PART 1

The basics of cardiac MR
**Introduction**

Magnetic resonance imaging (MRI) of the cardiovascular system continues to expand its application and importance as a diagnostic tool in adult and pediatric patients. In order to successfully apply this technology and interpret cardiac magnetic resonance (CMR) images, it is important to gain some basic understanding of the underlying physics, hardware, and methods used to generate and encode the images. This chapter provides a brief overview of the fundamental physics of MRI, as well as a summary of some techniques of particular importance in cardiac imaging.

**The source of the MRI signal**

MRI is based on the phenomenon of nuclear magnetic resonance (NMR). This is a “resonance” phenomenon in that a signal is emitted by the sample after radiofrequency (RF) energy is applied to it. The NMR signal is emitted by molecules of the tissue in the body, unlike X-ray imaging methods which rely on the attenuation of externally applied radiation by tissues or injected contrast agent, or nuclear imaging methods based on the detection of radiation from injected radioisotopes.

The atomic nucleus is comprised of protons and neutrons that have magnetic fields associated with their spin and charge distributions. The resonance phenomenon refers to the ability of some nuclei to selectively absorb and later release energy specific to the element and its chemical environment. Not all elements are capable of resonance; it requires an odd number of protons or neutrons for a nucleus to exhibit a magnetic moment associated with its net spin. The hydrogen atom, for example, consists of a single proton with no neutrons, giving it a net spin \( \frac{1}{2} \). While there are several biologically relevant candidates for MRI, such as \( ^{17}\text{O}, ^{19}\text{F}, ^{23}\text{Na}, \) or \( ^{31}\text{P} \), it is the hydrogen (\( ^{1}\text{H} \)) atom’s large magnetic moment as well as its isotopic abundance and biological abundance that make it the primary choice for MR imaging. Hydrogen is found in water molecules and in the methylene (\( \text{CH}_2 \)) groups of fat; both highly abundant in living tissue. The magnetic moment of each individual hydrogen proton is small, but the additive effect of many magnetic moment vectors makes it detectable in MRI.

**The Larmor equation**

Under normal conditions the net magnetization of a tissue sample is zero due to the random orientations of the individual protons or “spins” (Figure 1.1a), but this changes when the sample is placed in a strong magnetic field (Figure 1.1b). The magnetic field strength generated by clinical MRI systems ranges from 0.15 Tesla (T) to 7T, with CMR most commonly performed at 1.5T. In comparison, the Earth’s magnetic field is approximately .05 mT at the surface of the planet. When subjected to any external magnetic field, nuclear spins will align themselves with the applied field...
either in a low energy state parallel to the field, or a high energy state anti-parallel to the field. Moreover, they will precess around the direction of the magnetic field with a frequency proportional to the field strength, as described by the Larmor equation:

$$\omega = \gamma B_0$$

where $\gamma/2\pi$ is the gyromagnetic ratio and $B_0$ the external magnetic field, which in our case is the main magnetic field generated by the MRI system. The gyromagnetic ratio is unique for each atom. For example, the gyromagnetic ratio of hydrogen is 42.58 MHz/T, generating a precessional frequency of approximately 64 MHz at 1.5 T. Externally applied RF energy that matches the Larmor precessional frequency will cause some of the protons in the low energy state to flip to the high energy state. For protons at field strengths used for CMR, the Larmor frequency is in the “very high frequency” or VHF range of radio frequencies commonly used for FM radio and broadcast television. Radiation in this frequency range is non-ionizing, contributing to the inherent safety of MRI in comparison to radiographic techniques. Energy is proportional to frequency, and the RF energy used in MRI is many orders of magnitude lower than X-ray radiation, and is not known to cause the increased risk of DNA damage that is observed with the use of X-rays. Despite its lower energy, the MRI signal is detectable because there are so many hydrogen nuclei available in the body to contribute to the signal.

