Cardiac Magnetic Resonance: Physics, Pulse Sequences, and Clinical Applications

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Cardiac magnetic resonance (CMR) is a new and promising technique for image-based diagnosis in patients with known or suspected diseases of the heart. CMR allows clinicians to obtain relevant information on anatomy, function, perfusion, and viability of the myocardium. This technique offers the advantages of versatility, lack of ionizing radiation, and superior soft tissue contrast. The variety of clinical conditions that can affect the heart and the need to understand the time-varying movement of the heart in 3 dimensions adds challenges to interpretation of CMR above and beyond those present in understanding the imaging modality itself. The image intensities present in CMR scans can vary by orders of magnitude in the same subject depending on parameters set by the individual acquiring the data. These different appearances of images may reflect distinct pathophysiologic states and, therefore, an understanding of image acquisition is fundamental to the clinical diagnosis and assessment of disease.
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The need to explore new and potentially improved imaging modalities for screening, diagnosis, and prognostication of cardiac diseases is underscored by the observation that cardiovascular disease remains a foremost cause of death and morbidity in the Western world.[1] In recent years, advanced techniques such as multislice computed tomography (CT), positron emission tomography, 3-dimensional echocardiography, and magnetic resonance imaging of the heart have emerged as promising modalities that provide useful information for the practicing physician who recognizes the increasing incidence

of life-threatening cardiac conditions in virtually all patient populations. Of these new imaging methods, cardiac magnetic resonance (CMR) is an exciting diagnostic modality that provides high-quality images of the myocardium with unrivaled soft-tissue contrast.[2] Additionally, CMR allows clinicians to obtain relevant information on anatomy, function, perfusion, and viability of the myocardium.[3,4] This information, when interpreted appropriately, can be used to determine whether a particular disease state is likely or unlikely. This article will discuss the fundamental principles that underlie acquisition of magnetic resonance studies and review the literature that may facilitate interpretation of clinical CMR scans.

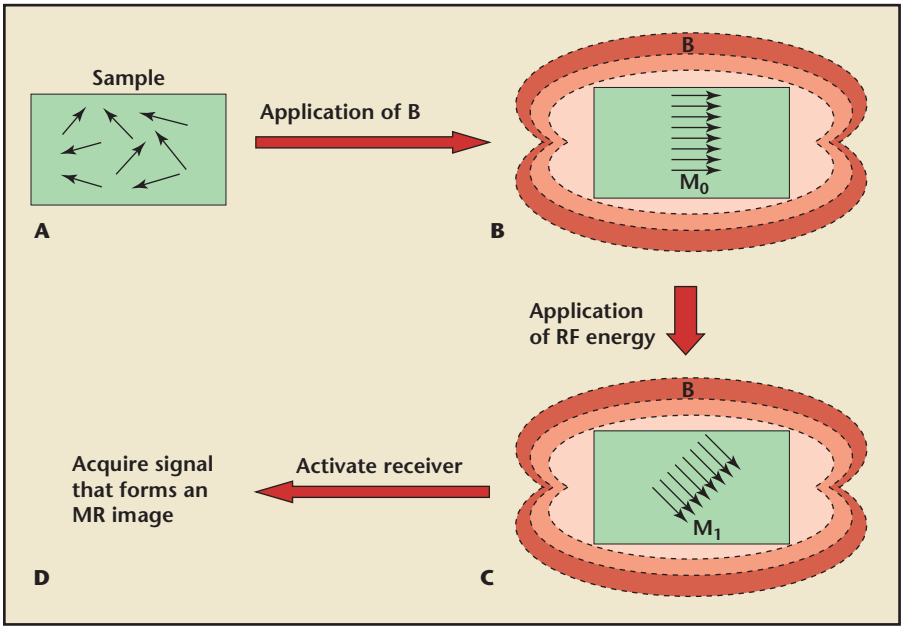
CMR is arguably as complicated as all other imaging techniques combined. The image intensities present in CMR scans can vary by orders of magnitude in the same subject depending on parameters set by the individual acquiring the data. These different appearances of images may reflect distinct pathophysiologic states and, therefore, an understanding of image acquisition is fundamental to the clinical diagnosis and assessment of disease. Furthermore, the unique and dynamic nature of the heart makes acquisition and interpretation of CMR images challenging. Although magnetic resonance imaging is an established modality for virtually every other body part, the variety of clinical conditions that can affect the heart and the need to understand the time-varying movement of the heart in 3 dimensions adds challenges to interpretation of CMR above and beyond those present in understanding the imaging modality itself.

