

**PART I**

# **General Considerations and Fundamentals of Imaging**



# Principles of CT and MRI

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X-ray computed tomographic (CT) imaging and magnetic resonance imaging (MRI) are relatively new methods for imaging the cardiovascular system. Although both methods can be used to produce high-quality tomographic (cross-sectional) images of the heart and vascular system, and there are some parallels in the underlying mathematics of the image reconstruction used for the two imaging methods, the basic physical principles of the two methods are quite different. In addition, they have different relative strengths and weaknesses. In this chapter, we will briefly review the basic concepts and principles of cardiovascular imaging with CT and MRI as a background for the more specialized chapters to follow.

## Cardiac CT Basics

Cardiac CT imaging, predominantly involving coronary computed tomographic angiography (CTA) imaging, has emerged in importance with the development of multidetector CT scanning technology and slip-ring technology [1]. These technologies allow noninvasive imaging of the body in greater detail with faster speed and higher temporal resolution. For cardiac imaging, this timing is critical because the anatomic structures of interest in the heart are very small and the heart is constantly moving. This section will review the technological advances that have brought cardiac imaging into prominence as an important tool in the evaluation of cardiac disease.

4 Part I General Considerations and Fundamentals of Imaging

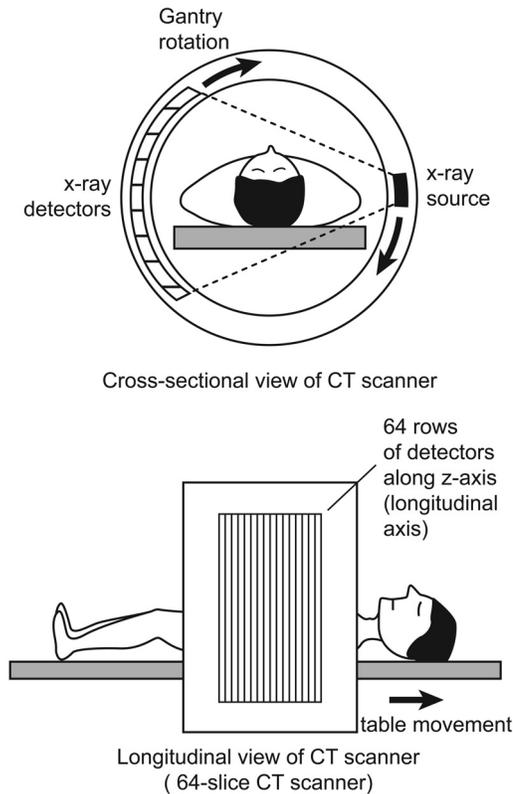


Fig. 1.1 Schematic of the principles of cardiac CT image acquisition.

### Physical Principles of CT Imaging

At present, a 64-slice multidetector CT scanner is the standard configuration used in cardiac CT examinations [1]. Newer developments, such as the dual-source CT scanner and the volume CT scanner, have been recently introduced and will likely supplant 64-slice multidetector CT as the future standard (see also discussion in Chapter 21). Nevertheless, this section will review 64-slice multidetector CT technology because it is the most commonly used device today. This configuration involves a rotating gantry that houses an x-ray source and multiple rows of x-ray detectors, located opposite each other (Figure 1.1). For a 64-slice multidetector CT scanner, x-ray transmission data are acquired simultaneously at 64 levels along the bore of the gantry, or z-axis. As the patient passes along the axis of the scanner on a moving table, the gantry rotates around the patient. While the gantry rotates, x-ray beams pass through the patient and data are collected at the detectors regarding the amount of attenuation of the x-ray beam that occurs along each projection. Because the patient and gantry are both

moving, the projection data are collected in a helical or spiral manner to cover the volume of interest. From this projection data, mathematical algorithms are used to reconstruct cross-sectional images of the body based on attenuation values of the different tissues. The *pitch* is defined as the table travel per complete rotation of the gantry divided by the x-ray beam width. Thus, a pitch between zero and one implies overlap of the projections, whereas a pitch greater than one implies gaps in the projections. The pitch is generally selected prior to the scan and based on the patient's heart rate. For faster heart rates, a higher pitch is applied. However, if the pitch is too high for a given heart rate, there will be gaps in the data set, decreasing image quality. As discussed in Chapter 3, if the pitch is too low for a given heart rate, the scan time, and consequently breath hold time, is unnecessarily increased along with the radiation dose.