**The Boltzmann distribution**

From the quantum mechanical point of view, hydrogen spins are found in either one of the two available energy states. However, there are only slightly more spins in the low energy spin state compared to the high energy state, and the excess spin number is, according to Boltzmann equilibrium probability, directly proportional to the total number of spins in the sample and the energy difference between states. The relation between the number of spins in high energy state ($N^-$) and the number of spins in the lower energy state ($N^+$) is given by the expression:

$$N^-/N^+ = e^{-E/kT}$$

where $E$ is the energy difference between the spin states; $k$ is Boltzmann’s constant, $1.3805 \times 10^{-23}$ J/K; and $T$ is the temperature in Kelvin. The energy of a proton, $E$, is directly proportional to its Larmor frequency $\nu$ (in Hz), such that $E = h\nu$, where $h$ is Plank’s constant ($h = 6.626 \times 10^{-34}$ J s). Following the Larmor equation (above), this results in a direct relationship between proton energy $E$, and magnetic field $B_0$, $E = h\gamma B_0$. When the energy delivered to the system matches the energy difference
between the spin states, spins from the stable lower energy states jump up to the unstable higher energy states. As these spins fall back into the lower energy state, they emit a detectable signal; this is the resonance phenomenon.

Only those excess spins in the low energy state are available for excitation, and able to generate MRI signal when they return to equilibrium position. There are only approximately nine more spins in the low energy state compared to the high energy state for each 2 million spins at 1.5 T field strength. Given that each ml of water contains nearly $10^{23}$ hydrogen atoms, the Boltzman distribution predicts over $10^{17}$ spins contributing to the MRI signal in each ml of water. As the magnetic field strength increases, the number of excess spins in low versus high energy state increases, and with it the magnitude of the MRI signal. The larger number of excitable spins leads to improvement in image quality (signal-to-noise ratio [SNR] and/or resolution) and is the driving force for imaging at higher magnetic field strengths like 3 T and 7 T.

**The RF pulse and signal reception**

When a specimen or subject is placed in the high magnetic field of an MRI system, the number of spins in the low energy level exceeds those at the high energy level (Figure 1.1b) creating a net magnetization aligned in the direction of $B_0$. Externally applied radio frequency (RF) energy with a frequency that matches the precessional or Larmor frequency of the spins will cause some of the protons in the low energy state to jump up to the high energy level. In terms of the net magnetization, the magnetic field component of the RF wave, $B_1$, which is perpendicular to the direction of $B_0$ and of μT order of magnitude, will tilt the longitudinal magnetization ($M_z$) to an angle that depends on the strength of the applied $B_1$ field and the duration of the RF pulse, which is usually from one to several milliseconds. A 90° RF pulse will rotate the net magnetization totally from the longitudinal plan (z) into the transverse plane (xy). It is this transverse component of the net magnetization that generates the MR signal detectable by a receiver coil. The MR signal is captured in the form of an induced voltage in a receiver antenna, or “coil”, placed perpendicular on the transverse plane. The precession of the transverse component of the magnetization, $M_{xy}$, generates an oscillating current in the receiver coil according to Faraday’s law of induction.

In summary, signal generation in MRI follows a few basic steps. As spins are subjected to the strong magnetic field, the net magnetization aligns with the direction of the applied field in the longitudinal (z) direction. The RF pulse with a frequency that matches the precessional frequency of the protons tilts the net magnetization from the longitudinal to the transverse plane (xy). Afterwards, the precession of spins around the axis of the main magnetic field induces a “resonant” signal in a receiver coil placed perpendicular to the transverse plane (Figure 1.2).

**Relaxation**

Through application of the RF pulse, which provides the energy necessary for spins to jump from the low to the high energy level, the protons are raised up to an excited, unstable state. While the magnitude of the MR signal depends on the net magnetization $M_{xy}$, the duration of the induced voltage is a function of the relaxation time constants $T_1$ and $T_2$, and/or $T_2^*$ of the sample. These relaxation parameters are different for different tissues and pathologies, and as such are primary sources of image contrast in CMR.

