CHAPTER 14

Instrumentation

A very important aspect of the production of MRI images is the instrumentation used in the measurement. Many MR systems are commercially available, each possessing different features and capabilities that are often difficult to evaluate and compare objectively. Many of these features are based on the operating software provided by the manufacturer, but certain hardware components are common to all systems. The following sections describe the basic subsystems of an MRI scanner and technical aspects to consider when comparing scanners from different manufacturers. The major components are a computer system, a magnet system, a gradient system, a radiofrequency system, and a data acquisition system (Figure 14.1).

14.1 Computer systems

From its very origin, MRI has been a computer-driven technology. Over the last 30 years, as computers have increased in speed and capabilities, MRI scanners have taken advantage of this performance increase to become less hardware-intensive and more software-driven. This has allowed manufacturers to provide products that are more reliable and stable in hardware performance and more robust and flexible in the types of images that can be acquired.

There are three primary tasks performed on an MR scanner that are computer-based: general scanner control (user interface), image processing, and data collection (measurement control). Depending on the particular manufacturer, there may be one, two, or three computers that are present. The main or host computer controls the user interface software. It will be connected to a keyboard and one or more monitors for displaying images and text information, known as the console. The operating software for the main console enables the operator to control all functions of the scanner, either directly or indirectly. Scan parameters may be selected or modified, patient images may be displayed or recorded on film or other media, and postprocessing such as region-of-interest measurements or magnification can be performed. Several peripheral devices are typically attached to the main computer. One or more hard disks are used to

MRI Basic Principles and Applications, Fifth Edition.
Brian M. Dale, Mark A. Brown and Richard C. Semelka.

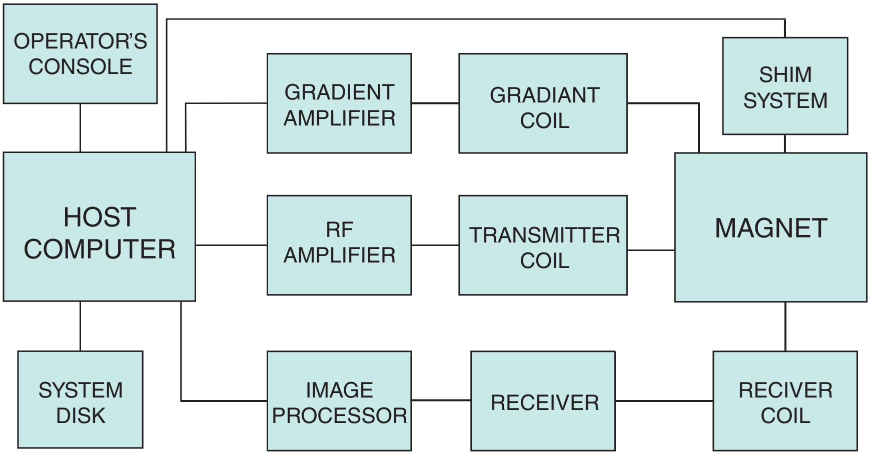

Figure 14.1 Block diagram of an MRI system.

store the patient images immediately following reconstruction. This disk or disks are used for short-term storage, with current disk drives able to store upwards of hundreds of thousands to millions of images, depending on the image size. A device for long-term archival storage, either CD or DVD, is usually included. A hard-copy camera may be connected via a network.

The image processing computer is used for performing the Fourier transformations or other processing of the detected data. This computer is synchronized with the host computer and the measurement controller. The raw data generated during the scan is stored from the receiver into memory in the image processor itself or on to a separate hard disk. Current image processing computers have multiple processors, either central processing units or graphical processing units, which allow for data processing in parallel, providing added speed in image reconstruction.

The third computer is the measurement controller, which interprets the operator-defined scan parameters and pulse sequence template files to generate the RF and gradient waveforms used to manipulate the spins. It also may be used as the monitor of the hardware to ensure its proper operation. This computer may be a separate processor or its functionality may be performed by one of the other two computers mentioned previously.