The spatial resolution of the images is determined by the image matrix size, the detector height and geometry, and the reconstruction algorithm. Because a volumetric data set is acquired, images can be reconstructed at different slice thicknesses. The lower limit of slice thickness will correspond to the height of a single row of detectors. Thicker slices can be reconstructed by combining thinner slices. The average current 64-slice multidetector CT scanner has detector heights ranging from 0.5 to 1.25 mm. The average spatial resolution is approximately  $0.4 \times 0.4 \times 0.4$  mm.

The temporal resolution of the images is determined by the gantry rotation speed and the reconstruction algorithm. For example, a partial scan reconstruction algorithm can decrease the temporal resolution to half the gantry rotation time. This technique uses  $180^\circ$  of data in parallel geometry to reconstruct the whole image. Another algorithm called multisegment reconstruction can decrease the temporal resolution by using data from more than one cardiac cycle to reconstruct the images. The resulting temporal resolution will depend on the heart rate. The average current 64-slice multidetector CT scanner has gantry rotation times ranging from 0.33 to 0.42 sec. The average temporal resolution ranges from 83 to 210 msec.

### Image Reconstruction Principles

Electrocardiogram (ECG)-gated data acquisition for cardiac CT examinations is necessary to reduce cardiac motion artifact and improve temporal resolution [2,3]. There are two primary methods of ECG gating. One method is prospective ECG triggering, in which data are acquired at a predefined point during the cardiac cycle, also known as a "step-and-shoot" technique. In this method, sequential axial slices are acquired through the heart. As discussed in Chapter 3, the advantages of this method include speed and a lower radiation dose. However, one disadvantage of this method is the dependency upon a regular heart rate. Variability in the heart rate will result in data acquisition at different points along the cardiac cycle, producing misregistration artifacts. Another disadvantage is the lower z-axis resolution, secondary to the slice-by-slice acquisition rather than a volumetric acquisition.

## 6 Part I General Considerations and Fundamentals of Imaging

The second method is retrospective ECG gating, in which data are acquired throughout the entire cardiac cycle. In this method, data acquired at specified points during the cardiac cycle are extracted to reconstruct the images. Two types of reconstruction algorithms can be applied. In one type, data can be extracted at a specified fraction of the R–R interval. For instance, image reconstruction can be performed at 10% intervals of the R–R interval from 0% to 90%. In another type, an absolute time interval before or after the R peak can be used, rather than a fraction, to reconstruct the images. In this manner, multiple image data sets are produced that allow the viewing of the cardiac anatomy during all phases of the cardiac cycle [2]. In addition, functional analysis can be performed by visualizing the data sets of the beating heart in a dynamic or cine form to evaluate systolic and diastolic function [2]. One advantage of the retrospective ECG gating method is the ability to edit the ECG reference points to adjust data acquired during an irregular or ectopic beat. Another advantage is the isotropic resolution of the data set, secondary to the volumetric data acquired. The disadvantage of this method is a higher radiation dose. However, there are strategies, such as ECG-triggered dose modulation, that can be used to reduce the total dose. By reducing the tube output during systole, when there is more cardiac motion and anatomical imaging is less useful, the total dose can be reduced without sacrificing data needed for functional analysis. Radiation dose considerations and strategies to reduce it are discussed in more detail in Chapter 3.