**$T_2$ and $T_2^*$ relaxation**

A RF pulse that tilts the net magnetization into the transverse plane also brings the spins into phase coherence with each other, resulting in a maximum current in the receiver antenna. As time passes, the spins that were initially precessing in phase with each other will lose phase coherence, resulting in a decrease in the net magnetization (Figure 1.2) and induced voltage. The loss of phase coherence is called transverse or spin–spin relaxation and is characterized by the $T_2$ time constant. The rate of loss in phase coherence among the individual protons is influenced by the chemical environment experienced by each. The presence of each spin slightly affects the local magnetic field, and as such the precessional frequency of the surrounding spins. Due to this spin–spin interaction protons will lose phase coherence and the transverse
magnetization \(M_y\) will decay exponentially from \(M_z\) at a rate defined as \(T_2\), described by the relationship:

\[
M_y(t) = M_y(0) \exp(-t/T_2)
\]

The transverse relaxation is highly dependent on the molecular structure of the sample. Small molecules in amorphous medium demonstrate a long \(T_2\) because fast and rapidly moving spins average out the intrinsic magnetic field inhomogeneities. Conversely, larger macromolecules that are subject to constrained molecular motion, exhibit much shorter \(T_2\) due to the accumulation of phase differences among spins, which are not canceled by rapid diffusion.

The spin–spin interaction is not the only factor responsible for the time-decay of transverse magnetization and acquired MRI signal. Extrinsic magnetic inhomogeneities, such as imperfections of the main magnetic field or susceptibility differences between adjacent tissues, also contribute to dephasing and loss of phase coherence among spins. The time-decay of signal in this case is characterized by the time constant \(T_2^*\), which is always shorter than \(T_2\). However, the \(T_2^*\) signal loss caused by static, extrinsic magnetic field inhomogeneities is corrected in spin echo sequences by the use of 180° RF refocusing pulses. RF refocusing can reverse the phase difference induced by static field inhomogeneity and re-establish phase coherence.

**T1 relaxation**

The end of the RF pulse begins the return to equilibrium. Immediately after the RF pulse, the excited spins will undergo relaxation through the same energy coupling process. The return of excited spins to the low energy, equilibrium state, which is accompanied by the recovery of the longitudinal magnetization \(M_z\), is part of the spin-lattice relaxation process. The rate of \(M_z\) recovery is a function of the relaxation time constant \(T_1\), which by definition, is the time necessary to recover 63% of the equilibrium magnetization \(M_z\) after a 90° RF pulse:

\[
M_z(t) = M_z(1 - \exp(-t/T_1))
\]

The return to equilibrium is directly related to how fast the excited spins release their energy to the tissue (lattice). This process depends significantly on the physical properties of the tissue. The energy transfer is possible only when the precessional frequency of spins overlaps the vibrational frequencies of the molecules embedded into the lattice. Depending on their physical characteristics (size) the vibrational frequency of the molecules span different frequency ranges. The less efficient this system is at transfer of energy from the excited spins to the lattice, the longer the T1 recovery time will be.

Moreover, T1 relaxation is dependent on the main magnetic field strength. At higher magnetic field the precessional frequencies of spins increase,
and as such there is lower spectral overlap with the molecular vibrational frequency spectrum of the sample, resulting in an increase in spin lattice relaxation time with $B_0$. The only exception from this rule is offered by free water, which has a vibrational frequency range that covers a large spectrum of precessional frequencies. However, for a specific tissue, there is always the following relationship among relaxation times, $T_1 > T_2 > T_2^*$, regardless of the magnetic field strength.

**Contrast agents**

Tissue contrast in MRI is fundamentally based on tissue specific parameters such as proton density (PD) and relaxation times $T_1$, $T_2$ and $T_2^*$ and can be further influenced by diffusion, perfusion, flow and motion. Contrast agents offer another important source of tissue contrast essential to many cardiac imaging techniques. Contrast agents generally work by shortening both $T_1$ and $T_2$, but with a predominant effect on either one or the other depending on the specific agent. $T_1$ shortening contrast agents enhance the MR signal (positive contrast) by increasing the signal in $T_1$ weighted images. The reverse is true for a predominantly $T_2$ contrast agent; shortened $T_2$ leads to decreased signal (negative contrast) in $T_2$ weighted images.

While the underlying processes by which contrast agents function is complex, their effects on $T_1$ and $T_2$ can generally be described in a simplified way by the equations:

$$\frac{1}{T_1} = \frac{1}{T_{1_0}} + r_1 C$$

$$\frac{1}{T_2} = \frac{1}{T_{2_0}} + r_2 C$$

where $T_1$ and $T_2$ are the tissue relaxation times after contrast agent administration, $T_{1_0}$ and $T_{2_0}$ are the relaxation times prior to contrast agent injection, $C$ is the contrast agent concentration and $r_1$ and $r_2$ are the longitudinal and transverse relaxivities of the contrast agent. However, $r_1$ and $r_2$ are field strength dependent and the linear relationship between relaxation time shortening and contrast agent concentration is no longer valid at high concentrations.