Additional consoles may be directly attached to the main computer, allowing convenient viewing or postprocessing of the images. More frequently, other viewing stations may be completely detached from the primary computer system and access the image data through a network connection. They may be used as a common viewing station for analysis of images from additional imaging modalities such as computed tomography, ultrasound, or traditional X-ray. In addition, there may be an archive server known as a PACS (picture archiving and communications system) unit. This system is used as a centralized digital repository for images. PACSs allow multiple users to access the images and patient data easily and can provide long-term archiving of the images in a central location.

MRI systems are most frequently incorporated into a computer network for the imaging facility. This allows images to be transferred directly from the MRI host computer to another computer in a remote location rather than using a removable medium (e.g., a CD). The interconnection between the two computers is normally a high-speed Ethernet connection. This is typically used for connecting computers within a local area network, typically consisting of a single building or small group of buildings; for example, between two scanners or between the scanner and a viewing station. This local network is often connected to the Internet, which is used for long-distance transmission of images and data.

Most computer networks use a communications protocol known as TCP/IP (Transmission Control Protocol/Internet Protocol) for the actual data transfer between the computers. Each computer on the network will have a unique identification number assigned to it by the network administrator, known as the IP address. This is a series of four numbers, each less than 256, (written as, for example, 121.232.22.21), which act as an electronic "street address" for the computer on the network. There will also be one computer or device on the network known as a router, which facilitates the data transfer. Data transfer is initiated from one computer (e.g., the MRI host computer) with the IP address of the destination computer (e.g., viewing station). The data are initially sent to the router, which will direct the data to the destination computer based on the IP address. Both the initiating and destination computers must have each other's IP address in order to transfer data in both directions.

While the usage of TCP/IP provides a means for transferring data between computers, there must be a common format (similar to a language) for writing the data if it is to be interpreted correctly. This is of particular importance if the two computers use software from different manufacturers, as each manufacturer will use their own proprietary format for data storage. One format that has become the industry standard for image and medical data transfer to facilitate such transfer is known as DICOM, which stands for Digital Imaging and Communications in Medicine. It is the result of a joint committee of the American College of Radiology and the National Electrical Manufacturers Association (ACR-NEMA). The DICOM standard provides a framework that allows equipment (scanners, digital cameras, viewing stations) from different manufacturers to accept and process data accurately. Images written using this standard have the basic measurement information stored so that any vendor can read and properly display the images with the correct anatomic labeling and basic measurement parameters. Manufacturers who subscribe to the DICOM standard have available a DICOM Conformance Statement, which provides details on their implementation. Programmers writing software that use DICOM-formatted images should consult the Conformance Statement for the particular manufacturer to ensure proper interpretation of the image header variables.

While a discussion of the complete DICOM standard is beyond the scope of the current discussion, two aspects of it are likely to be encountered with current MRI systems. The first is the format in which images are stored. Images stored using the DICOM format are written with the measurement parameters and other information organized in a fashion as specified in the DICOM Conformance Statement. This allows image display and analysis programs to be written without knowledge of the manufacturer's proprietary methods. Programs for reading, manipulating, and displaying DICOM-format images are available from several companies, with some programs available for free while others available for a fee.

The second aspect of DICOM that is frequently encountered is in the nature of the data transfer between computers. The DICOM standard has features that control the communication relationship between systems. This is critical to ensure confidentiality of patient information. Four communication protocols are commonly used. The first protocol controls the data transfer between a scanner and a hard-copy device, such as a laser camera. It is known as DICOM Basic_Print. This protocol enables the camera to be detached from the scanner, yet receive and process the image data over the computer network. Two other protocols control data transfer between workstations or a scanner. These work in tandem, known as the DICOM Service Class User and Service Class Provider (commonly known as Send/Receive). These protocols allow one computer system (e.g., a scanner) to send images to another system or to receive images from another system. This is normally used to connect scanners of different modalities (i.e., MR to CT). The other level of connectivity is known as the DICOM Query Service Class (commonly known as Query/Retrieve). This allows a remote computer to query the image database on a scanner and retrieve the images without requiring operator intervention. This is the normal connectivity between two scanners of the same modality or between a scanner and a PACS server. The DICOM connectivity is controlled by the Application Entity Title (AET) and both computers must have matching AETs in order for the transfer to be successful. The connectivity relationships are assigned by the network administrator during system installation or configuration based on the preferences and policies of the facility.