Several excellent reviews of the current status of CMR in clinical practice have been published in the past few years.[2-5] Accurate appraisal of results obtained by CMR requires an understanding of 3 topics: physics of magnetic resonance, application of basic pulse sequences to the heart, and knowledge of the current literature that supports the use of CMR to evaluate particular pathologies of the heart. Clearly, an ability to identify normal and pathologic cardiac anatomy, knowledge of clinically relevant cardiac disease states, and an understanding of cellular pathophysiology are also necessary to achieve an accurate diagnosis using CMR. As most readers will already be familiar with cardiac anatomy, disease states, and cellular dysfunction, information in this review will not encompass these issues. Rather, this review will emphasize applications of CMR to problems in the current practice of cardiology. It should be emphasized that interpretation of CMR studies should embrace all of the above-mentioned information. Recommendations about when particular CMR exams should be ordered by physicians caring for patients with complex cardiac conditions will be alluded to but not exhaustively detailed. The reader is referred to clinical guidelines, current medical literature, and advanced texts for additional information on specific topics of interest.[2,4,6,7]

Physical Basis of Magnetic Resonance

A rigorous account of the physical principles that describe magnetic resonance is beyond the scope of this article but can be found elsewhere.[6,7] This section will provide a rudimentary set of principles that can be applied to the study of CMR. The basic sequence of events that describes how a signal that is based on nuclear magnetic resonance is used to obtain data is outlined in Figure 1. The presence of protons (^1H) on water molecules (H_2O) provides the matter that

Figure 1. *Diagram of the basic sequence of events whereby information by MR is obtained. The sample* **(A)** *when placed in a large magnetic field (***B***, where the magnetic field is denoted by dotted lines) develops a net magnetization, M_0, which is excited by an outside energy source causing a new magnetization state, M_1* **(C)**. *The relaxation of this new state back to the original state results in a signal* **(D)** *that forms the basis of an image. MR, magnetic resonance; RF, radiofrequency.* www.medreviews.com



allows for acquisition of data by magnetic resonance. Charges associated with protons exhibit a peculiar physical phenomenon known as *precession*, whereby they rotate. Rotating charges within water molecules cause a magnetic field that is specific to protons and is characterized by a particular rotational frequency (f, in units of cycles/sec or Hertz). Because water molecules in the liquid phase exhibit random motion, the net sum of the magnetic fields created by the protons is zero. However, when water molecules are placed in a large magnetic field (denoted "B_0," referred to as the *static* field, which has units of Tesla), such as the bore of a clinical scanner, the ensemble of spins develops a net bulk magnetization in the same direction as the magnetic field in which they were placed. Common static field strengths in clinical CMR imaging are 1.0, 1.5, and 3.0 Tesla. In theory, image intensity increases with higher field strength systems, assuming all other things are equal. The magnetization is written as a vector (denoted "M") and is the basis for data acquired using magnetic resonance. Importantly, the Larmor equation states that the frequency of precession of the protons is equal to a constant known as the *gyromagnetic ratio* (denoted "γ," in units of Hertz per Tesla) multiplied by the field strength that the protons experience. In other words, the rate at which the protons rotate is directly proportional to the strength of the field in which they are placed. It should be noted that the magnetic field used in clinical scanners is often created by current traveling in a circular pattern in wires that run outside the bore and are cooled by liquid cryogens. This current, by Faraday's law of induction, creates B_0 and causes the development of M within a sample that contains protons. It should also be noted that B_0 is a large and potentially dangerous field; only well-trained individuals with knowledge of dangers associated with magnetic fields should enter the area. Particular care should be exercised with any metallic objects (including indwelling devices, such as pacemakers, insulin pumps, and neurologic stimulators) that come within range of the magnetic field.[8]

The magnetization, M, described above is detectable after it has been excited by application of external energy, as suggested by Figure 1. Usually, this energy is in the form of a radiofrequency pulse that is transmitted onto the sample to excite protons. Following this excitation, M can be sensed by a receiver coil that is located near the water molecules that were excited. As shown in Figure 1, the magnetization is excited (or "tipped") by radiofrequency energy that is transmitted at the same frequency as the frequency (f) of the bulk magnetization of the spinning protons noted above. This stimulation causes the bulk magnetization vector (M) to shift from its initial orientation (M_0) along the main magnetic field to a new position (M_1). The angle between vectors M_0 and M_1 is a quantity referred to as the *tip* or *flip angle* (α, in units of degrees). After transmission of radiofrequency energy, M returns to its original position over a short period of time and, in doing so, emits energy that can be detected by a receive coil placed near the protons. The frequency of precession of the protons in the sample, the frequency of the transmitted radiofrequency pulse, and the frequency that characterizes the received signal are all similar and, hence, the term *resonance* is used to describe this sequence of events.