Most contrast agents used for clinical CMR are the paramagnetic chelates of gadolinium (Gd$^{3+}$). Gadolinium has unpaired orbital electron spins and a very large magnetic moment. Gadolinium shortens the $T_1$ relaxation time by allowing free protons to become bound and to create a hydration layer, which helps energy release from excited spins and accelerates the return to equilibrium magnetization. A number of CMR applications are dependent on exogenous contrast agents, including angiography, first-pass perfusion, late gadolinium enhancement (LGE), and characterization of tumors and masses.

**Image encoding**

**Magnetic field gradients**

The Larmor equation is at the heart of image encoding. The main magnetic field, $B_0$, generated by the MRI system is engineered to be as homogeneous as possible. Homogeneity of about 1 part per million over a roughly spherical region of $\frac{1}{2}$ meter in diameter is typically achieved, depending on the particular magnet design. Within this homogeneous volume, all protons precess at the same frequency (disregarding tissue susceptibility differences and other sources of field distortion). By precisely controlling the strength of the magnetic field as a function of both location and time, the frequency and phase of precession also become functions of location and time. Using this principle, the MR signals coming from different locations within the body can be distinguished from one another, and an image can be formed. Special gradient coils are embedded within the bore of the MRI system to create controlled, linear variations in the $B_0$ field strength in each of the three orthogonal directions in the Cartesian $(x,y,z)$ spatial coordinate system (Figure 1.3). By applying current to these coils simultaneously in appropriate ratios, a linear gradient in the magnetic field can be generated in any arbitrary direction. This linear change in magnetic field translates into a linear change in resonant frequency depending on location in that direction.

**Slice selection**

In order to generate a magnetic resonance signal that can be detected, the magnetization must be
tipped away from the longitudinal axis and into the transverse plane by RF excitation, as described earlier. The process of slice selection limits RF excitation to a plane of tissue of any desired thickness. Recall that a spatially linear gradient or ramp in the magnetic field establishes a linear relationship between proton precessional frequency and location. In order to excite or tip the magnetization of precessing spins, an RF pulse must oscillate at the precessional frequency of those spins. Physically, RF pulses are of finite amplitude, duration, and bandwidth. The amplitude and duration of the RF pulse will control the resulting flip angle; the longer the pulse and higher the amplitude, the greater the tip angle of the net magnetization. The RF pulse center frequency can be shifted to match a specific location along the gradient, and the bandwidth of the pulse can be limited to selectively excite the protons with a narrow range of frequencies around the center frequency, as shown in Figure 1.4. Thus a slice of arbitrary thickness and location along the direction of the slice select gradient can be selectively excited to generate the signal used to form the MR image. The direction of the slice selection gradient, and therefore the orientation of the slice, can also be arbitrarily chosen by appropriate combination of the physical x, y, and z gradient fields. Following slice selective excitation, the signal detected by the MRI receiver coil will come from the excited slice only. The amplitude of the signal emitted by the slice is directly proportional to its thickness; this sets the lower practical limit on slice thickness at about 2 mm. Thinner slices can be achieved by 3D encoding, which is addressed in the section on phase encoding.

**Frequency encoding**

The process of slice selection excites the slice or slab of tissue that will generate the MR signal; the next steps of frequency and phase encoding serve to encode this tissue into individual discrete two-dimensional picture elements (pixels), or three-dimensional volume elements (voxels). Once again, linear field gradients and the Larmor relationship between field strength and precessional frequency are used to encode spatial location information into the MRI signal. After a slice-selective RF pulse tips the magnetization into the transverse plane, an MR signal is emitted by all of the spins contained within the slice and some method of encoding is required to distinguish the signals coming from the individual voxels. A linear magnetic field gradient is switched on in one of the in-plane directions, perpendicular to the slice select gradient. This gradient has the effect of frequency encoding. While this gradient is on, preces-
sional frequency has a linear distribution along the gradient direction, and thus every location along the gradient can be distinguished by the frequency of the signal it emits. The MR signal is detected through the receiver coils and digitally sampled using an analog-to-digital converter (ADC) during the application of a constant frequency encoding gradient. This detected signal is the sum of all of these frequency components. Fourier transformation is used to separate out the individual frequency components in the detected signal, and thus decode the signal from the entire slice into individual signals coming from discrete locations along the frequency encoding gradient.