14.2 Magnet system

The magnet is the basic component of an MRI scanner. Magnets are available in a variety of field strengths, shapes, and materials. All magnet field strengths are measured in units of tesla or gauss (1 tesla = 10,000 gauss). Magnets are usually categorized as low-, medium-, or high-field systems. Although the categorization is not fixed, low-field magnets are usually considered to be magnets that have B_0 less than 0.5 T. Medium-field systems have B_0 between 0.5 T and 1.5 T, high-field systems have fields between 1.5 T and 3.0 T, and ultra-high-field systems have B_0 greater than 3.0 T. The magnetic field is one area where caution should be exercised. Refer to Chapter 16 for a description of safety issues regarding the magnetic field.

Magnets are also characterized by the metal used in their composition. Permanent magnets are manufactured from metal that remains magnetic for extremely long periods of time (years). They can be solenoidal (tube shaped) or have a more open design such as a C-arm or double-doughnut. Permanent magnets have minimum maintenance costs because the field is always present. However, care must be taken to keep ferromagnetic material away from the magnet. Such material will be attracted forcefully into the magnet and the magnetic field cannot be turned off to allow its extraction. Permanent magnets also have their mass concentrated over a small area. Additional structural support of the scan room may be necessary in some situations. The scan room temperature must also be very stable as fluctuations in the temperature of the magnet metal will cause the field strength to change.

The other types of magnet are electromagnets in which the flow of electrical current through wire coils produces the magnetic field (current flowing through a wire generates a magnetic field perpendicular to the direction of the current flow). The magnetic field is present as long as current flows through the magnet windings. Traditional electromagnets are made of copper wire wound in loops of various shapes. They may be also solenoidal or open-type design. A power supply provides a constant current source. Due to the limited amount of current that copper wire can carry, copper wire-based electromagnets are low-field systems. They are also sensitive to room temperature variations.

The most common type of magnets are solenoidal electromagnets using niobium–titanium alloy wire immersed in liquid helium as the magnet wire. This alloy, which has the property known as superconductivity, has no resistance to the flow of electrical current below a temperature of 20 K. It also is capable of carrying large amounts of current, enabling high magnetic field strengths to be achieved. The magnet cryostat, which contains the liquid helium, may be a double dewar design with a liquid nitrogen container surrounding the helium container or a helium-only design with a refrigeration system to reduce the helium boiloff. Refrigeration systems used with current magnets allow essentially zero helium boiloff in the course of normal operation. The cryostat and helium reservoir insulates the magnet wire so that the magnetic field is very stable and less sensitive to room temperature fluctuations. Cryogens are a second area where caution should be exercised. Refer to Chapter 16 for a description of safety issues regarding the presence of cryogens.

The primary consideration in magnet quality is the homogeneity or uniformity of the magnetic field. High homogeneity means the magnetic field changes very little over the specified region or volume. The protons in this region resonate at the same frequency in a coherent manner and thus induce the maximum possible signal. One factor affecting magnetic field homogeneity is the magnet design. Large-bore solenoidal magnets generally have the best homogeneity over the largest volume. Short-bore magnets tend to have smaller regions of good homogeneity due to the reduced number of magnet windings used. Open-design magnets will also have reduced regions of good homogeneity. Magnetic field homogeneity is usually expressed in ppm relative to the main field over a certain distance. It is assessed by measuring the field value at various locations inside the magnet and using equation (2.4) to calculate the field variation, replacing the frequencies with the measured magnetic field values. Great effort is taken during magnet manufacturing and installation to ensure the best homogeneity possible. However, manufacturing imperfections or problems with the scan room (e.g., nearby steel posts, asymmetrical metal arrangements) may produce significant field distortions. To analyze and compensate for this, the distortions are characterized by the mathematical shape of the field corrections required as a function of distance away from the magnet center. This classification is referred as the order of the field or shim correction used. First-order or linear corrections in each direction are achieved using the imaging gradient coils described in Section 14.3. Second- or higher-order corrections are nonlinear in nature and most MRI systems use a coil known as a shim coil to correct for them. The design of the shim system may be passive in that it holds pieces of metal (shim plates) or small magnets that correct the field distortions or active in that there are loops of wire through which current passes to correct the field distortions (usually through second order only). In some systems, both types of shim correction may be used. Passive shimming is generally performed at the time of magnet installation as a one-time event. Active shimming (also called electrical shimming) is usually performed on a regular basis during system maintenance and also for each individual patient. Field homogeneity is an important factor to consider when evaluating an MRI system as inadequate homogeneity can cause problems with fat saturation or even general imaging.