### Image Contrast and Contrast Agents

Most cardiac CT examinations will require administering an intravascular contrast agent (“contrast”) to perform a complete examination. Contrast administration serves several functions. First, it allows identification of the vessels and delineates their endoluminal anatomy [4]. In coronary CTA examinations, contrast opacification of the lumen is required to determine the presence and degree of luminal stenosis secondary to atherosclerotic plaque. Without the presence of contrast, the blood pool, vessel wall, and noncalcified atherosclerotic plaque will have similar attenuation values, preventing accurate assessment of the vessel lumen. Contrast also opacifies the cardiac chambers, allowing identification of space-occupying lesions such as intraluminal masses or thrombus. Opacification of the lumen also allows functional analysis of the ventricles. The ventricular volumes can be determined in both systole and diastole to calculate an ejection fraction. Contrast also perfuses the soft tissues, demonstrating their enhancement patterns. For example, the enhancement pattern of the myocardium can assist differentiation of normal myocardium from scarred myocardium, which might occur following a myocardial infarct. Thus, contrast administration is an important element of the cardiac CT examination.

There are several contrast agents available for cardiac CT examinations. These contrast agents are water-soluble and contain high concentrations of iodine. The

high x-ray absorption by the contrast agent attenuates the x-rays from the CT scanner to a greater degree than soft tissues but less than bone. On a CT image, the contrast will appear as a shade of gray depending upon its concentration. The radiodensities of structures in a CT image are measured in Hounsfield units, where water is set to the value zero. Soft tissues have Hounsfield values of approximately 40 to 80, and bone has a Hounsfield value of approximately 1000. The contrast within the vessel lumen has values ranging from 100 to 400, depending upon concentration and volume. For a more detailed discussion on the clinical use of contrast agents in cardiac CT, please refer to Chapter 2.

### **Imaging Protocol**

Patient preparation is an essential step in any imaging protocol. Patients are instructed to remain NPO for three hours prior to the examination and refrain from caffeine intake. After obtaining patient history and informed consent, vital signs are recorded. If necessary, beta-blockers are administered to lower the heart rate. Beta-blockers may be relatively contraindicated in patients with asthma, aortic stenosis, heart block, or severe left ventricular dysfunction. After achieving an adequate heart rate, the patient is placed supine on the CT table. Patients must be able to lie on their back for the duration of the examination, which typically ranges from 5 to 20 min. For a coronary CTA examination, a large-bore intravenous catheter is inserted into a large arm vein, preferably an antecubital vein. This large-bore catheter is necessary to accommodate the high rate of contrast infusion, ranging from 2 to 6 cc/sec. ECG leads are placed on the patient's chest to acquire an ECG tracing. The decision to perform prospective or retrospective ECG triggering gating should be made after considerations of the advantages and disadvantages of each method discussed previously and the type of CT scanner available. To facilitate the visualization of small-caliber coronary arteries, nitroglycerin is generally administered at the start of the examination if there are no contraindications [4]. After obtaining a scout image of the thorax to identify anatomic landmarks, a coronary calcium scan (i.e., without IV contrast) may be performed to determine the calcified atherosclerotic burden. An extremely elevated calcium score or proximal and clumped distribution of the calcium can render the coronary CTA unable to be interpreted and preclude contrast administration. Once this scan is completed, the coronary CTA portion of the examination can be performed. A typical coronary CTA protocol for a 64-slice CT scanner is outlined in Table 1.1.

### **Image Display and Analysis**

An advanced three-dimensional (3D) workstation is essential for analyzing the image data set accurately and efficiently. Multiple post-processing techniques, including bi-orthogonal projections, maximum intensity projections (MIPs), curved planar reformations, and surface- or volume-rendered displays can be used to evaluate the anatomy of interest [3]. For coronary CTA examinations,