Frequency encoding can also be described in terms of spatial frequency, and this alternative description is also helpful to understand phase-encoding, the method used to encode the other in-plane dimension of the image. Spatial frequency expressed as cycles/cm is directly analogous to the perhaps more familiar concept of temporal frequency expressed in units of cycles/sec or Herz. Whereas temporal frequency pertains to a time varying signal, spatial frequency can be used to describe a signal varying with position, for example, an image. An individual spatial frequency then describes a sinusoidal variation in pixel intensity across an image. A complex image can be expressed as the linear combination of many spatial frequencies. Lower spatial frequencies determine the gross features and contrast in the image, while higher spatial frequencies determine image details and sharpness. The Fourier transform can be used to go back and forth between image space and spatial frequency, or “k-space”, in the same manner it is used to describe the frequency component of a time-varying signal.

Before the frequency encode gradient is switched on, all of the spins are precessing at the same frequency and in phase with each other. As soon as the frequency gradient is switched on, the spins begin to precess at a frequency linearly dependent on position, and this linear distribution of frequencies causes a sinusoidal distribution of phase across the slice in the direction of the frequency encoding gradient (Figure 1.5). It is the integral or
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The accumulated area under the frequency encoding gradient pulse that determines the instantaneous sinusoidal distribution of phase. This sinusoid across space describes a single spatial frequency, often referred to as $k_x$. As time progresses while the gradient remains on, the area under the frequency encoding gradient increases and progressively higher spatial frequencies are mapped out in the distribution of phase. Thus, each digital sample of the MR signal corresponds to the signal attributable to a unique spatial frequency component in the image, that is, each sample corresponds with a distinct value of $k_x$.

**Phase encoding**

The third spatial dimension (second in-plane dimension) must also be encoded in order to distinguish the signal from each individual voxel and complete the process of image formation. Phase encoding is also based on the Larmor equation, and on discrete sampling of the spatial frequency content of the image. Frequency encoding, as described earlier, samples spatial frequency components in rapid succession as the area under the frequency encoding gradient pulse accumulates over time. Phase encoding, however, is instead typically accomplished by applying a series of gradient pulses of successively increasing amplitude, each designed to encode a single specific spatial frequency component, $k_y$, of the image (Figure 1.6). The phase encode gradient pulse amplitude is incremented to encode a different spatial frequency component $k_y$, prior to each frequency encoding gradient. This completes the concept of two-dimensional spatial frequency encoding. The matrix of sampled image data represents the two-dimensional spatial frequency content of the image and is often referred to as $k$-space, and the process of image data acquisition can be thought of as filling of $k$-space. Each phase encoded line of data corresponds to a specific spatial frequency in the phase encoded direction, and contains all spatial frequencies in the frequency encoding direction. With the sampling of each MR signal, all values of $k_y$ are encoded for a single value of $k_x$. Thus, each phase-encoded line of data corresponds to a raster line in $k$-space. The two-dimensional Fourier transform is utilized to convert this spatial frequency information into the image domain (Figure 1.7). Phase-encoding can also be applied in the slice direction to encode thinner sections of tissue than possible using selective excitation alone. 3D data acquisition incorporates the process of phase encoding into the slice direction as well as one in-plane direction.

**Basic pulse sequences**

The pulse sequence defines the sequence of events on a microsecond scale controlling all factors
understanding of pulse sequences can be gained by looking at the five basic components of any CMR pulse sequence: magnetization preparation, echo formation, k-space trajectory, k-space segmentation, and image reconstruction. Some of the options used in CMR within each of these categories are listed in Table 1.1.

**Magnetization preparation**

Magnetization preparation refers to the various methods available to impart sensitivity of the pulse sequence to specific characteristics of the tissue within the imaged slice. For example, the inversion

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**Figure 1.6** Encoding of spatial frequencies in the phase-encode direction is analogous to frequency encoding, but rather than a constant amplitude gradient waveform incrementing in time, individual gradient pulses of incrementing amplitude are applied. Once again, the encoded spatial frequency is directly related to the area under the gradient pulse. The highest spatial frequency is encoded by the highest phase encode gradient pulse \((G_{PE1})\), and progressively smaller amplitude phase encode pulses encode lower spatial frequencies. With no phase encode gradient pulse \((G_{PE3})\), all spins are in-phase. This is called the “zero phase-encode line” or “center phase-encode line.”

