure they all use the same pulse sequence.

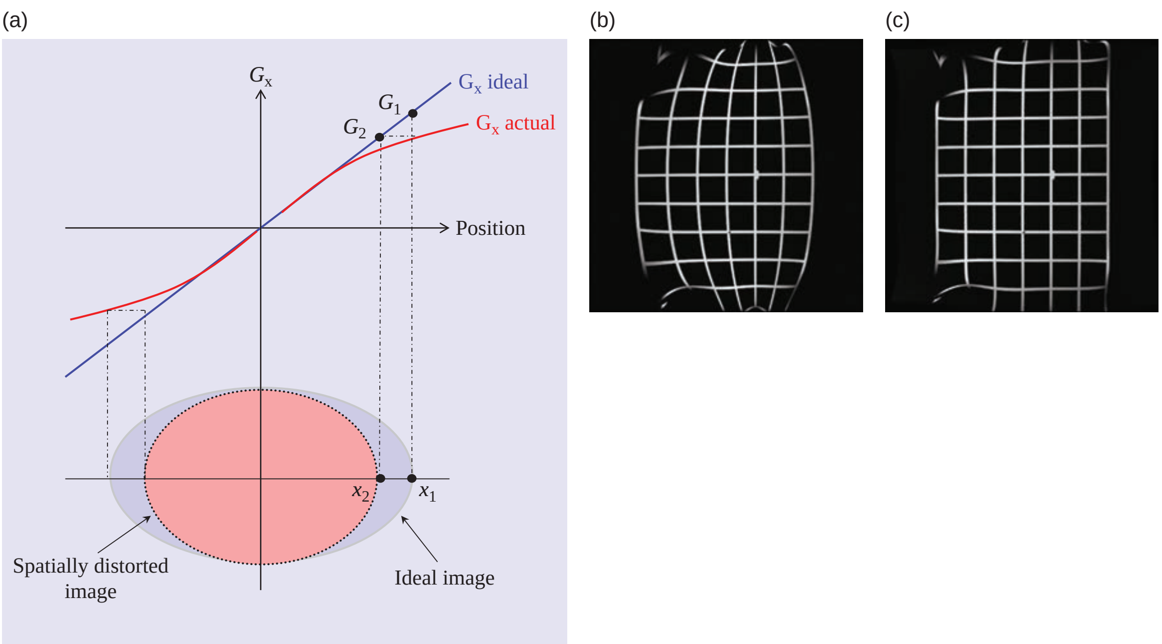

Figure 10.8 (a) Gradient field non-linearity; (b) a grid phantom scanned without correction for non-linearities; and (c) the same phantom with correction applied.

10.3.3 Gradient Linearity

Linear variation in the gradient field is required for accurate spatial encoding; however, the gradient coils have a finite length and diameter and so they produce non-linear fields close to their edges. Linearity usually decreases fairly rapidly towards the edge of the imaging volume. The consequence of non-linearity on an image is misplaced signal and geometric distortion (Figure 10.8a). Most manufacturers make use of computer algorithms that warp the images after reconstruction to compensate for the gradient non-linearities (Figure 10.8b).

Meet Eddy, the Current Guy

Whenever a changing voltage (or current) is experienced in a conductor, according to Lenz's law, a *back-emf* is generated (*emf* = electro-motive force, a nineteenth-century term for voltage). The back-emf is caused by eddy currents, which are localized currents running in the conductor. Eddy currents, in their turn, are induced by the changing magnetic field generated by the changing forward current, and they can be induced in nearby conductors too. Eddy currents generally cause energy losses in the form of heat, and since they are also changing with time, they generate time-varying magnetic fields of their own. The back-emf always acts to oppose the changing current, i.e. it slows down the rate of switching, meaning that you don't get the gradient you asked for, at least not exactly at the right time. So, all the rapid switching of the gradients causes an avalanche of electromagnetic effects in the gradients themselves and in the surrounding magnet, which can result in image artefacts and signal loss. There are several engineering ways to reduce these problems: active shielded gradient coils, pre-emphasis on the driving waveforms, and careful design of the gradient to minimize its inductance.

All modern MR systems use active shielding on the gradients, very similar technology to the active shielding of the magnet coil itself. Additional 'secondary' or 'shield' coils surround the primary gradient coils and are driven with the opposite gradient waveform. This aims to cancel the gradient field outside the shield coils, magnetically isolating the gradients from the cryostat so that eddy currents cannot be induced. The shield coils make the entire gradient assembly larger, reducing the free space available inside the bore, and also require more power to generate a given gradient amplitude.

Although active-shielded gradients induce fewer eddy currents, they do not prevent the back-emf in the coil. This effect can be minimized by *pre-emphasizing* the gradient waveforms so that, when combined with the eddy current field, the resultant is close to the ideal gradient waveform (Figure 10.9). Pre-emphasis uses extra electronic circuits that add additional voltages, with adjustable amplitudes and

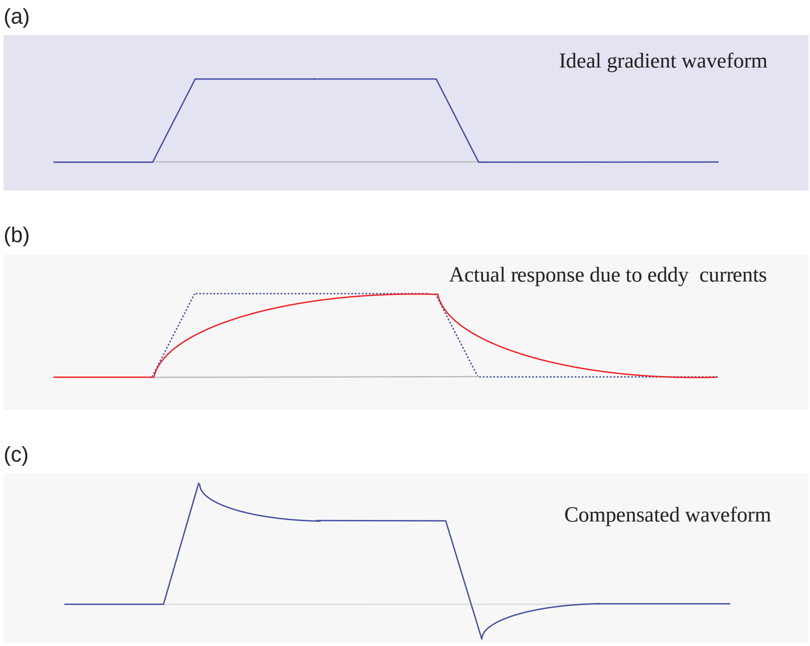

Figure 10.9 Gradient pulses. (**a**) Ideal gradient waveform; (**b**) actual response due to eddy currents; and (**c**) compensated waveform.