**8 Part I General Considerations and Fundamentals of Imaging**

**Table 1.1** Typical Coronary CTA Examination Protocol: 64-Slice Multidetector CT

<b>Coronary Calcium Scan</b>	
Scan Parameters	
Range	Tracheal bifurcation to the bottom of the heart
Voltage (kV)	120
Effective current (mA)	310
Slice collimation (mm)	1.2
Slice width (mm)	3
Pitch	0.2
ECG gating	Prospective
<b>Coronary CTA</b>	
Patient Preparation	
NPO	3 hours prior to exam
Beta-blockade*	Metoprolol PO (50–100 mg) 1 hour prior to exam 5 mg IV prn (20 mg total maximum)
Nitroglycerin <sup>†</sup>	0.4 mg SL
IV catheter	18 gauge (preferably in antecubital fossa)
Scan Parameters	
Range	Tracheal bifurcation to the bottom of the heart
Voltage (kV)	120
Effective current (mA)	700–900
Detector collimation (mm)	0.6
Slice thickness (mm)	0.75
Pitch	0.2 (0.18 for heart rate < 50 bpm)
Rotation time (sec)	0.33 (0.37 for heart rate < 50 bpm)
Reconstruction interval (mm)	0.5
ECG gating	Retrospective or prospective
* Use with caution in patients with asthma, aortic stenosis, atrioventricular block, or severe left ventricular dysfunction.	
<sup>†</sup> Recent use of PDE5-selective phosphodiesterase inhibitors (sildenafil, tadalafil, and vardenafil) for treatment of erectile dysfunction is a contraindication.	

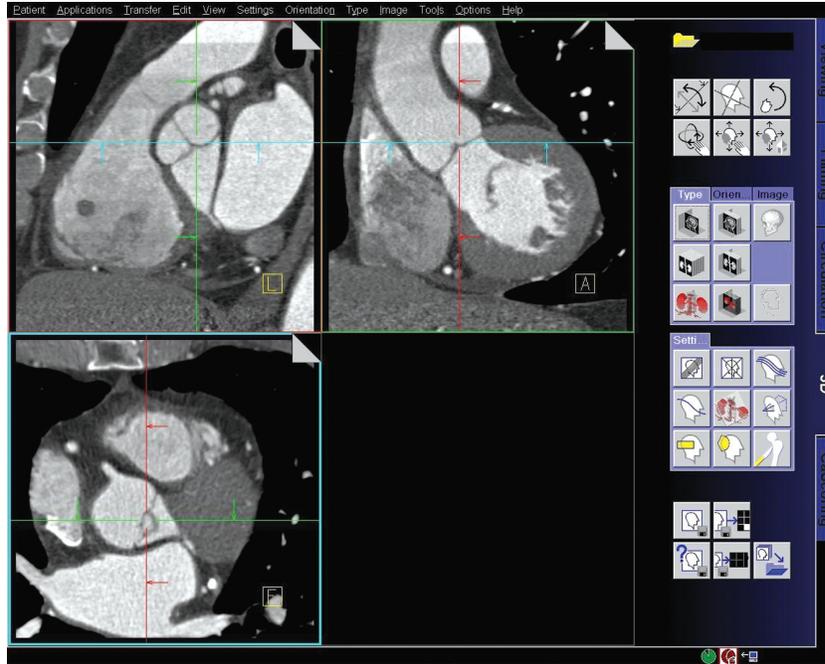


Fig. 1.2 Multiplanar image reformats from a cardiac CT data set.

these techniques assist with the identification of the coronary anatomy, the detection of atherosclerotic disease, and the diagnosis of any significant stenosis [3]. A screenshot of a 3D workstation display is shown in Figure 1.2.

A four-dimensional (4D) analysis can be performed when image data are acquired over the entire cardiac cycle. The 3D data set can be visualized as a dynamic or cine motion display. This dynamic capability allows the evaluation of cardiac function in addition to the cardiac anatomy. Left ventricular wall motion can be assessed. For instance, patients with significant stenosis of the LAD might demonstrate thinning and hypokinesis of the anterior wall consistent with myocardial infarct or hibernation. Cardiac valvular function can be assessed as well. For instance, dynamic images might demonstrate the prolapse of the mitral valve or the incomplete apposition of the aortic valve leaflets, suggesting the presence of aortic regurgitation. In addition to these qualitative assessments of cardiac function, quantitative information can also be obtained. For instance, measurements of the left ventricular volume at end-systole and end-diastole allow the calculation of the stroke volume and ejection fraction. These techniques enable a thorough evaluation of both cardiac anatomy and function.