When M is excited, its new position, M_1, can be described by the sum of 2 vectors: one is oriented longitudinally in the direction of the bore (M_{long}) and the other is oriented transversely in a direction perpendicular to the bore (M_{tran}), as shown in Figure 2. As M_1 returns to M following the excitation pulse noted in Figure 1, M_{tran} becomes zero according to an exponential decay function characterized by time constant T_2 (the transverse relaxation time constant, in units of milliseconds). Concurrently, M_{long} becomes M according to another exponential function characterized by time constant T_1 (the longitudinal relaxation time constant, in units of milliseconds). Importantly, the energy that is emitted as M_1 and returns to M_0, and that can be detected by the receive coil depicted in Figure 1, comes exclusively from the transverse component of magnetization, M_{tran}, shown in Figure 2. Interestingly, the constants T_1 and T_2 are independent and have characteristic values in different water environments. For example, at 1.5 Tesla, the T_1 and T_2 values of heart muscle are approximately 650 ms and 60 ms, respectively.[9-12] Because particular tissue types (eg, fat, connective tissue, myocardium, blood) have certain expected values of T_1 and T_2, magnetic resonance imaging methods can be designed to look for particular pathophysiologic disease states. The static field, B_0, is ideally homogeneous but, in reality, has magnetic inhomogeneities. These variations in the field cause the observed T_2 (known as *T_2-star* and written as "T_2*") to be shorter than the actual T_2 of the protons excited. The relaxation of the magnetization from M_1 to M_0 alluded to above is described by a differential equation known as the *Bloch equation*.

By convention, the direction of the bore of the scanner is depicted as the z-direction; thus, M_{long} is the component of M in the z-direction, and M_{tran} is the component of M in

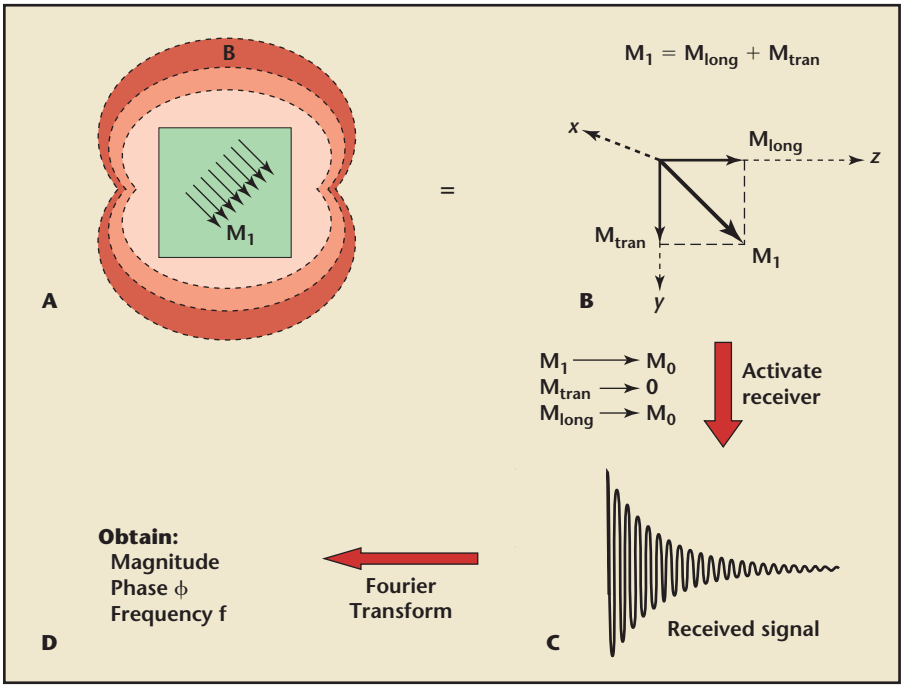


Figure 2. *Diagram of data acquisition. The excited sample, M_1* **(A)**, *is considered to have longitudinal and transverse components* **(B)**, *the latter of which decays to zero after excitation and results in a received signal* **(C)**. *The signal can be conveniently represented as a magnitude and phase (ϕ) at a particular frequency (f)* **(D)**. *M, magnetization.* www.medreviews.com