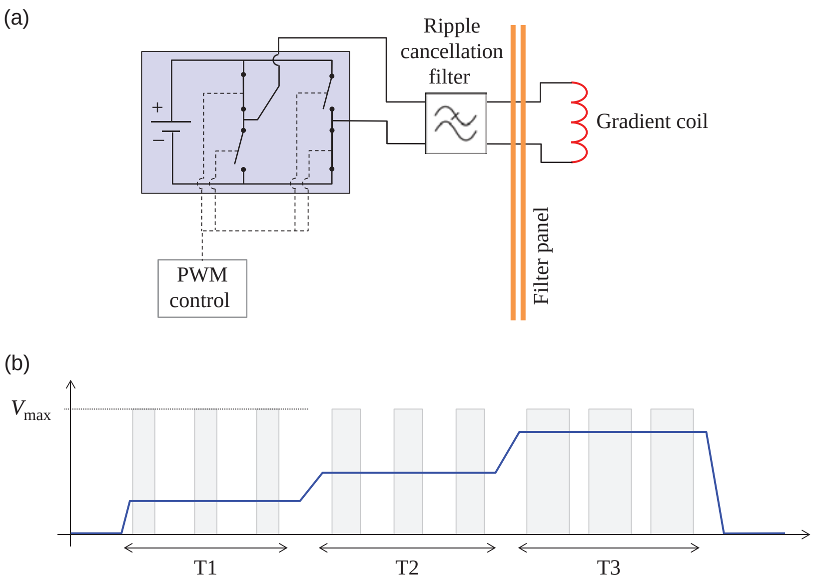

Figure 10.10 (a) Simplified diagram of a PWM amplifier. When the switches are closed, the full output power of the amplifier is driven through the gradient coil. The switches are implemented using transistors and controlled digitally by the PWM circuit. (b) Example of PWM waveform (grey line) switching on and off the amplifier voltage at different frequencies. In period T1 the on:off ratio is approximately 25%, giving an average output voltage of 0.25 × V_{max}. In period T2 the ratio is 50%, while in T3 it is 80%. The output voltage waveform is shown in blue.

time constants, to the gradient driving waveform. Using electromagnetic simulations, pre-emphasis can also help to compensate for concomitant gradient fields, i.e. field variations induced on the other axes.

Finally, the back-emf depends on the rate of change of the current, and on the inductance L of the gradient coil:

$$\text{emf} = L\frac{dI}{dt}$$

So to minimize eddy currents and reduce the back-emf, gradient coils are always designed to have minimal inductance. Specially designed software is used to calculate the optimal path for the primary and secondary gradient windings, so that L is minimized and the fields outside of the gradient assembly are close to zero, while maintaining the required current density to generate the gradient fields within the magnet field of view.

10.3.4 Gradient Amplifiers

Gradient amplifiers operate at audio frequencies and could also be used to hold your very own rock concert – one of the authors has been known to connect his guitar to the MR system, using the gradient coils as a loudspeaker! (Don't try this yourself, it's dangerous: he had help from professional service engineers.) The requirement for high gradient amplitudes means that the amplifier must be capable of producing large electrical currents through the coils. Furthermore, the requirement for short switching times means that this current must be rapidly increased from zero to the maximum and then back down. The amplifier therefore needs to generate a sufficient driving voltage to meet this and of course to overcome the back-emf.

Analogue amplifiers similar to high-precision music amplifiers were widely used in the early years of MRI. Duty cycles achievable with this amplifier type are limited due to losses in the semiconductors. **P**ulse-**W**idth **M**odulated (PWM) amplifiers overcome this drawback and have become the standard for MRI gradient drivers in the meantime. The basic idea behind PWM amplifiers is that the amplifier always produces its maximum voltage, and its output is modulated by a binary switching pattern at frequencies ranging from 20 to 200 kHz or even higher (Figure 10.10). When the switching frequency is high enough, the gradient coil reacts to the time-averaged output of the amplifier. Filters are needed to smooth out the modulated output. PWM amplifiers are capable of delivering very high voltages (or currents) with only modest heat dissipation.

10.3.5 Acoustic Noise

The switching of gradient currents is the source of acoustic noise, which is irritating at best and, at worst, can reach levels which may damage hearing. When a gradient is on (i.e. non-zero current in the coil), it interacts magnetically with the main field due to Lorentz forces. Typically the forces act radially to compress or expand the gradient set. As the gradient currents change during the pulse sequence, the gradient set creates complex oscillatory patterns which in turn generate acoustic noise within the system bore. The noise level depends on the gradient strength, the switching time, and the mechanical properties of the gradient set. The noise in the bore can easily be higher than 85 dB (A), a noise level which is high enough to cause temporary hearing damage. That's why you should always encourage patients to use hearing protection.

There are various ways to limit the noise generation, and many manufacturers now offer special 'quiet' pulse sequences. The technology uses a mixture of limiting slew rates, avoiding gradient resonant frequencies, and so-called 'bridging' gradients. The noise reduction can bring it down to as low as 70 dB(A), which is experienced by the patient as a far nicer scan experience.

10.4 Radiofrequency Transmit Subsystem

The radiofrequency (RF) system comprises a transmitter and amplifier at the right frequency, and a transmit coil. We will first look at the transmit side in this section, and consider the receive side of the RF chain in the next section. The purpose of the RF transmit system is to generate homogeneous B_1 fields over the imaging volume, at the Larmor frequency.

10.4.1 Transmitter and RF Amplifier

The transmitter has to generate RF pulses with appropriate centre frequencies, bandwidths, amplitudes and phases in order to excite nuclei within the desired slices or slabs. The slice position and the strength of the slice-select gradient at that location determine the centre frequency of the pulse. The bandwidth, or the range of frequencies within the pulse, controls the thickness of the excited slice. The shape and duration of the RF pulse envelope determines the bandwidth. The amplitude of the RF pulse controls how much the magnetization is flipped by the pulse, while the phase controls along which axis the magnetization is flipped (in the rotating frame of reference). In modern MRI systems the RF pulse envelope is generated digitally (see Box 'On the Air: Transmitter Theory').

On the Air: Transmitter Theory

As we saw in Chapter 8, the slice-selective RF pulse is amplitude modulated by a function $S(t)$ to create a slice. It has a frequency slightly offset from Larmor in order to select the required slice position, and may also have a phase angle. The required output is

$$S(t)\cos(\omega_{SS}t+\phi)$$

where

$$\omega_{SS}=\omega_0\pm\Delta\omega$$

To avoid the possibility of stray RF reaching the receive coil from the transmitter (and completely bypassing the patient!), its internal frequency source does not operate at the Larmor frequency but at a fixed frequency ω_{fix}. For that reason ω_{SS} is only generated when the RF pulse is about to be applied, by combining a variable offset frequency ω_{off} with the fixed frequency ω_{fix}:

$$\begin{aligned}\omega_{SS}&=\omega_{\text{fix}}-\omega_{\text{off}}\\ \omega_{\text{off}}&=\omega_{\text{fix}}-\omega_0-\Delta\omega\end{aligned}$$

This combination is performed in an electronic 'mixer' that multiplies the two frequencies and generates the sum and difference frequencies, known as side-bands, using the trigonometric identity

$$\begin{aligned}2\cos(\omega_{\text{fix}}t)\cos(\omega_{\text{off}}t)=&\cos((\omega_{\text{fix}}+\omega_{\text{off}})t)\\ &+\cos((\omega_{\text{fix}}-\omega_{\text{off}})t)\end{aligned}$$

We want the lower side-band at (ω_{fix}–ω_{off}), and the unwanted side-band at the higher frequency (ω_{fix}+ω_{off}) is filtered out.

To generate the RF pulse, the transmitter (Figure 10.11) takes the digitally generated amplitude-modulation function $S(t)$ and mixes it with the offset frequency ω_{off} and phase ϕ. This waveform is then passed to a digital-to-analogue converter (DAC). The power of this analogue RF pulse coming from the transmitter will only be about 0.1 mW. In order to perturb the spins in a patient this is amplified using an RF power amplifier.