14.3 Gradient system

As mentioned in Chapter 4, small linear distortions to $\boldsymbol{B}_0$, known as gradient fields or gradients, are used to localize the tissue signals. Three gradients are used, one each in the x, y, and z directions, to produce the orthogonal field variations required for imaging. They are each generated by the flow of electrical current through separate loops of copper wire mounted into a single form known as the gradient coil. Variations in gradient amplitude are produced by changes in the amount or direction of the current flow through the coil. The current for each gradient axis is provided by an amplifier or power supply. The gradient amplifiers and gradient coils are actively cooled, either by air or chilled water flowing through the components, due to the heat that is generated by the current flow. Gradients are an area where caution should be exercised. Refer to Chapter 16 for a description of safety issues regarding gradients.

One of the major criteria for evaluation of an MRI scanner is the capabilities of the gradient system. There are four aspects that are important in assessing gradient system performance: maximum gradient strength, rise time or slew rate, duty cycle, and techniques for eddy current compensation. Gradient strength is measured in units of mT m^{-1} or G cm^{-1} (1 G cm^{-1} = 10 mT m^{-1}), with typical maximum gradient strengths for current state-of-the-art MRI systems being 40–100 mT m^{-1} (4.0–10.0 G cm^{-1}). These maximum gradient strengths allow thinner slices or smaller FOVs to be obtained without changing other measurement parameters. Another quantity often used to describe maximum gradient amplitudes is the effective gradient amplitude G_{eff} (Figure 14.2). This is the instantaneous vector sum of all three gradients when applied during the scan:

$$G_{eff} = (G_x^{\,2} + G_y^{\,2} + G_z^{\,2})^{1/2} \tag{14.1}$$

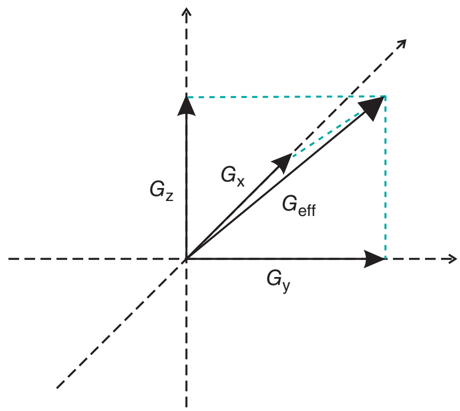

Figure 14.2 The effective gradient G_{eff} is the vector sum of the gradients (G_x, G_y, G_z) in all three directions.

Care must be made to distinguish between the effective gradient and the actual gradient being applied. For example, for a slice selection, orthogonal slices will only use one gradient axis, while only double oblique slices will require all three axes to be active during RF transmission.

The response of a gradient coil to the flow of current is not instantaneous. Gradient pulses require a finite time known as the rise time to achieve their final value. This rise time is nominally 0.1–0.3 ms, which determines a rate of change or slew rate for a gradient pulse. If the desired gradient pulse amplitude is 20 mT m^{-1} with a 0.2 ms rise time, the slew rate is 100 mT m^{-1} ms^{-1} or 100 T m^{-1} s^{-1}. Gradient rise times and/or slew rates are often used to evaluate the performance of the gradient amplifier or power supply producing the current. High-performance amplifiers allow shorter rise times (faster slew rates), enabling shorter gradient pulse durations and/or interpulse delays within a pulse sequence. As a result, the minimum *TE* may be reduced for a given technique while maintaining a small FOV.

The duty cycle of the gradient amplifier is another important measure of gradient performance. The duty cycle determines how long the amplifier can sustain its response to the demands of a pulse sequence. Duty cycles of 100% at the maximum gradient amplitude are typical for state-of-the-art gradient amplifiers for normal imaging sequences. Large duty cycles allow high-amplitude gradient pulses to be used with very short interpulse delays. Low duty cycles mean that the *TE* times will be longer for the scan to allow the gradient amplifiers to return to a standard operating state.