the x-y plane (Figure 2). The signal that is received by the detector in Figure 1 when M_1 is returning to M_0 takes the form of a decaying exponential function known as a *free induction decay*, as shown in Figure 2. The rate at which the waveform goes to zero is given by the time constant T_2, as described above. The function is usually expressed as a magnitude that oscillates at a characteristic frequency (f) and is offset by a phase (ϕ); these values can be obtained from the free induction decay by applying a mathematical tool, the Fourier Transform. Thus, after the spins present in state M_0 are excited by a radiofrequency pulse, the magnitude of the spins, the frequency at which they oscillate (f), and their associated phase (ϕ) can be obtained conveniently using the Fourier Transform, as suggested by Figure 2. The reason this mathematical tool is applied is that the free induction decay function depicted in Figure 2 is a complicated, time-varying signal; following transformation, the signal becomes a more manageable mathematical expression. In the example described in Figures 1 and 2, because there is 1 magnetic field (B_0), and because the Larmor equation states that the frequency of precession (f) of the protons excited is directly proportional to the field, the Fourier Transform of the acquired signal will return 1 value of magnitude, frequency, and phase.

Spatial Gradients

The basic principles of magnetic resonance described above can be applied to form an image. If an additional magnetic field, known as a *spatial gradient* (which has units of millitesla per meter), is applied to the original object in addition to the static fields in Figures 1 and 2, there will be different magnetic fields at different locations of the object and, as predicted by the Larmor equation, the precession frequencies within the object in the magnet bore will also be different. If, for example, there are 2 different fields (B_1 and B_2) created within the same object by the magnetic gradient, protons located in different regions of the object will exhibit different frequencies of rotation proportional to the strength of the field the spins experience. After application of a radiofrequency pulse, known as a "hard" or nonselective pulse, in which all spins are excited, the received signal will represent the relaxation of spins rotating at 2 different frequencies, created by B_1 and B_2, located at 2 different locations within the magnet bore. The received signal that results when spins with 2 different frequencies of precession relax is the sum of the 2 oscillating exponential decay functions, which, as mentioned, is a complicated mathematical expression; however, the relative magnitudes, frequencies, and phases of spins at both locations in the magnet bore can be determined by applying the Fourier Transform. Thus, in Figure 3, the magnitudes and phases that represent protons at the 2 particular frequencies will reflect information at the 2 different areas of the bore that can be used to create an image. Extending this principle further, many frequencies of precession can be created by applying different magnetic fields throughout an object. For example, if a gradient is applied such that there are *n* different magnetic fields within the object of interest, after excitation, the received signal will be a sum of spins at *n* frequencies all tending towards M_0 after a nonselective radiofrequency pulse is applied. The received signal can be converted into relative magnitudes and phases of spins at the *n*