**Figure 1.7** The 2D Inverse Fourier transform of the raw k-space data (a), obtained by means of phase and frequency encoding, is used to reconstruct this MR image of a four chamber view of the heart (b).
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For delayed-enhancement imaging post-contrast, since time is available in between IR pulses to allow for signal recovery, the SR pulse is more appropriate for perfusion imaging when multiple slices are prepared and imaged each cardiac cycle. The double-inversion or black-blood preparation is commonly used to suppress the blood signal [1,2]. It works by effectively inverting the blood outside of the imaged slice, without effecting the magnetization within the imaged slice. When the inverted blood flows into the slice, the pulse sequence can recover (IR) technique precedes data acquisition with a 180° RF pulse to provide high sensitivity to differences in T1 (see Figure 1.8). Saturation recovery (SR) preparation using a 90° RF pulse is commonly used in first-pass perfusion imaging. It provides moderate T1-weighting that is not as strong as the IR technique, but does not require a wait period in between pulses to allow magnetization to recover since longitudinal magnetization is essentially set back to zero with each saturation pulse. While IR imaging has been very successful for delayed-enhancement imaging post-contrast, since time is available in between IR pulses to allow for signal recovery, the SR pulse is more appropriate for perfusion imaging when multiple slices are prepared and imaged each cardiac cycle. The double-inversion or black-blood preparation is commonly used to suppress the blood signal [1,2]. It works by effectively inverting the blood outside of the imaged slice, without effecting the magnetization within the imaged slice. When the inverted blood flows into the slice, the pulse sequence can

Table 1.1 CMR pulse sequences can be broken down into the components listed in the five columns of the table. Each application involves a different combination of these components to achieve specific imaging goals.

<table>
<thead>
<tr>
<th>CMR pulse sequences</th>
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<tbody>
<tr>
<td><strong>Magnetization preparation</strong></td>
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<td>• Inversion recovery</td>
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<td>• Saturation recovery</td>
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<td>• Double inversion recovery (black blood)</td>
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<td>• Fat suppression</td>
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<td>• T2-preparation</td>
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<td>• Diffusion weighting</td>
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<td>• Tagging</td>
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<td>• DENSE</td>
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<td>• Magnetization transfer</td>
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<tr>
<td>• Velocity encoding</td>
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<td>• T1, T2, T2* mapping</td>
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<td><strong>Echo formation</strong></td>
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<td>• Spin echo</td>
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<td>• Turbo Spin Echo</td>
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<td>• SPACE</td>
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<td>• Gradient Echo</td>
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<td>• Steady-State Free Precession (SSFP)</td>
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<td>• Echo Planar Imaging (EPI)</td>
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<td>• Spin echo EPI</td>
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<td>• Gradient echo EPI</td>
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<td><strong>k-space trajectory</strong></td>
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<td>• Linear</td>
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<td>• Centric</td>
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<td>• PROPELLER</td>
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<td><strong>k-space segmentation</strong></td>
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<td>• Segmented</td>
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<td>• Single-shot</td>
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<td><strong>Image reconstruction</strong></td>
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<td>• Partial Fourier</td>
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<td>• Parallel Imaging</td>
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<td>• SENSE</td>
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<td>• SMASH</td>
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<td>• GRAPPA</td>
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<td>• TSENSE/TGRAPPA</td>
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<td>• k-t methods</td>
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Figure 1.8 The inversion (a) and saturation (b) recovery magnetization curves of fat, myocardium and blood. Observe the increased magnetization difference, resulting in better T1 contrast between different tissue types when inversion recovery sequence is used as compared with saturation recovery technique.
be timed to acquire data just as the blood signal is crossing through the zero or null point in its T1 recovery curve. Velocity encoding can also be considered as a magnetization preparation, although unlike the others listed in the table it incorporates specific gradient pulse design to control the motion and flow sensitivity of the pulse sequence rather than a separate preparation module [3].