Radiofrequency amplifiers are characterized by the frequency bandwidth, power range, and linearity of response. However, designing an amplifier for good linearity typically makes it inefficient for power, and vice versa. Phase and amplitude stability must be maintained during the whole pulse sequence, i.e. linearity is more important than power. To convince yourself of this, imagine that the amplifier's output amplitude is unstable, and generates different RF powers in each TR. This would mean different flip angles, and therefore time-varying behaviour of the MR signal. After Fourier transformation, that becomes a series of ghost images, interfering with

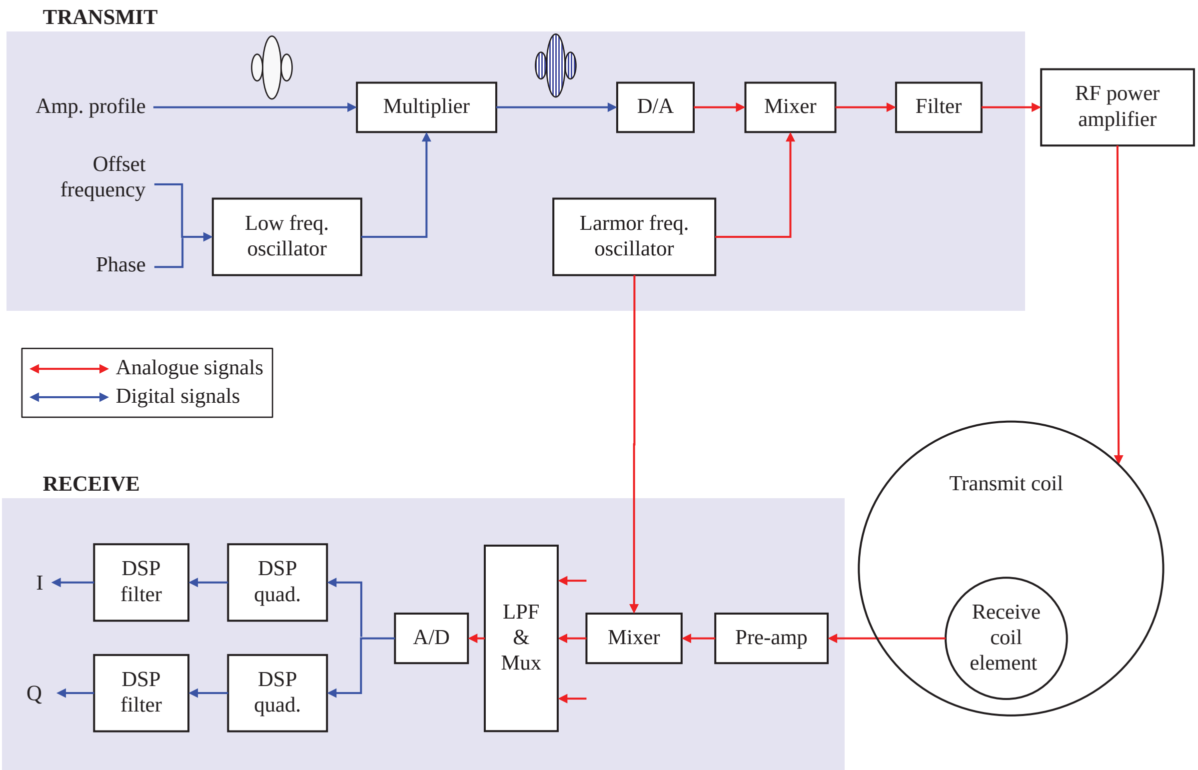

Figure 10.11 Digital radiofrequency transceiver. D/A denotes digital-to-analogue converter; A/D, analogue-to-digital converter; LPF, low-pass filter; Mux, multiplexer. See text for details on the theory of operation.

the diagnostic quality of the final image. Similarly, if the output phase is unstable, the MR response will vary per TR, leading to ghosting, loss of SNR and changes in image contrast. Modern solid state amplifiers use MOSFET transistors which can provide up to 1 kW power with a flat frequency response up to 500 MHz. Several transistor pairs are combined together to give a total output of up to 30 kW. These transistors generate a lot of heat, and the RF amplifier often has a cooling system using chilled water (like the gradient subsystem) or forced air.

The latest MR systems use parallel transmit technology to improve B_1 homogeneity, which is a particular problem at 3 T. The RF wavelength at 3 T is approximately 25–30 cm, which leads to more standing waves within the body. This leads to flip angle variations across the field of view and 'shading' effects, particularly noticeable in the liver, breast (Figure 10.12a) and in the thoracic spine. Conventional systems drive the transmit coil in quadrature, by splitting the output from a single RF amplifier and applying a 90° phase difference between two ports on the coil, which are also 90° apart (Figure 10.12b). Parallel transmission works by driving the two ports on the coil by two separate amplifier chains (Figure 10.12c). This allows for fully independent control of amplitude and phase on each channel, and the B_1 field can be optimized to suit the anatomy of the patient in the bore, producing more uniform signal intensity (Figure 10.12d). Multi-transmit systems have two RF amplifiers, which must also be synchronized with each other with sub-nanosecond accuracy. In principle, parallel transmit systems can be designed with more than two channels; however, the benefits are limited at 3 T. In research systems at e.g. 7 T and higher, multiple channels may be necessary to achieve a uniform flip angle excitation in the body.

10.4.2 B_1 Mapping

On systems with full multi-transmit capabilities, it is important to map the spatial distribution of the transmit field B_1^+ in vivo as a calibration scan. This allows the system to adjust the output amplitudes and phases of the RF chain so that the optimum B_1 field can be generated for each patient and each anatomy.

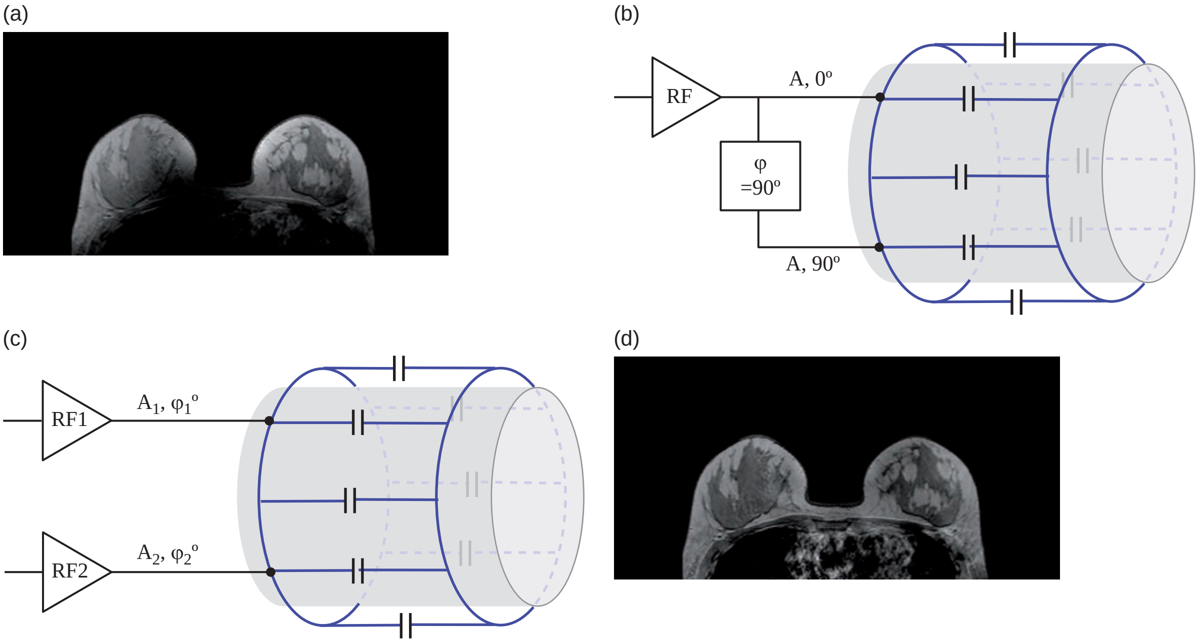

Figure 10.12 (a) Dark shading in breast image caused by B_1 inhomogeneity at 3 T. (b) Conventional transmit subsystem. (c) Modification of amplitude and phase in two independent transmit channels creates a more homogeneous B_1 field resulting in (d) a more uniform breast image.