Another complication of gradient pulses is eddy currents. Eddy currents are electric fields produced in a conductive medium by a time-varying magnetic field. In MRI systems, eddy currents are typically induced by the ramping gradient pulse in the body coil located inside the gradient coil and the cryoshield (the innermost portion of the magnet cryostat) outside the coil. These currents generate a magnetic field that opposes and distorts the original gradient pulse. In addition, once the gradient plateau is reached and the ramp is stopped, the eddy currents begin to reduce in amplitude. The net result is that the distorting field will change with time as the eddy currents decay. Therefore the magnetic field homogeneity and the corresponding frequencies change with time as well. Correction of these eddy-current-induced distortions is known as eddy current compensation. Two approaches are commonly used for compensation. One method predistorts the gradient pulse so that the field variation inside the magnet is the desired one. This predistortion may be done via hardware or software. A second approach uses a second set of coil windings surrounding the main gradient coil. This approach is called an actively shielded gradient coil, analogous to the actively shielded magnet described previously. The current flow through the shield coil reduces the eddy currents induced in structures outside the coil. Typical state-of-the-art scanners use both methods of eddy current compensation.

14.4 Radiofrequency system

The RF transmitter system is responsible for generating and broadcasting the RF pulses used to excite the protons. The RF transmitter contains four main components: a frequency synthesizer, the digital envelope of RF frequencies, a high power amplifier, and a coil or antenna. As discussed in Chapter 5, each RF pulse has both a frequency and a phase defined for it. These features are determined by the combination of frequency and phase from the frequency synthesizer and the RF envelope defining the pulse shape.

The frequency synthesizer produces the center or carrier frequency for the RF pulse. It also provides the master clock for the measurement hardware during the scan. The frequency synthesizer also provides a phase reference for the scan. Many pulse sequences alternate the phase of the excitation pulse for each line of data by 180° to help reduce stimulated echo artifacts caused by pulse imperfections. Spin echo sequences also typically have the refocusing RF pulses shifted in phase 90° relative to the excitation pulse (known as a Carr–Purcell–Meiboom–Gill, or CPMG, technique). This phase variation may be done through modulation of the RF envelope or of the carrier frequency. More sophisticated synthesizers allow phase changes of 1–2° increments. This finer control also allows for coherence spoiling through incremental phase change of the transmitter, a process known as RF spoiling, used in spoiled gradient echo techniques (Chapter 9).

The RF envelope is generated as a discrete envelope or function containing a range or bandwidth of frequencies. It is made analog and mixed with the carrier frequency prior to amplification to produce an amplitude- or phase-modulated pulse centered at the desired frequency. For some scanners, the final frequency is produced exclusively by the frequency synthesizer, while for other scanners, the RF envelope is modulated to incorporate a frequency offset into the pulse. In either case, the final frequency is determined based on equation (4.1) and generated as a phase coherent signal by the synthesizer.

The RF power amplifier is responsible for producing sufficient power from the frequency synthesizer signal to excite the protons. The amplifier may be solid state or a tube type. In both cases, the amplification is nonlinear in amplitude and phase, causing distortions of the waveform being broadcast. It is necessary to perform some type of nonlinearity correction for the output of the transmitter. Consult the manufacturer for details on the type of correction that is used. Typical RF amplifiers for MR scanners are rated at 2–40 kW of output power at the ^{1}H frequency, with less power required for other nuclei. The actual amount of power required from the amplifier to rotate the protons from equilibrium depends on the field strength, coil transmission efficiency, transmitter pulse duration, and desired excitation angle. This energy deposition in the patient is an area where caution should be exercised. Refer to Chapter 16 on the safety issues regarding the monitoring of RF power deposition.

The final component of the RF system is the transmitter coil. All MR measurements require a transmitter coil or antenna to broadcast the RF pulses. Although transmitter coils can be any size and shape, the one requirement that must be met is that they generate an effective $\boldsymbol{B}_1$ field perpendicular to $\boldsymbol{B}_0$. Another feature of most transmitter coils is that they can produce uniform RF excitation over a desired area; that is, a volume can be defined within the coil where all protons experience the same amount of RF energy. Solenoidal MR systems use either a saddle or birdcage coil design, which produces uniform RF excitation even though the coil opening is parallel to $\boldsymbol{B}_0$. These coils are often adjusted or tuned to the patient to achieve the maximum efficiency in RF transmission. Two types of coil polarity are used: linear polarized (LP) and circularly polarized (CP), also called quadrature. In an LP system, a single coil system is present and the RF pulse is broadcast as a plane wave. A plane wave broadcast at a frequency ω_{TR} has two circularly rotating components, rotating in opposite directions at the same frequency ω_{TR} (Figure 14.3a). For MR, only the component rotating in the same direction as the protons (in-phase) induces resonance absorption. The other component (out-of-phase) is absorbed by the patient as heat. In a CP transmitter system, two coils are present, one rotated 90° from