## Cardiac MRI Basics

### Nuclear Magnetic Resonance Phenomena

The fundamental physical phenomenon underlying MRI is the fact that certain types of atomic nuclei (including hydrogen, which is abundant in the body in the form of water and fat) have a quantum-mechanical property called *spin* [5–7]. Such nuclei are often informally referred to as “spins.” The nuclear spin property leads to the nuclei exhibiting a collective magnetization in the presence of a strong external polarizing magnetic field. This nuclear magnetization is quite weak (the equilibrium magnetization depends on the strength of the field; it is equivalent to a net alignment of the nuclei with the external field of only around one part per million at typical imaging system field strengths), and it is not readily detectable in its equilibrium state, i.e., aligned with the external field.

The nuclear spin property also leads to the nuclei exhibiting an effective angular momentum. This angular momentum combines with the nuclear magnetization to create the possibility of exciting a nuclear resonance condition at a characteristic resonance frequency that is proportional to the magnetic field strength. For the field strengths used in MRI systems, this resonance frequency is in the radiofrequency (RF) range. For example, for a typical imaging field strength of 1.5 T (approximately 30,000 times stronger than Earth’s magnetic field), the hydrogen resonance frequency is approximately 64 MHz.

In the phenomenon of nuclear magnetic resonance (NMR), if a suitable short-duration pulse of RF magnetic field oscillating at the resonance frequency is applied to the nuclei, the nuclear magnetization can be induced to rotate away from its equilibrium orientation (“to be excited”). The net rotation angle produced will depend on the strength of the applied oscillating field and its duration. This resonance phenomenon will only be exhibited for a narrow band of frequencies around the central resonance frequency. When the RF magnetic field is turned off, the nuclear magnetization is left at a final net angle called the *flip angle* relative to the external polarizing magnetic field. The component of the nuclear magnetization that is perpendicular to the external polarizing field (*transverse magnetization*) will then rotate around the external field at the resonance frequency; this spinning magnetization will induce a weak but detectable signal in a suitable electrically conducting receiver coil. This signal is the basis of MRI and other applications of NMR [5,6]. Differences in the local tissue magnetization due to different amounts of nuclei (*proton density*) can provide a source of image contrast; for example, the lungs normally have less tissue density than the mediastinal tissues, and appear correspondingly darker. The magnetic resonance signal will not persist indefinitely but decays over time. Typically, the magnetic resonance signal decays exponentially with time, with a characteristic time constant,  $T_2$ , that depends on the state of the tissue; for example, fluids typically have a larger  $T_2$  than solid tissue, and edematous tissue typically has a larger  $T_2$  than normal tissue. These differences in the values of  $T_2$  provide

another potential source of image contrast if the signal detection is delayed to a later time; those regions with T2 values greater than or on the order of signal observation time will tend to appear brighter than regions with lesser values of T2, all other factors being equal (*T2 weighting*). The nuclear magnetization will gradually recover toward its equilibrium state, aligned with the polarizing field (*longitudinal magnetization*). Typically, this recovery is exponential in time with a characteristic time constant, T1, that also depends on the state of the tissue. The T1 value is on the order of or greater than the T2 value. As MRI typically involves the use of repeated excitations, if these excitations are repeated with a repetition time, TR, that is on the order of or smaller than T1, the magnetization will only recover part of its magnetization between consecutive excitations and will appear correspondingly darker in the image. The larger the flip angle, the larger this *T1-weighting* effect will be.