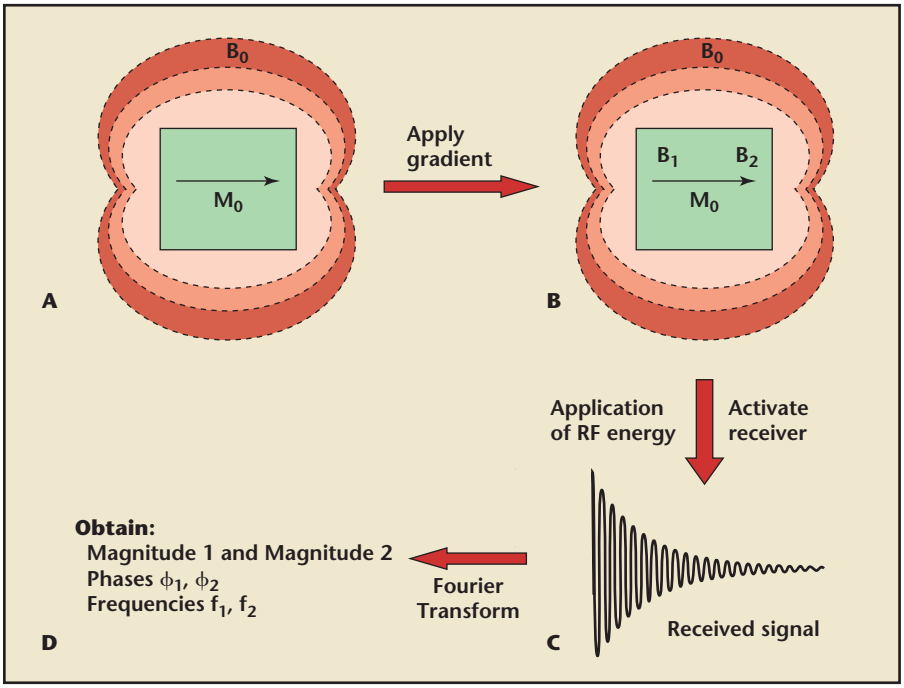


Figure 3. *Diagram of data acquisition with a magnetic gradient. The sample, M_0* **(A)**, *is placed in both a static field, B_0, and a field gradient that causes 2 different magnetic fields, B_1 and B_2* **(B)**. *After excitation and activation of a receiver, a signal is acquired* **(C)** *that, after Fourier Transformation, is represented as magnitudes (Magnitude 1 and Magnitude 2) and phases (ϕ_1 and ϕ_2) at corresponding frequencies (f_1 and f_2)* **(D)**. *M, magnetization; RF, radiofrequency.* www.medreviews.com

different frequencies by applying the Fourier Transform.

Awareness of spatial magnetic gradients that create different fields within the object of interest is critical to understanding how nuclear magnetic resonance can be applied to form an image. The gradient (G, in units of millitesla per meter) is applied, by convention, as a linearly-varying magnetic field over a particular distance. For clinical magnetic resonance scanners, a magnetic gradient is described by how fast it turns on and by the size of the additional magnetic field that is imparted; these parameters are combined into one term called the *slew rate* (in units of Tesla per second). The relationship between the magnetic field gradient, G, the frequency range that is excited, Δf, and the distance, *d* (in units of meters), that corresponds to the frequency range that is excited is described by the following imaging equation: $G = \Delta f / (\gamma\, d)$, where γ is the gyromagnetic ratio. Magnetic field gradients can be used to encode spatial information based on creation of different precession frequencies at different points within the object; this type of acquisition is termed *frequency encoding* or *read out*. If, on the other hand, a range of frequencies is created by a magnetic gradient and an excitation pulse is transmitted that excites only a narrow band of the frequencies within the sample created by the gradient, the acquisition is termed *slice selection*, the excitation pulse is termed a *slice selective* (or *soft*) *pulse*, and the magnetic field applied is a second type of gradient termed a *slice selection gradient*. The fields created by frequency encoding and slice selection gradients are perpendicular to one another. A third magnetic gradient whose direction is perpendicular to both frequency and slice selection gradients can be applied that imparts a phase to the sample prior to data acquisition; this is termed *phase encoding*. These 3 gradients allow for information to be encoded in such a manner that images can be formed. Following Fourier Transform, information on the magnitudes and phases of spins in different locations of the original sample can be derived and displayed in the form of a diagnostic image.

Pulse Sequences

Timed combinations of radiofrequency pulses, slice selection, phase encoding, and frequency encoding gradients are termed *pulse sequences* and are described by diagrams in which the exact sequence of events is outlined. Pulse sequences are implemented in a manner such that data throughout the object of interest are acquired. A typical pulse sequence will begin with radiofrequency transmission coupled with slice selection, followed by a phase encoding step, and then frequency encoding. During the last step, an analogue to the digital converter is activated whereby the data are sampled and converted to digital form for additional processing. The time between data samples is referred to as the *sampling time* (in units of microseconds), and the reciprocal of this quantity is the bandwidth (in units of megahertz) associated with the particular pulse sequence. In general, there are many excitations that are each followed by frequency encoding steps, and this scheme is continued until enough information about the region of interest within the object is acquired. The actual data acquired are usually in the form of a matrix of numbers and are termed *k-space*. The time between excitation pulses is termed the *repetition time* (TR, in units of milliseconds), and the time between the excitation pulse and the data collection is referred to

as the *echo time* (TE, in units of milliseconds).