Alternatively, specially shaped B_1 fields can be created to selectively excite a sub-volume within the patient's body. Many quantitative imaging procedures also rely on accurate flip angles, so B_1^+-mapping is sometimes used in those techniques too (Chapter 19). There are two main ways to map B_1^+: a dual-TR method (see Box 'Dual-TR Method for B_1 Mapping'), and a double angle method which is easier to try for yourself since it can be performed using standard pulse sequences. The double angle method, as the name suggests, involves the acquisition of two gradient-echo images with flip angles of α and 2α. This produces two images S_α and $S_{2\alpha}$ with signal intensities depending on $\sin\alpha$ and $\sin 2\alpha$ respectively. To produce the B_1^+ map, we find the flip angle α by first taking the ratio $S_{2\alpha}/S_\alpha$, and then using the relationship $\sin 2\alpha = 2\sin\alpha\cos\alpha$:

$$\frac{S_{2\alpha}(x,y)}{S_\alpha(x,y)} = \frac{2\sin\alpha\cos\alpha}{\sin\alpha}$$
$$\alpha(x,y) = \cos^{-1}\left[\frac{S_{2\alpha}(x,y)}{2\cdot S_\alpha(x,y)}\right]$$

A limitation of this method is that the longitudinal magnetization must recover nearly to equilibrium to prevent different T_1 weightings in the two images, so the TR must be rather long. Also the technique is rather inaccurate for $80° < \alpha < 100°$ as $\sin\alpha$ hardly varies over this range; also at low flip angles $<20°$, the SNR is rather low and noise bias will reduce the accuracy. Specialized pulse sequences can remove these limitations, but they are not always commercially accessible.

Dual-TR Method for B_1 Mapping

The dual-TR method, proposed by Yarnykh in 2004, uses two interleaved scans with different TRs (Figure 10.13). Using the Bloch equations to describe the steady state signals, we again use a ratio of the two images:

$$\frac{S_2}{S_1} = \frac{1 - E_1 + (1 - E_1)E_2\cos\alpha}{1 - E_2 + (1 - E_2)E_1\cos\alpha}$$

where $E_1 = \exp(-TR_1/T_1)$ and $E_2 = \exp(-TR_2/T_1)$. Using a Taylor expansion for the exponentials, and assuming that both TR_1 and TR_2 are very short compared with T_1, this ratio r reduces to

$$r = \frac{S_2}{S_1} = \frac{1 + \eta\cos\alpha}{\eta + \cos\alpha}$$
$$\alpha = \cos^{-1}\frac{\eta\cdot r - 1}{\eta - r}$$

where $\eta = TR_2/TR_1$. Provided $TR_1 < TR_2 < T_1$ of the tissue, the accuracy of the flip angle calculation is independent of T_1, so the dual TR method is particularly useful in vivo.

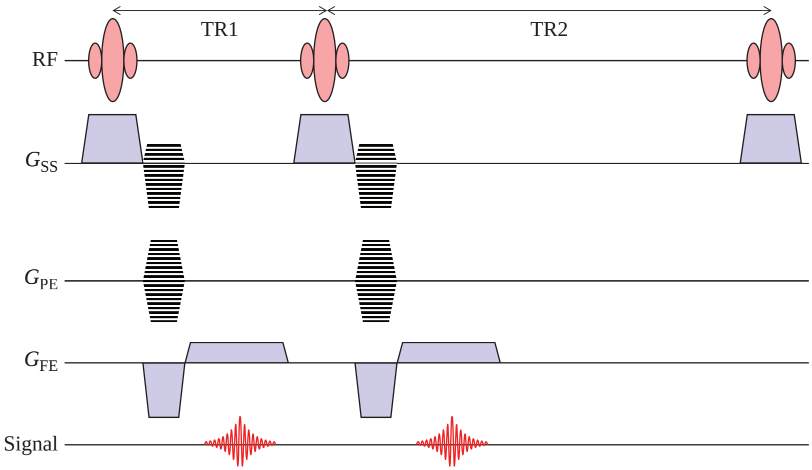

Figure 10.13 Pulse sequence diagram for dual-TR B_1-mapping technique.

10.4.3 Transmit Coils

The coils used to excite the MR signal must produce a uniform field B_1 at right angles to the static magnetic field. Transmit coils are usually large to optimize their uniformity, and encompass a significant volume of tissue. A transmit coil may also be used to receive the MR signals, provided an appropriate **T**ransmit/**R**eceive (T/R) switch is used, but this is relatively uncommon (because it has lower SNR than using local receive coils). The T/R switch protects the receive circuitry from the very high voltages applied during transmit and also prevents the small MR signal from being lost in the electronic noise generated by the transmitter even in its off state.

The main transmitting coil is usually the body coil, which surrounds the entire patient. This is usually built into the scanner bore and is not generally visible. Since this coil is large it has a very uniform transmission field, but this also means that it is not particularly sensitive if used as a receive coil. In some systems other coils, e.g. head or knee, may also be used for transmission, in which case less power is required to flip the magnetization, but excitation uniformity may be sacrificed.

In typical cylindrical MR systems, the body coil usually has a birdcage design (Figure 10.14a). The number of elements is optimized so that the current distribution over the surface of the coil varies sinusoidally. Birdcage coils are always driven in quadrature, meaning that the output from the RF amplifier is connected to the coil at two ports, not just one.

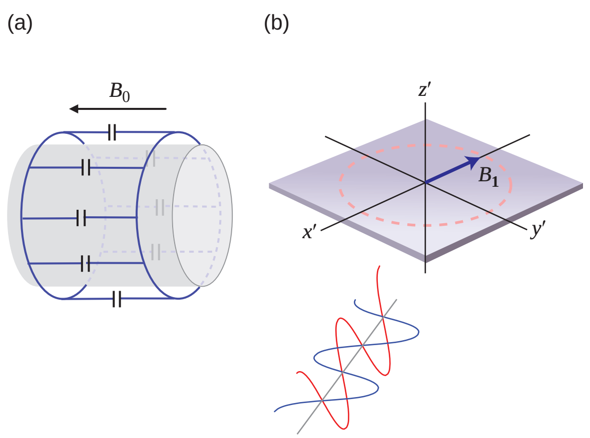

Figure 10.14 (**a**) Low-pass birdcage coil. (**b**) Quadrature transmission creates a circularly polarized B_1 field.

The two currents are 90° out of phase, and this combination generates just one rotating B_1 field, rotating in the same sense as the protons (Figure 10.14b).