Although the resonance frequency is proportional to the magnetic field experienced by the nucleus, the different screening effects of the electron orbitals can result in different nuclear positions within a molecule having different characteristic resonance frequencies; this effect is known as a *chemical shift*. The chemical shift is a small effect; for example, there is approximately a 3.4 ppm difference between the resonance frequencies of hydrogen nuclei in water and fat. In the presence of different frequencies, the magnetic resonance signal will not decay as a simple exponential function of time, as there will be interference between the signals at different frequencies. The resulting complex signal as a function of time can be used to recover the corresponding distribution of different frequencies in the signal through a mathematical operation, the Fourier transform. The distribution of frequencies due to chemical shift differences can be used for magnetic resonance spectroscopy studies. If there is a wide range of frequencies present, the net observed signal can appear to disappear prematurely due to the resulting interference between them. However, the signal can be induced to transiently reappear at the full strength (limited by the T2 relaxation time) by applying suitable refocusing RF pulses; this reappearance of the signal is known as a *spin echo*, and the time of appearance of the echo is the echo time, TE. If a series of exciting RF pulses is applied at repetition times, TR, on the order of or smaller than the T2 time, the residual transverse magnetization can be refocused by the subsequent excitation pulses, adding to the new transverse magnetization being produced and leading to a condition known as *steady-state free precession*, with an increased signal that depends on the ratio of T2 to T1.

Image contrast depends on a combination of the relative contributions of the different relaxation times of the tissue and the different repetition and echo times and the flip angle used in the imaging. In addition to the T1- and T2-weighting effects described previously, we can create additional T1-weighting effects through the use of an additional RF pulse applied immediately before the signal excitation pulse, so that the recovery from the effects of the pulse at the time of the signal excitation will reflect the local T1 value. For example, such

## 12 Part I General Considerations and Fundamentals of Imaging

a pulse could have a net flip angle of  $90^\circ$  (saturation recovery, as the starting state of the magnetization is zero, or saturation) or  $180^\circ$  (inversion recovery, as the starting state of the magnetization is inverted). In addition, we can alter the image contrast through the administration of magnetic resonance contrast agents that shorten the tissue relaxation times. As will be discussed further, motion can also affect the magnetic resonance signal.

### Imaging Methods

Although the basic principles of the NMR phenomena have been known for over 50 years, their use for imaging was relatively more recent [7]. In part, this delay was because the NMR signal cannot be focused or collimated like other medical imaging signals. The use of localized signal detection coils can provide a limited amount of localization of the signal sources, but this cannot be used for imaging by itself (although it can be useful in speeding up more conventional imaging, as described later). The key to performing MRI is the use of magnetic field *gradients*, supplementary magnetic fields that are computer-controllable and that are designed to increase linearly along some desired direction in space [7]. With the use of gradients, we can couple the position of the signal sources along the direction of the gradient to the corresponding resonance frequency of the signal. The resonance frequency of a source will vary proportionally to the component of its position along the gradient direction and the strength of the gradient. The direction and strength of the gradient can be rapidly and flexibly changed by the MRI system computer during the data acquisition, and can be turned on and off as pulses.

The most straightforward use of gradients in MRI is for the frequency encoding of position by applying the gradient during signal detection. The distribution of the strength of signal sources along the direction of the gradient will be reflected in the resulting distribution of signal frequencies. These source locations can then be separated by performing a Fourier transform on the signal detected as a function of time, to find the corresponding distribution as a function of frequency (similar to its nonimaging use of the Fourier transform to find the distribution of chemical shift frequencies in magnetic resonance spectroscopy). These locations are then mapped to the corresponding distribution as a function of position along the gradient by knowledge of the strength of the applied gradient. By itself, the frequency-encoded signal in the presence of a gradient only provides information about the component of the image positions along the gradient direction, i.e., the projection of the signal sources onto the gradient direction. However, by repeating the signal excitation and detection process with the gradient direction being systematically changed to provide the projection data along different directions in space, we can build a set of radially oriented projection data analogous to that used for x-ray CT imaging. Image reconstruction using this approach, called *projection-reconstruction imaging*, was originally carried out using the same mathematical approaches as x-ray CT.

However, now computational methods are generally used that effectively interpolate the acquired data set onto an equivalent rectangular grid of samples of the Fourier transform of the image. The image is then reconstructed by the use of Fourier transform processing of this data.