Particular pulse sequences will affect the nature of the signal that is received and, indeed, pulse sequences can be tailored, or "weighted," to reflect T_1-, T_2-, or proton-density information or other data. Following Fourier Transformation of the k-space data, frequency information is converted to spatial information, and both magnitudes and phases of the spins at the spatial locations sampled are obtained. This information is then rendered digitally to form an image. Usually, only magnitude images are viewed clinically, but phase information can also be considered. The area of interest that is imaged in magnetic resonance is described by a slice thickness (in units of millimeters) and a field of view (in millimeters) in each of the in-plane directions. Also, the number of points that are acquired in each of the in-plane directions (termed the *matrix size*) is an important parameter in magnetic resonance imaging. The field of view divided by the matrix size provides the resolution or pixel size (in units of millimeters). The product of pixel size in each in-plane direction and slice thickness provides the voxel size (in units of cubic millimeters). Importantly, the signal per pixel that is apparent in a magnetic resonance image is increased by longer imaging times, is decreased by higher spatial resolutions, and is a fundamental quantity that affects diagnostic capability of CMR. The signal is often expressed as a quotient of signal to noise magnitudes (the signal-to-noise ratio [SNR]). In practice, pixel sizes of 1 to 2 millimeters in-plane with associated slice thicknesses of 5 to 10 millimeters are routine in clinical applications of CMR performed at 1.5 Tesla.

Even a cursory review of the specific pulse sequences that are relevant to cardiac diagnosis is beyond the scope of this article, and more advanced texts should be studied.[2,6] However, there are several points that can be emphasized. Pulse sequences that are commonly used for diagnosis of cardiac conditions vary according to the scanner manufacturer but, in general, include spin echo and gradient echo variants. These imaging pulse sequences are named according to the method by which the signal is obtained. For spin echo, sequential radiofrequency pulses are transmitted in order to provide particular excitation to the object of interest; for gradient echo, sequential gradient waveforms are programmed such that magnetization is maintained and relevant clinical information can be acquired. The choice of which pulse sequence may be used for a particular application is often guided by magnetic resonance manufacturer recommendations, and pulse sequences are organized in a manner that provides information in a given clinical setting. For example, information about basic anatomy can often be obtained using a spin echo sequence. For applications in which rapid imaging of a particular part of the heart is necessary, gradient echo sequences may be preferred. On occasion, pulse sequence choice is influenced by the rate at which the changing electromagnetic fields heats the patient; this parameter can be determined by ascertaining the specific absorption rate (SAR, in units of Watts per kilogram) of a particular pulse sequence.

Magnetic Resonance of the Heart

Although the principles of magnetic resonance described above demonstrate that images can be acquired, visualization of the heart presents several additional challenges. The cardiac cycle causes rapid motion of the myocardium that must be accounted for to obtain quality images. This can be done, in part, by performing CMR using a triggering signal (usually the R-wave of an electrocardiogram [ECG]). The triggering signal times the acquisition of data to an electrical signal that is related

Figure 4. *CMR images are often timed to a triggering signal, such as the R-wave of an ECG* **(A)**. *Following this signal, the excitation pulse with associated flip angle and slice selection is played, followed by phase encoding, and, finally, frequency encoding to acquire K-space lines* **(B)**. *These lines form the raw data that are reconstructed to form an image* **(C)**. *CMR, cardiac magnetic resonance; ECG, electrocardiogram; RF, radiofrequency.*