Birdcage Coils

The birdcage coil physically consists of two circular end rings connected by N equally spaced straight conductors. The straight conductors and segments of end rings connecting adjacent conductors can be treated as inductors. In a birdcage coil the necessary sinusoidal variation in current is generated by the components acting to delay the current around the conductors. In the low-pass configuration each of the straight conductors includes a capacitance C, while in the high-pass configuration the capacitance is placed on the end rings. Breaking the two end rings and unrolling the coil results in an equivalent circuit of N elements, each introducing a phase shift of $\Delta\phi(\omega)$. The requirement for a sinusoidal current distribution is that the total phase shift around the coil must equal a multiple of 2π

$$N\Delta\Phi(\omega) = 2\pi M$$

where M is the resonant mode number and $1 \leq M \leq N/2$. A standing wave in the $M = 1$ mode produces the most homogeneous B_1 field.

10.5 RF Receiver Subsystem

It can be helpful to consider the RF receiver subsystem separately from the transmit chain, since in modern MR systems they are effectively independent. The receive chain is responsible for detecting the tiny MR signal, amplifying and digitizing it without introducing unnecessary noise, and then sending it to the reconstructor for image calculation. The receiver subsystem consists of receive coils, preamplifiers, digitizers, and digital signal processing to prepare the signal for reconstruction (Figure 10.11).

10.5.1 Receive Coils

The function of a receive coil is to maximize signal detection, while minimizing the noise. Usually the major source of noise is from the patient's tissue (from the Brownian motion of electrolytes). To minimize the noise, and maximize the SNR, it is necessary to minimize the coil dimensions, i.e. the coil's volume should be filled as much as possible by the sample. A compromise needs to be made between adequate RF homogeneity and high SNR.

Play that Tune

An RF coil is essentially a tuned electrical circuit that comprises an inductor (the actual coil wires) and a capacitor connected in parallel (Figure 10.15a). The inductor has an electrical reactance $X_L = i\omega L$, where $i = \sqrt{-1}$ and L is the inductance in henrys. The capacitor has a reactance $X_C = -i/\omega C$, where C is the capacitance in farads. The parallel tuned circuit gives a sharp frequency response (Figure 10.15b), peaking when the reactances cancel, at a resonant frequency of

$$f_0 = \frac{1}{2\pi\sqrt{LC}}$$

At this frequency the impedance of the tuned circuit is a pure resistance

$$Z_p = \frac{LR}{C}$$

RF signals are usually piped around circuits using transmission lines, a typical example being coaxial

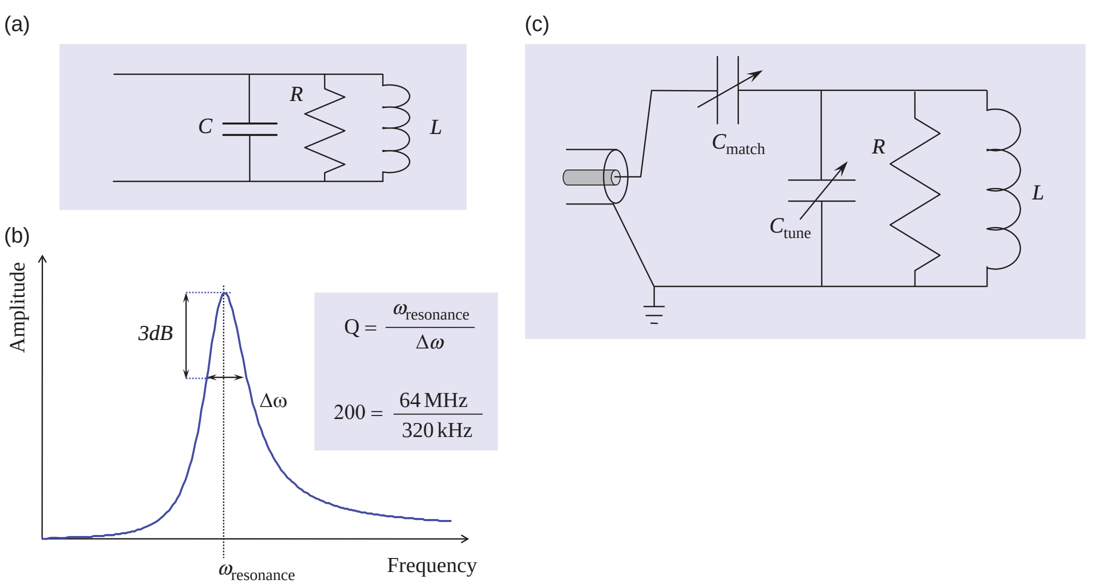

Figure 10.15 (a) Tuned circuit and (b) typical frequency response. (c) Tune-and-match circuit. A matching capacitor is added to cancel the inductive reactance at ω_0 and make the impedance of the circuit appear as a pure 50 Ω resistance.

cable. The transmission line will have a characteristic impedance of typically 50 Ω; therefore, for efficient power transfer the RF circuits should also have an input and output impedance of 50 Ω. In order to match the tuned circuit to the 50 Ω output of the power amplifier, the impedance of the coil needs to be matched to 50 Ω. At some frequency off-resonance the coil will have a resistance of 50 Ω in series with an inductive reactance. The addition of a series (matching) capacitor will cancel this inductive reactance, resulting in a pure 50 Ω resistance at this frequency (Figure 10.15c). The parallel tuning capacitor may then be adjusted to make this frequency the desired Larmor frequency. A measure of the tuned circuit (coil) performance is the Q or quality factor, which is the ratio of stored energy to dissipated energy. It may also be written as

$$Q = \frac{\omega L}{R}$$

Q is also a measure of the current or voltage magnification achieved by the tuned circuit. Q can be measured by dividing the centre frequency by the frequency difference at the half power points, or the −3 dB bandwidth (for voltage).

Coils should have reasonably high Qs, typically 200 when empty, but not too high since the circuit will continue to oscillate after the RF pulse. A good Q means that the frequency response is quite narrow and the coil is behaving like a band-pass filter, eliminating noise from outside the bandwidth of interest.

When a conducting sample, such as a human body, is placed inside the coil the Q (the loaded Q) decreases. The coil inductance also changes due to mutual inductance between the coil and the conducting tissues, changing its resonant frequency and impedance. In some systems the unloaded Q is deliberately decreased so that the coil has a fixed tuning for more operational convenience.

Birdcage coils used for transmit can also receive in quadrature. During reception the signals from the two-quadrature modes add constructively, while the noise from each is uncorrelated, i.e. it 'averages out', resulting in a $\sqrt{2}$ improvement in SNR over a comparable linear coil. Further consideration of SNR is given in Chapter 11. Note that both transmitter and receive coils require tuning and matching during manufacture, described in Box 'Play that Tune'.

Some receive coils are single loops or figure-of-eight coils which are placed directly over the anatomical region of interest. The signal response of a surface coil is non-linear with depth, resulting in an intensity fall-off into the patient. Surface coils are therefore only useful for imaging structures that lie relatively close to the surface of the patient. Flexible surface coils are very useful since they can be wrapped around the region of interest. Care must be taken to ensure that the surface coil is orientated perpendicular to B_0 otherwise no signal will be detected.

10.5.2 Phased Array Coils

The majority of receive coils in modern systems consist of multiple elements, each acting independently. These coils are formally known as 'phased array' coils, but since they are now so common, we often just use the word 'coil'.

These coils need to be designed very carefully so that the individual elements do not interact with each other, a phenomenon known as coupling which reduces SNR in the images. One way of effectively 'decoupling' one element from its neighbours is to geometrically overlap the coils in a particular way. Ideally, each element is connected to an entirely separate preamplifier and receiver (see Sections 10.5.3 and 10.5.4) which has the advantage that the noise in each receiver is completely different, i.e. uncorrelated, resulting in a higher SNR in the final image (Figure 10.16).