Whereas projection-reconstruction imaging is simpler to understand, most MRI is currently carried out with *phase-encoded imaging*. In phase encoding, a pulse of magnetic field gradient is applied after the excitation but prior to the signal detection. Immediately after the excitation, in the absence of any gradients, the signal sources will all have the same resonance frequency and will be synchronized with the same signal phase. While the pulse is applied, the local resonance frequencies will vary along the direction of the gradient with a resulting accumulation of differences between the phases of the signal from different locations along the gradient. After the gradient is turned off, the frequencies will revert to their initial value, but the signal sources will be left with a difference in their phases that depends on their position and is proportional to the strength and duration of the gradient. If we now acquire the signal data in the presence of a frequency-encoding gradient along another direction, as described previously, the resulting projection data along that direction will be altered compared to the values that would be obtained in the absence of the phase-encoding gradient. In fact, by repeating this process and acquiring a set of such phase-encoded data with a set of different values of the strength of the phase-encoding gradient along the same direction, we can build up a set of samples of the Fourier transform of the image along that direction that can be used to reconstruct the image with corresponding Fourier transform processing. Typically, phase-encoding and frequency-encoding approaches are used jointly in MRI, with signal detections in the presence of a fixed frequency-encoding gradient pulse along one direction being used together with a set of preceding phase-encoding gradient pulses along a perpendicular direction.

As the MRI data are acquired at discrete sampled locations in the Fourier transform of the final image, the number and locations of these samples will determine the final achievable resolution and field of view (FOV) limits of the final image. Specifically, the extent of the data acquired in the Fourier transform domain along a given direction limits the corresponding resolution along that direction in the reconstructed image. To increase the resolution, we must either use stronger gradients or apply them for a longer time. The sparseness of the spacing between the samples in the Fourier transform domain along the phase-encoded direction limits the achievable imaged region FOV along that direction. Any structures located beyond the FOV will appear as if they were “aliased” back into the region, and can be superimposed on the images of structures that are really located in the region being imaged.

We can use magnetic field gradient pulses applied during the excitation process, together with RF pulses that are designed to have a limited frequency content, to achieve a corresponding spatially limited *selective excitation* of a

## 14 Part I General Considerations and Fundamentals of Imaging

desired slab or slice to be imaged. As the direction of the slice-selection gradient in space can be freely chosen, as can the frequency content of the RF pulse, we can very flexibly choose the orientation, location, and thickness of the slice. Thus, we can reduce a potentially 3D image reconstruction task to a two-dimensional one, with the imaging gradients applied along the directions in the plane of the slice.

As the repeated excitations and signal detections that are required for acquiring the data used in magnetic resonance image reconstruction take time, and there is a limited signal-to-noise ratio (SNR) of the detected signal, there are trade-offs that must be considered in choosing the imaging parameters. In particular, as only a limited number of data samples can be acquired in a given time period, there will be a trade-off in choosing how to optimize the resolution versus the FOV. Furthermore, as the signal per picture element (pixel) decreases with the decreasing size of the pixel, higher spatial resolution will come at the cost of a decreased SNR.

An obvious difference in imaging the heart compared to other organs is the cyclical contractions it undergoes. As the time required to acquire enough data to reconstruct a high-resolution image can be a large fraction of a cardiac cycle, there would be significant motion blur degrading the image. To avoid this motion blur, we generally use the synchronization of the data acquisition with the cardiac cycle, called *cardiac gating*, and gather the data needed for imaging each phase of the cardiac cycle from a set of relatively small temporal windows (or segments) within each of a series of heart beats. Cardiac gating can be applied in either a prospective or a retrospective manner. In *prospective gating*, the data acquisition is initiated at fixed times after detection of a synchronizing trigger, e.g., derived from detection of the QRS complex in the ECG. In *retrospective gating*, the data are acquired continuously and then interpolated onto equivalent consistent cardiac cycle times as part of the image reconstruction processing. Another potential source of motion blur in cardiac MRI is respiratory motion. To avoid this distortion, we can either keep the imaging times short enough that the patient can suspend respiration during the data acquisition or we can perform an additional gating process, only acquiring data during relatively consistent and motionless phases of the respiratory cycle (generally near end-tidal expiration).