**Echo formation**

A few basic methods of echo formation are employed in CMR, each with distinct advantages and disadvantages that have led to specific applications for each. Spin Echo, including Turbo- or Fast Spin Echo and SPACE [4], is used primarily as a black-blood method for cardiovascular morphology and tissue characterization based on T1 or T2 changes. Gradient Echo is commonly used for delayed-enhancement and first-pass perfusion, in combination with IR and SR prep pulses, respectively. Gradient echo cine is used for phase velocity mapping, tagging, and in some cases for visualization of valve function. Most cine imaging applications, however utilize steady-state free precession [5] (SSFP or TrueFISP), due to its inherently high blood-to-myocardium contrast, high signal-to-noise ratio, and high imaging efficiency [6]. Echo Planar Imaging (EPI) has found widespread application as a method for perfusion imaging due its high efficiency [7].

**k-space trajectory**

CMR applications are dominated by Cartesian k-space sampling, that is, a linear raster scanning of k-space accomplished by conventional phase encoding each line of data. Virtually all CMR sequences in broad clinical use employ this conventional sampling strategy. Some alternative trajectories that have been employed in CMR are shown in Figure 1.9. The spiral trajectory has some advantages in speed and insensitivity to motion, and has been utilized for coronary artery imaging [8]; however it is highly sensitive to field inhomogeneities, and has not seen widespread application for that reason. Radial imaging has some efficiency advantages over Cartesian sampling, and is experiencing some gain in popularity for cine imaging [9] due to its ability to achieve higher spatial resolution for a given number of acquired lines. PROPELLER [10] combines some of the advantages of Cartesian and radial acquisitions, and while it is popular for neuro-imaging, CMR applications have not yet been fully developed.

**k-space segmentation**

Segmentation refers to the strategy of segmenting data acquisition over multiple cardiac cycles [11]. The degree of segmentation can range from one k-space line per cycle (non-segmented), up to acquisition of all of the lines needed to reconstruct an image (single-shot). Any level of segmentation can be used with each of the basic methods of echo formation (gradient echo, spin echo, SSFP, EPI). Overall acquisition time is inversely related to the number of lines acquired per image each cardiac cycle (lines per segment), that is, the more lines per segment, the shorter the scan time. The trade-off is in temporal resolution; the more lines per segment, the poorer the temporal resolution. Modern CMR sequences for cine, flow, and delayed-enhancement are designed to acquire enough lines per segment to reduce the scan time to a reasonable breath-hold. The success of segmented imaging depends not only on patient breath-hold, but also on a
regular cardiac rhythm to ensure that the k-space data from each cardiac cycle is capturing the heart in the same respiratory and cardiac positions. In patients with severe arrhythmia or an inability to breath-hold, real-time or single-shot methods are commonly used due to their insensitivity to respiratory motion effects.

**Image reconstruction**

Partial Fourier or partial k-space acquisition has been used for a number of years as a means of reducing scan time at the expense of signal-to-noise [12]. More recent advances in image reconstruction methods have played a large part in the improved image quality and efficiency of CMR. Parallel Acquisition Techniques (SENSE [13], SMASH [14], GRAPPA [15], and TSENSE [16]) have become an integral part of virtually all commonly applied CMR pulse sequences. These methods allow reconstruction of full field-of-view and full resolution images while sampling only a fraction of the full k-space data matrix. This results in a significant time savings that can be directly beneficial as shortened scan time, or traded for higher spatial or temporal resolution. This entails a direct trade-off of signal-to-noise ratio, so the acceleration attainable using parallel imaging is generally limited to a factor of two or three, but that can make the difference between a 10 second or a 20 second breath-hold, and so represents a significant gain in imaging performance.

Table 1.1 lists some of the many possibilities for each of these pulse sequence components. Elements from each column of the table can be combined to construct pulse sequence variations for specific applications. For example, double inversion recovery, turbo spin echo, and segmented, linear, Cartesian k-space trajectory with GRAPPA parallel imaging reconstruction is a common application for T2-weighted imaging of the heart. Or, saturation recovery, gradient echo EPI with centric, Cartesian, single-shot trajectory and TSENSE reconstruction [17] is commonly used for first-pass perfusion imaging. Delayed-enhancement imaging is commonly performed using inversion recovery, gradient echo, with linear, Cartesian, segmented k-space acquisition [18]. The list of possibilities goes on, and while there is not space in this chapter to delve into the details of the numerous combinations in common use in CMR, hopefully Table 1.1 helps to illustrate the wide range of sequence combinations available, and even some possibilities that have not yet been investigated. Most modern MR systems have very flexible interfaces and pulse sequence control software that allow the user to easily mix and match components from these categories. Unfortunately, for every useful combination there are many more that have no value, contributing to the complexity of CMR and the need for every CMR practitioner to gain some basic understanding of the underlying physics.