Phased array coil imaging generates more raw data than imaging with a simple single receive coil, and needs extra reconstruction memory in order to keep reconstruction times reasonable. A key advantage of multi-element coils is the possibility of using parallel imaging techniques such as SMASH or SENSE (see Chapter 14 for full details). Parallel imaging is extremely useful for many techniques, particularly 3D CE-MRA, steady state imaging and for high-field (3 T) imaging since it offers another way to reduce SAR.

It is useful to be able to connect two or more array coils simultaneously, particularly for whole-body screening where the patient is imaged from head to toe using several stations. With this technique, a particular FOV is scanned (e.g. the head and neck), then the patient is automatically moved into the magnet by a fixed distance and the next station is scanned (e.g. the upper thorax). This is repeated up to six times to provide complete coverage without having to change coils. It is worthwhile to take extra care when

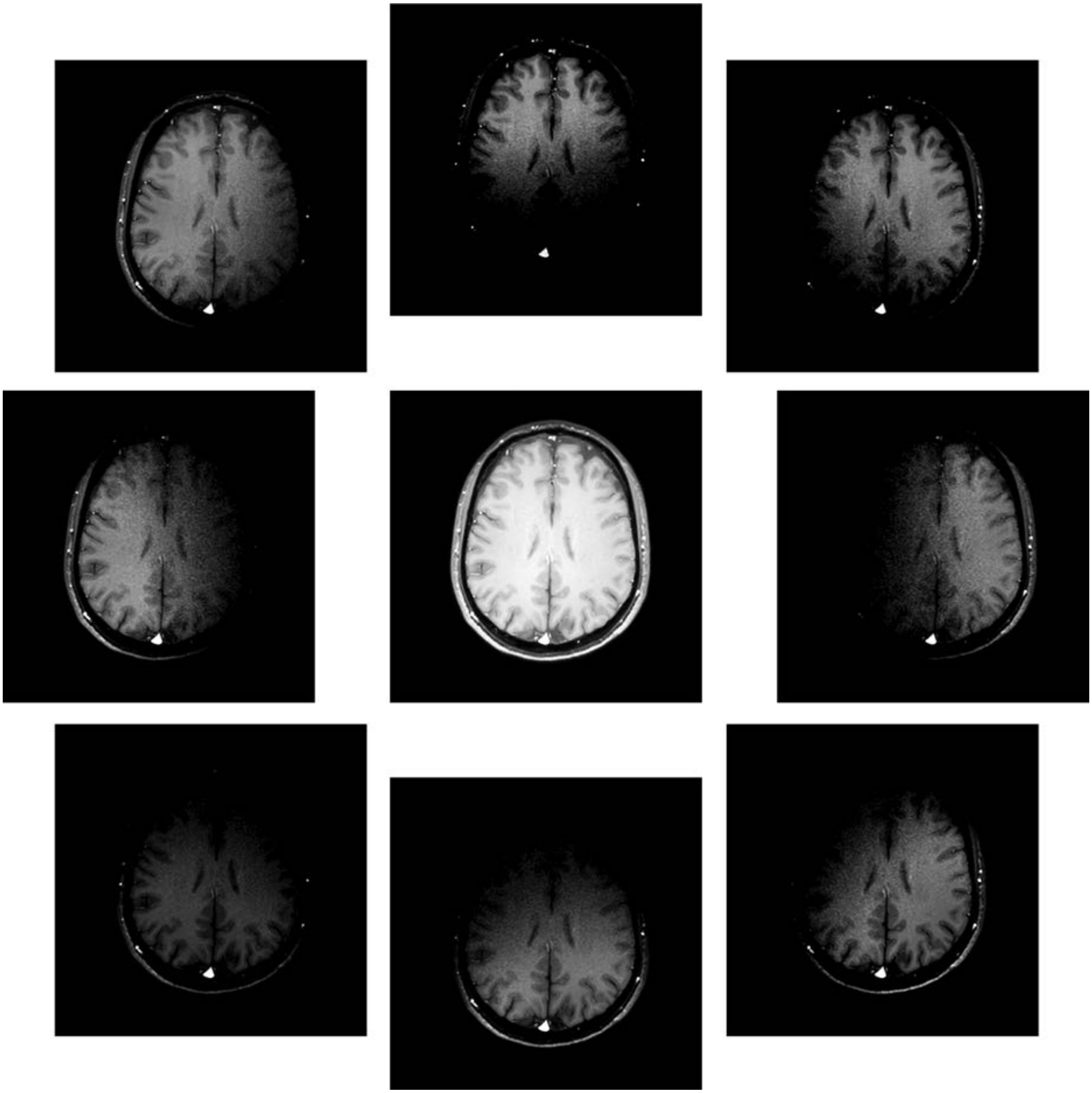

Figure 10.16 Separate images from each of the eight channels of a head phased array coil. The final combined image is also shown.

comparing receive systems: manufacturers' brochures often specify the maximum number of coil elements which can be connected at one time. But can you actually use them all at the same time, or is this just useful for reducing patient set-up time? Also be careful to check how many actual receivers (A/D channels) are provided. It is possible to combine two or three coil elements through a single receiver, but you should be aware that this reduces SNR and limits the receiver bandwidth. Buying a system with a limited number of channels may be cheaper initially, but it may have significant financial impact in the future if you need to buy more channels to support new phased array coils with many more elements.

10.5.3 Preamplifier

The small MR signal detected by the coil needs to be boosted by an extremely low-noise preamplifier before being fed to the receiver. Preamplifiers are often mounted on the coil assembly to avoid degradation of the signal by the leads. The quality of an amplifier is given by its noise figure, which is measured in decibels (dB, see Appendix). The noise figure is defined as

$$\text{NF} = \frac{\text{SNR at output}}{\text{SNR at input}}$$

and should be less than 1 dB typically.

10.5.4 Receiver (Digitizer)

The MR signal is within a narrow frequency range of interest $\Delta\omega$, embedded in or carried by the Larmor frequency ω_0 (with an offset due to the frequency encoding). In many MR systems, it is not possible to digitize directly at the Larmor frequency, so the MR signal has to be demodulated down to a lower frequency (e.g. 125 kHz). The signal is then low-pass filtered before being converted to a digital signal (Figure 10.11). The signal is then separated into the two quadrature channels using digital signal processing – see Box 'Digital Dexterity: Receiver Theory' for details. In these systems, the A/D converter is the limiting factor in the receiver chain. To achieve maximum SNR, each element of the receive coil should be connected to a separate A/D converter. However, it is practical to use a multiplexer ('Mux' in Figure 10.11) to share a single receive channel (A/D) between two or three coil elements. This limits the minimum receive bandwidth and also sacrifices SNR.

Analogue-to-digital converters can be much faster, operating at very high frequencies (tens of MHz), and still allow at least 16-bit digitization. This allows for direct digitization at the Larmor frequency, avoiding the first analogue demodulation step. With further advances in miniaturization of electronics, these powerful receivers can be mounted close to the scanner or even within the coil itself, avoiding the need for lossy copper cables carrying the MR signal into the technical room. This also has the advantage of minimizing image artefacts caused by drift in the older analogue receiver circuitry (Figure 10.17).