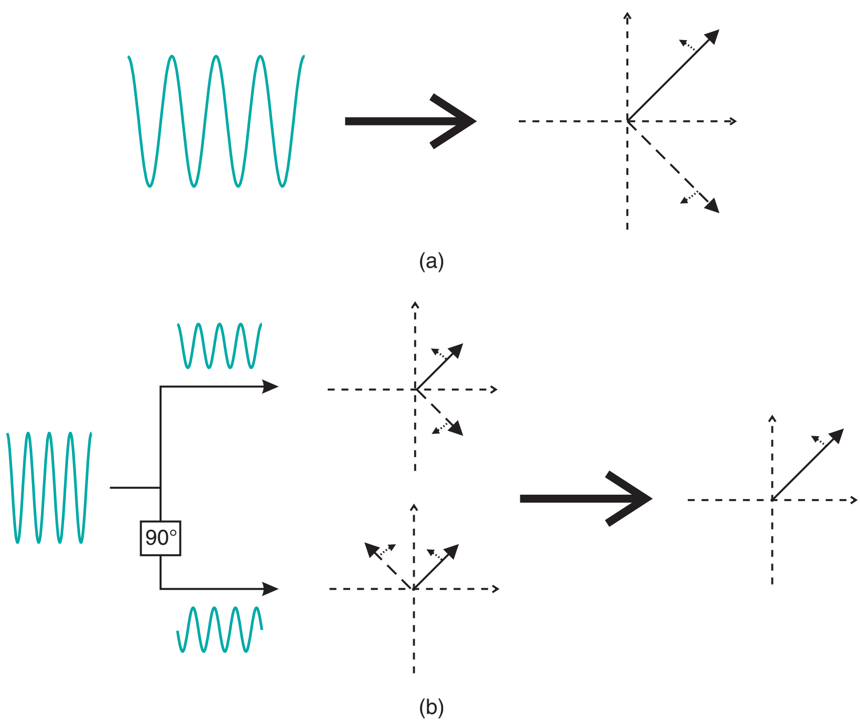

Figure 14.3 Polarization of RF wave. (a) Linear polarization. A plane wave can be thought of as the sum of two circular waves rotating in opposite directions. Because the nuclear precession is in only one direction, only one component interacts with the precessing spins (solid arrow) and is effective at resonance absorption. The other component will be absorbed as heat. (b) Circular polarization. A plane wave can be divided into two parts. If one part is shifted 90° relative to the other wave, each part will have one component interacting with the precessing spins (solid arrows), which add together. The counter-rotating components (broken arrows) are always 180° from each other and cancel. The resulting wave is a circular or helical wave with no out-of-phase component.

the other. Phase-shifted RF pulses are broadcast through each coil. The out-of-phase components cancel each other while the in-phase components add coherently (Figure 14.3b). The patient absorbs only the energy from the in-phase components from each coil. A 40% improvement in efficiency from the transmitter system is thereby achieved for a CP system relative to an equivalent LP system for the same proton rotation.

As indicted in Figure 14.3b, the RF waveform broadcast in a CP system is the same waveform except for a phase shift. An alternate approach is to send different waveforms through the two antennas. The waveforms are designed so that the combination provides the desired RF excitation. This concept is referred to as parallel transmission. It has the advantage that a limited volume of tissue can be excited, thereby reducing the SAR. However, multiple transmitter systems are necessary, increasing the cost and complexity of the scanner. Commercial systems implementing parallel transmission have two to four transmitters while ultra-high field systems can have eight or more transmitters.