**Cardiac and respiratory synchronization**

**ECG triggering and gating**

In addition to the categories outlined in Table 1.1, CMR sequences can be further subdivided with respect to depiction of cardiac motion: dynamic or static. Dynamic methods include any cine techniques designed to represent the heart or flow patterns at multiple phases throughout the cardiac cycle. Static methods generate images of the heart at a single-phase of the cardiac cycle. Static techniques are generally applied to depict cardiovascular anatomy, or to characterize tissue by generating images sensitive to any of a variety of contrast mechanisms. In either case, synchronization with cardiac motion, generally using an ECG signal, is necessary to time each image to a specific phase of the cardiac cycle. This is not the case in real-time cine methods that acquire dynamic images asynchronous with the cardiac cycle. Some sequences, like first-pass perfusion and dynamic 3D MR angiography, can be thought of as hybrids between dynamic and static imaging. These techniques create a dynamic series of images, but each image depicts a different cardiac cycle, not a different phase of the cardiac cycle.

The R-wave of the ECG is typically used to generate a trigger signal indicating the beginning of the cardiac cycle with the initiation of ventricular systole. CMR pulse sequence events are timed relative to that trigger to acquire data at specific time points within the cardiac cycle. Static images are acquired using prospective triggering. That is, one or more lines of k-space data are acquired beginning at a specific time-delay relative to the trigger.
pulse. Dynamic cine images can be acquired by either prospective triggering or retrospective gating. With prospective triggering, phases of the cardiac cycle are defined by a fixed time after the R-wave, regardless of the duration of each individual cardiac cycle. With retrospective gating, each cardiac phase is defined as a certain percentage of the cardiac cycle, allowing the actual duration of each phase to vary flexibly with variation in cardiac cycle.

**Respiratory motion compensation**

Respiration causes variation in the position of the heart from beat to beat, and leads to motion artifact in segmented acquisitions scanned over multiple cardiac cycles. There are four basic strategies to deal with respiratory motion artifact; signal averaging, breath-holding, respiratory gating, and single-shot imaging. In many common CMR applications like cine, velocity mapping, late gadolinium enhancement, and black-blood imaging, segmented acquisitions that are fast enough to be performed in a reasonably short breath-hold are widely available and commonly used as the standard method. In small children unable to voluntarily breath-hold, signal averaging is successfully used to average out respiratory motion artifact. Advances in gradient hardware and parallel acquisition techniques have dramatically improved the speed and quality of single-shot and real-time imaging techniques, and these now often become the method of choice for scanning patients unable to breath-hold. Applications like coronary angiography and other three-dimensional acquisitions that require high resolution and acquisition times are unsuitable for breath-hold or single-shot methods; in these circumstances respiratory gating is useful, most commonly in the form of navigator echo gating [19].

**MRI hardware**

The main field-generating components of the MRI system include the main magnet ($B_0$ or $B_z$ field), the RF transmitter coil ($B_1$ field), and the gradient coils ($G_x$, $G_y$, $G_z$ fields). Additional second-order shim coils are also often employed to achieve a more homogeneous $B_0$ field. Separate computer systems are typically used to control the MRI field-generating units (measurement-control system), reconstruct the acquired data, and provide an interactive interface to link the user to the MRI system control and the reconstructed images.

Advances in hardware continue to drive advances in pulse sequences. This is especially true in CMR where acquisition speed is critical to avoid the deleterious effects of motion. Gradient amplitude and switching speeds have hit the limits of physiological stimulation. RF receiver systems continue to advance in the number of channels used to increase the capacity of parallel imaging techniques. Multi-channel array coils designed specifically for cardiovascular applications can improve SNR and parallel imaging performance.

**References**

PART I The basics of cardiac MR