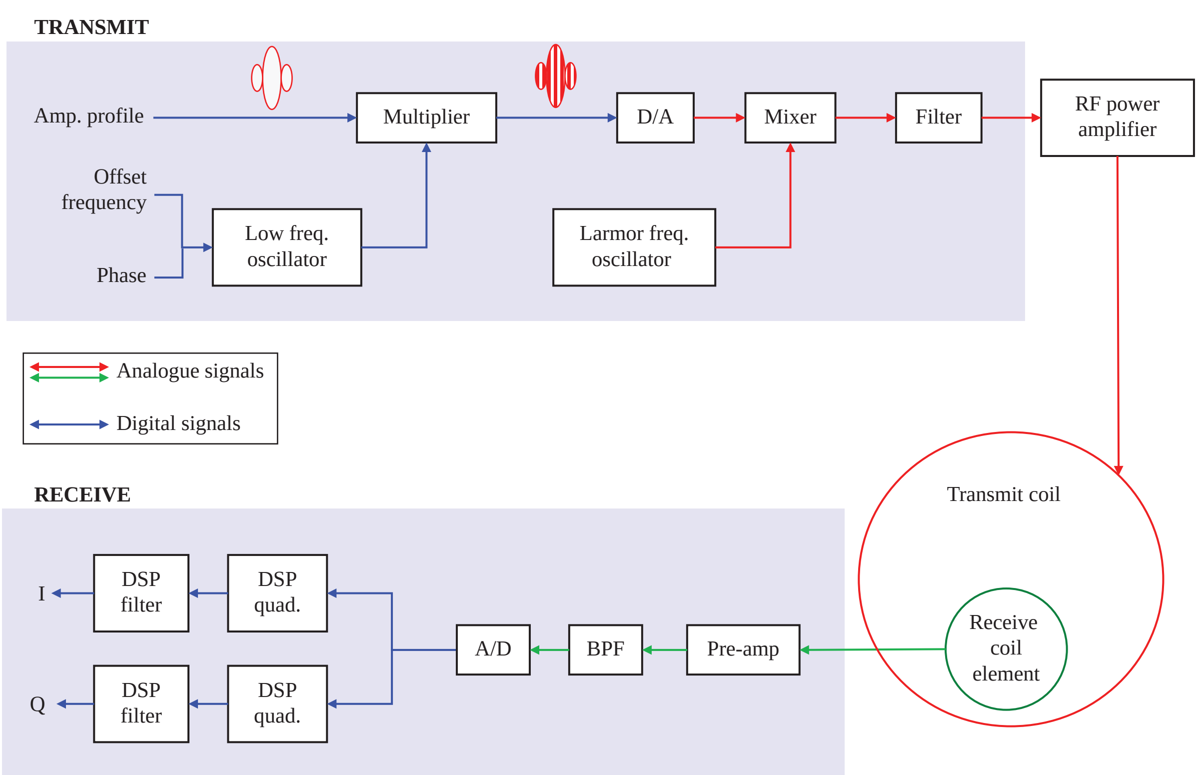

Figure 10.17 Broadband digital radiofrequency transceiver. D/A denotes digital-to-analogue converter; A/D, analogue-to-digital converter; BPF, band-pass filter. See text for details on the theory of operation.

The latest 'broadband digital' MR systems use digitization as close as possible to the coil, and carry the signal via fibre-optic cables back to the data acquisition system or reconstruction. With this receive architecture, the coil designer has complete freedom to optimize the element layout for the anatomy and can guarantee maximum SNR from the coil by including a dedicated receiver on each element. Eliminating all the analogue circuitry has made a huge improvement in the SNR available from modern scanners and allows you to buy new coils without the hassle (and expense!) of upgrading the whole receiver system.

Digital Dexterity: Receiver Theory

Although the analogue-to-digital converters are working at very high frequencies, they are typically lower than the Larmor frequency, which means they measure a series of alias frequencies ω_A of the true signal,

$$\omega_A = (\omega_0 + \omega_{FE} \pm \Delta\omega) - n \cdot \omega_S$$

where ω_S is the sample frequency of the ADC, and n is an integer. Provided that ω_S and n are chosen such that the bandwidth of the received MR signal is well within the sampled bandwidth, i.e.$\Delta\omega \ll \omega_S$, the MR signal is accurately measured.

Back in the technical room, the digital signal is processed to remove the Larmor frequency, and to separate the real and imaginary channels. The latter process is known as 'quadrature detection' and involves splitting the incoming signal and creating a 90° phase shift between the two components. In traditional MR systems this is done with analogue components, combining the signal with sine and cosine waves at the Larmor frequency, and this is still a good way to understand what is happening. Mathematically we describe the MR signal as

$$\cos(\omega_0 + \omega_{FE} \pm \Delta\omega)$$

using the cosine because it is symmetrical about the mid-point of the readout window. This signal is split and then combined with sine and cosine waves at the Larmor frequency:

$$\begin{aligned}\cos(\omega_0 + \omega_{FE} \pm \Delta\omega) \cdot \sin(\omega_0) &= \frac{1}{2}\sin(\omega_0 + \omega_{FE} \pm \Delta\omega + \omega_0) \\ &\quad + \frac{1}{2}\sin(\omega_0 + \omega_{FE} \pm \Delta\omega - \omega_0) \\ &= \frac{1}{2}\sin(2\omega_0 + \omega_{FE} \pm \Delta\omega) + \frac{1}{2}\sin(\omega_{FE} \pm \Delta\omega)\end{aligned}$$

$$\begin{aligned}\cos(\omega_0 + \omega_{FE} \pm \Delta\omega) \cdot \cos(\omega_0) &= \frac{1}{2}\cos(\omega_0 + \omega_{FE} \pm \Delta\omega + \omega_0) \\ &\quad + \frac{1}{2}\cos(\omega_0 + \omega_{FE} \pm \Delta\omega - \omega_0) \\ &= \frac{1}{2}\cos(2\omega_0 + \omega_{FE} \pm \Delta\omega) + \frac{1}{2}\cos(\omega_{FE} \pm \Delta\omega)\end{aligned}$$

Looking at both terms, you can see that one is at a very high frequency $>2\omega_0$, while the other term is just slightly offset from zero (by ω_{FE}). The high-frequency terms are discarded, leaving the real and imaginary components of the signal:

$$\cos(\omega_0 + \omega_{FE} \pm \Delta\omega) \cdot \sin(\omega_0) = \underbrace{\frac{1}{2}\sin(2\omega_0 + \omega_{FE} \pm \Delta\omega)}_{\text{high frequency side-band}} + \underbrace{\frac{1}{2}\sin(\omega_{FE} \pm \Delta\omega)}_{\text{real component}}$$

$$\cos(\omega_0 + \omega_{FE} \pm \Delta\omega) \cdot \cos(\omega_0) = \underbrace{\frac{1}{2}\cos(2\omega_0 + \omega_{FE} \pm \Delta\omega)}_{\text{high frequency side-band}} + \underbrace{\frac{1}{2}\cos(\omega_{FE} \pm \Delta\omega)}_{\text{imaginary component}}$$

Analogue quadrature detection relies on having a perfect 90° phase difference between the sine and cosine waves – controlled by another piece of electronics, which is susceptible to temperature changes and other interference. In today's scanners, quadrature detection is done digitally: alternate digital samples are sent to the real and imaginary channels, and then every other point on each side is inverted. In effect the real channel is the input signal multiplied by {0, 1, 0, –1}, the imaginary by {1, 0, –1, 0}, which corresponds to a sine and cosine respectively at the intermediate frequency, with perfect 90° phase difference between them. With digital receivers, artefacts such as quadrature ghosts or DC offset artefacts belong in the past; you will see them described in online artefact galleries, but you will never see them on your own scanner.