14.5 Data acquisition system

The data acquisition system is responsible for measuring the signals from the protons and digitizing them for later postprocessing. All MRI systems use a coil to detect the induced voltage from the protons following an RF pulse. The coil is tuned to the particular frequency of the returning signal. This coil may be the same one used to broadcast the RF pulse, or more commonly it may be a dedicated receiver coil. The exact shape and size of the coil are manufacturer-specific, but its effective field must be perpendicular to $\boldsymbol{B}_0$. The sensitivity of the coil depends on its size, with smaller coils being more sensitive than larger coils but covering a smaller region. Also, the amount of tissue within the sensitive volume of the coil, known as the filling factor, affects the sensitivity. For large-volume studies, such as body or head imaging, the transmitter coil can serve as the receiver coil. For smaller-volume studies, receive-only surface coils are usually used. These coils are small, usually ring-shaped, have high sensitivity but limited penetration, and are used to examine anatomy near the surface of the patient's body. Phased array coils use two or more smaller surface coils to cover a larger area. The small coils are configured so that there is minimal interference between them. This arrangement provides the sensitivity of the small coil but with the anatomical coverage of the larger coil. Current generation scanners have coil arrays containing 4 to 128 individual coils, often referred to as elements.

The signals produced by the protons are usually nV to μV in amplitude and MHz in frequency. In order to process them, amplification is required, which is usually performed in several stages. The initial amplification is performed using a low-noise analog preamplifier located inside the magnet room or built into the coil itself. Further amplification will be performed by the receiver module. Two types of receivers are used in MRI systems. Analog receivers amplify and then demodulate the measured signal relative to the input frequency from the frequency synthesizer to produce a quadrature signal (real and imaginary) with frequencies between 1000 and 250, 000 Hz (audio frequency). The signals will be filtered with bandpass filters, then digitized using analog-to-digital converters (ADCs). The ADCs digitize each analog signal at a rate determined by the sampling time and number of data points specified by the user. Typical ADCs can digitize a 10 V signal into 16 bits of information at a rate of 0.1 μs per data point. Digital receivers demodulate the signal to an intermediate frequency and then digitize using an ADC. The final signal amplification, application of a low pass filter, demodulation to an audio range, and quadrature formulation are done on the digital signal. Digital receivers allow for better signal fidelity of the final signal composed to analog receivers. The digitized data are stored on to a hard disk or on to computer memory for later Fourier transformation. Phased array coils typically have a separate preamplifier and ADC for each coil in the array.

Although not formally part of the data acquisition system hardware, an important component of an MRI scanner is RF shielding of the scan room. The weak MR signals must be detected in the presence of background RF signals produced by local radio and television stations. To filter this extraneous noise, MRI scanners are normally enclosed in a copper or stainless steel shield known as a Faraday shield. Maintaining the integrity of this Faraday shield and eliminating in-room sources is very important to minimize noise contamination of the final images.

14.6 Summary of system components

Following is a list of general system features or characteristics to consider in comparing MRI systems, according to subsystem. Individual software features offered by a manufacturer are not included.

Computer systems

- Main computer processor speed (GHz)
- Capacity of short-term storage disk (Gbytes)
- Type of archive device and capacity (Mbytes)
- Number and speed of image processors (s image^{-1})
- Number of consoles and method of interconnection
- Network capability
- Filming capabilities
- Level and nature of DICOM compliance

Magnet system

- Field strength (T)
- Field homogeneity measured over a specified diameter of a spherical volume (dsv) (ppm)
- 0.5 mT (5.0 G) distance from isocenter (in all directions) (m)
- Cryogen capacity and evaporation rate (l He day^{-1})

Gradient system

Maximum gradient amplitude per axis (mT m^{-1} or G cm^{-1})

Duty cycle (percentage)

Maximum slew rate (T m^{-1} s^{-1})

Method(s) of eddy current correction

Radiofrequency system

RF spoiling capabilities (phase behavior)

Maximum output power (kW)

Type of transmitter coils (CP, LP)

Number of transmitter systems

Operating frequency range (if multinuclear imaging or spectroscopy is planned)

Data ccquisition system

Number and type of receiver channels

Digitization speed of ADCs (minimum μs/point)

Dynamic range of receiver system, maximum number of available digital bits

Raw data storage capacity (MGbytes)

Types of receiver coils (CP, LP, phased array)

Nature and quality of RF shielding