There are two principal potential sources of effects of motion on the magnetic resonance signal due to blood flow or myocardial motion in cardiovascular MRI: "time-of-flight" effects and phase shifts. *Time-of-flight effects* reflect the fact that alterations of the local magnetization (either transverse or longitudinal) will persist for times on the order of the corresponding relaxation times and will move with the underlying blood or tissue; this effect can lead to motion-induced changes in signal intensity. *Phase shift effects* reflect the changing magnetic fields experienced by excited nuclei as they move along magnetic field gradients. Although the signal phase is not displayed in conventional magnetic resonance images, the imaging methods can be modified with suitable additional gradient pulses so that a magnetic resonance image is produced whose phase is directly

proportional to the local velocity component along some desired direction in space. Note that due to the limited range of possible phases, the velocity sensitization may need to be adjusted for a given flow condition to avoid aliasing of the apparent velocity.

### Imaging Systems

A central component of an MRI system is the imaging magnet itself. This magnet must be both very strong (generally using a superconducting magnet) to boost the magnetization and thus the corresponding achievable SNR, and uniform so that there are minimal pre-existing field gradients [7]. The associated system for generating magnetic field gradients must be both strong and fast-switching (within safety limits for rapidly changing magnetic fields) as well produce gradient fields that are quite linear over the imaging region.

The RF coils (which may or may not be different physical coils for transmitting excitation pulses and receiving the resulting signal) must be designed to be uniform and efficient over the desired response region. The coil can actually be composed of an array of multiple elements. There are safety limits on the amount of RF power that can be deposited in the human body per unit time. The RF receiver electronics must be of high quality, and may need to be multichannel to handle the signals from multiple receiver coils. The whole system is controlled by using a computer or computers to synchronize the RF and gradient pulses for signal excitation, as well as for the image reconstruction and other associated data handling.

### Speeding Up Imaging

To get around the imaging speed limitations referred to previously, we need to optimize the system hardware and software performance and tailor the imaging to the individual [7]. This customization includes choosing the smallest FOV dimension we can for orienting the phase-encoded (the “slowest”) direction, and optimizing associated trade-offs between spatial and temporal resolution. Acquiring multiple sets of samples of the Fourier transform from each excitation by creating multiple spin echoes, each with a different value of the phase-encoding gradient, is used in echoplanar imaging. Various approaches can be applied to reduce the amount of data that must be acquired, such as using the limited spatial localizing information available from the use of arrays of receiver coils to eliminate the aliasing that would otherwise result from undersampling the data; this is often referred to as *parallel imaging*. In *serial imaging*, similar and other approaches can be used to allow relative undersampling in the time dimension.

### Imaging Protocols

In a cardiovascular MRI examination, protocols generally include initial quick localizing images to define the desired region of focus as well as some overall survey images of the general region [7]. Images are then acquired in multiple

orientations to define the anatomy and function of different regions. These images can be acquired with different sensitivity to relaxation times to vary the image contrast. Contrast agents are often administered to assess both the early enhancement phases, as a relative measure of perfusion, and the delayed phases, as a sensitive (although not specific) indicator of local processes such as scarring or inflammation. Velocity imaging is often useful to assess flows, e.g., in the great vessels as a means to find the cardiac output.

### Image Display and Analysis

A typical cardiovascular examination will generate hundreds or even thousands of individual images; many of these are best viewed as movie frames with a dynamic display. The calculation of functional parameters such as cardiac volume dimensions or velocities requires additional specialized analysis programs.

### Summary

In this chapter, we have presented an overview of the basic concepts and principles of cardiac CT imaging and MRI. Knowledge of these fundamental principles is important when selecting the best examination for a particular patient and clinical scenario. Future technological advances will build upon these fundamental principles and address current weaknesses and limitations of each modality. In the chapters that follow, the clinical applications that use current technologies will be discussed.

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