10.6 Computer Systems

The multi-tasking nature of MR means that it is impractical to control the many processes requiring accurate timing directly from the main or host computer, so many subsystems will have their own microprocessors whose commands can be downloaded from the host (refer back to Figure 10.1). A typical MRI system will have a host computer on which the operator will prescribe the scan in terms of the pulse sequence, its timing and various geometry factors, etc. These parameters will then be converted into commands that are transferred to another microprocessor system, known as the **P**ulse **P**rogrammer (PP) or **D**ata **A**cquisition **S**ystem (DAS) that controls the hardware. The PP or DAS ensures that the RF, gradients and data acquisition are all properly synchronized. Once the data have been acquired, the image reconstruction may also be performed on a separate computer, or on a specialist array processor or GPU (just like the graphics card in your favourite games console). The finished images are sent back to the database on the host computer, which also manages the image display, processing, filming, archiving and networking.

In a purchasing decision, the most important factors to know are the image database capacity, and the reconstruction speed and capacity. You should look for enough database capacity for 4–5 days of patient data (this obviously will depend on how busy your unit is). Reconstruction capacity should be as big as possible – the larger the better – especially if you will regularly use coils with 20 elements or more. Reconstruction speed is not easy to measure; look for the reconstruction 'latency', which is the time between the end of the scan acquisition and the last image being available to view.

10.7 Siting and Installation

When purchasing an MRI system there are a number of considerations with respect to the siting and installation of the system that need to be addressed. Obviously you need to allow enough space in the magnet room to contain the system and, ideally, the fringe fields up to the 0.5 T (5 G) line. The technical room must be next to the magnet room and will contain air conditioning and the magnet compressor unit, as well as a dedicated mains electrical supply. It may be possible to contain some of the fringe field in the technical room, but you should take advice from the manufacturer's siting specialists. Finally you need space for the operator's console; don't make it too small, as there will often be visitors who need to view the images (referring clinicians, radiologists, trainee radiographers, etc.). When choosing a location, take into account the safety zones described in Chapter 2, and make sure you will be able to control access adequately. Remember that the fringe field extends in three dimensions, so consider the floors above and below if necessary.

10.7.1 Radiofrequency Shielding

To avoid any extraneous sources of RF from interfering with the MR signals, all whole-body scanners are placed in an RF-shielded enclosure, also known as a Faraday cage. Typically this consists of a room lined on all six sides with copper sheeting. All electrical connections between the magnet system and the system's electronics cabinets are connected via electrical filters through a 'penetration panel' or 'filter panel' in the RF shield. Special wire-embedded glass is used for windows, and any doors need to make a proper electromagnetic seal with their frame. It is possible to allow tubes or fibre-optic cable through the screen via the use of waveguides (metal pipes of a specific length and diameter through which unwanted RF signals below a certain frequency cannot pass). However, conducting wires, including power cables, should not be passed through waveguides since they will act as aerials, picking up unwanted RF interference outside the room and radiating it inside. This can create zipper artefacts on the MR images.

10.7.2 Technical Environment

Every iron or steel object in the stray field of the magnet will have a temporary magnetization induced into it. The stray fields from these items will in turn affect the homogeneity of the magnet. Providing they are static then the effect can be counteracted, up to a limit, by shimming of the magnet. Proximity to moving objects such as elevators or trucks can make siting even more complex. You should not allow architects to make structural changes around the proposed MRI unit, unless it is done in consultation with the manufacturer's installation specialists.

MRI systems also have quite significant environmental requirements, particularly with respect to adequate cooling of the equipment racks and reliable electric power. Adequate air conditioning must be installed to maintain temperature and humidity within the equipment room. The air-conditioning unit should also provide the chilled water required by the compressor and other cooling units. A good-quality electrical supply is required since the sensitive electronic components do not react well to sudden losses of power, particularly if one phase of the three-phase supply fails. If necessary, additional transient suppression devices, line conditioners or even uninterruptible power supplies (UPS) may be required. In the event of power failure, the cold head will switch off and this will result in increased helium boil-off; however, the magnet will still be at field. It is worthwhile to install a control panel in the operator room to show the status of the chilled water supplies, air-conditioning and compressor function if possible, since these services may not automatically switch back on when the power is restored.

Fluorescent lights should not be used in the magnet room since they produce RF interference. Low-voltage halogen or LED lighting is far better to avoid artefacts and also improve longevity of the bulbs.

10.8 Other Types of MRI Systems

In this book we mainly talk about cylindrical MR systems based on superconducting magnets. However, there are others, and in this section we will briefly review the main differences.

10.8.1 Open MRI Systems

Open systems are much more patient-friendly than the 'tunnels' of superconducting systems, and this is their main advantage. Such systems also offer the possibility of MRI-guided interventional procedures, for example image-guided breast or liver biopsy. Open systems are often based on permanent magnets or iron-cored electromagnets, designed to operate primarily between 0.1 T and 0.3 T. There are some open superconducting systems available in the range 0.5–1.0 T, but they are becoming rarer.

Many open systems have vertically orientated magnetic fields and have specially designed flat planar gradient coils. The RF transmit coil can have a solenoid design, which is 30–40% more efficient than a birdcage coil with the same dimensions. This can produce higher SNR, but do not expect it to behave like a 1.5 T system. There are one or two open systems that allow for imaging during weight bearing, i.e. the patient stands or sits in the imaging volume instead of lying supine.

Since open systems tend to have lower field strengths, their fringe fields are smaller and they can be located in a smaller footprint. It is not helpful to make direct comparisons between an open system and a cylindrical one; the conventional system will almost always have better gradients and a wider range of receive coils. The main reason to choose an open system is the greatly improved patient experience; during the purchase process you should pay extra attention to these details, while still ensuring that the system is able to produce the right image quality for the proposed referrals.

10.8.2 Interventional (Therapy) Systems

There is continuing interest in the use of MRI for image-guided interventional procedures. Applications have included brain biopsy and neurosurgery, catheter-guided cardiac interventions, and vascular applications. Although there have been attempts to develop dedicated interventional systems they haven't generally developed into mainstream products. Therefore most interventional procedures are now carried out using conventional or open systems. It is extremely expensive to install an MR system dedicated to only surgical interventions; many centres try to use the MR system for diagnostic imaging as well, in order to make a better business case. However, this inevitably leads to compromise for both the radiology and surgical departments, and the purchase process for an interventional MRI suite is often long-winded and complex.

10.8.3 Niche Systems

In addition to conventional whole-body systems there is also ongoing interest in niche systems for particular applications. For example, special systems are commercially available for imaging the arms or legs (joint imaging), head, breast or neonatal imaging. New prototypes appear fairly regularly at trade shows like the **R**adiological **S**ociety of **N**orth **A**merica (RSNA) or **E**uropean **C**ongress of **R**adiology (ECR); however, some of them never make it to a commercial product. Since one of their main advantages is reduced size (and therefore significantly reduced cost and installation requirements), many systems are based on low-field permanent or electromagnets. Such systems often use a more open magnet design. As with open whole-body systems, the purchasing decision is usually based on a business case for imaging only a certain type of referral in a certain location, e.g. where it is impossible to site a whole-body system. Once again, there is little point making direct comparisons between niche systems and conventional whole-body systems, since they are designed for very different purposes.

See also:

- Early daze: your first week in MR: Chapter 2
- But is it safe? Bio-effects: Chapter 20

Further Reading

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ISMRM members should also browse the 'MR Systems Engineering' educational sessions for past annual meetings: for example www.ismrm.org/13/WK09.htm [accessed 28 March 2016].