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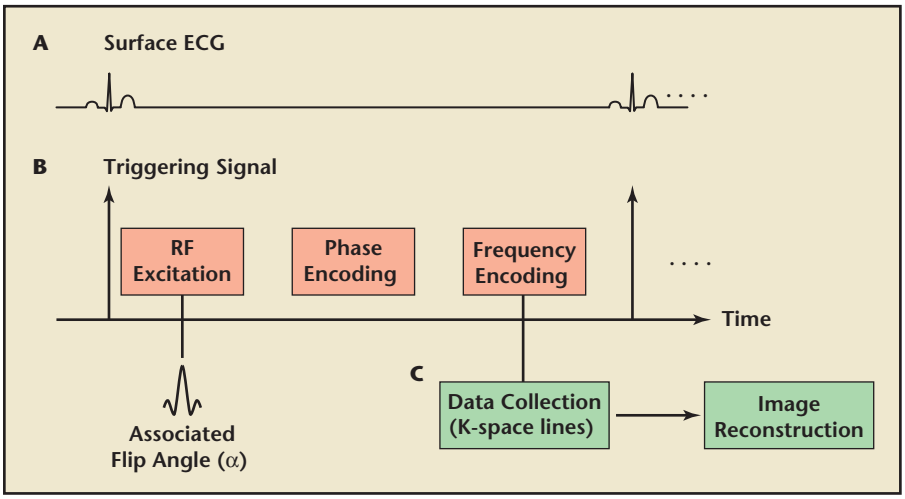
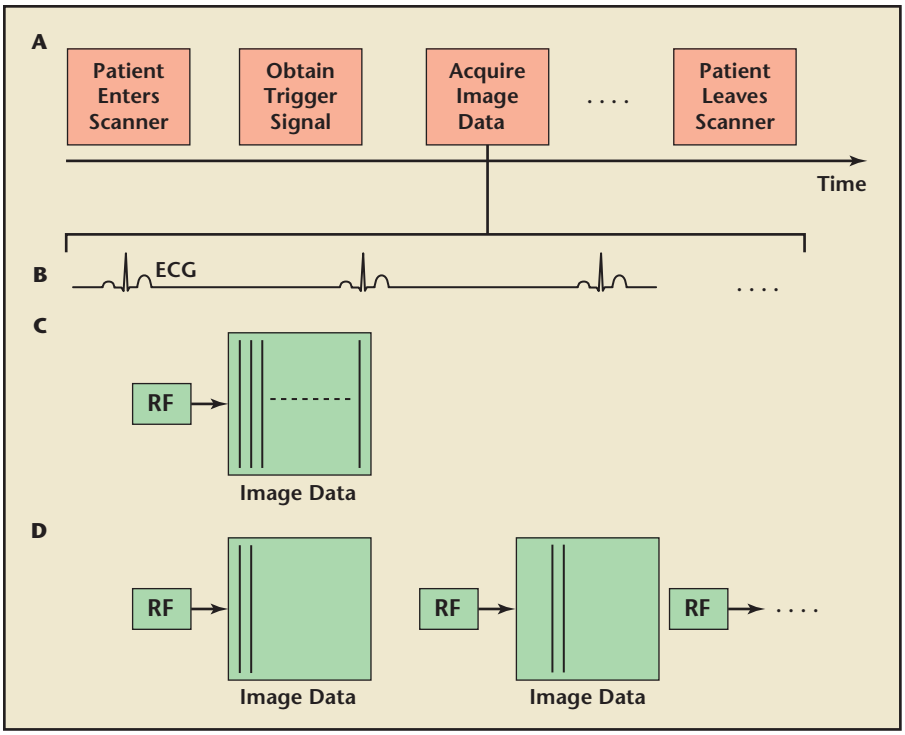


to cardiac motion, as shown in Figure 4. Other potential gating methods include pulse oxygenation, arterial blood pressure, and magnetic resonance navigator pulses, but these are not routinely used in clinical practice. The heart not only moves during the cardiac cycle, but it also translates within the thorax during respiration. Accordingly, CMR is often performed during a short (usually 10 to 20 seconds) breath-hold. Because breath-holds are required, the time a patient can hold his or her breath is a fundamental limitation that affects CMR image resolution for most applications. Finally, patients often need to be coached during a CMR examination. Emphasis should be placed on the need for good communication throughout the scan, consistent breath-holds, and the requirement of no movement once the patient is placed on the scanner table.

In addition to the technical obstacles present when imaging the heart, there are additional challenges to overcome when CMR exams are undertaken. Acquisition of standardized cardiac views is necessary for obtaining clinically useful data and for comparing CMR with other imaging modalities. Knowledge of the heart anatomy and the appearances of short- and long-axis views are crucial. Also, as shown in Figure 5, once the heart is positioned in the center of the scanner, CMR data can be acquired after excitation where the entire image is obtained ("single-shot" imaging) or can be acquired piecewise over several cardiac cycles for improved resolution and image quality ("segmented" imaging). Deciding which technique is appropriate requires user experience and may depend on the clinical information desired. Challenges in obtaining quality CMR data may occur in patients who have irregular heart rates, who have difficulty holding their breath consistently, or who are unable to lie still in the scanner. Adding another complexity to the CMR exam, an intravenous chelated gadolinium-based contrast agent may be administered to depict various structures or tissue types within the heart and, therefore, knowledge of how and when to perform contrast-enhanced CMR is important. Contrast agents cause changes in T_1 and T_2 of the blood and affect the appearance of tissues where contrast agents accumulate. Contrast agents allow for angiography, perfusion, viability, and other evaluations of the heart and should be employed where clinically indicated. An excellent review of contrast-enhanced CMR is provided by Edelman.[13] Finally, pharmacologic stress agents may be desirable for certain studies obtained by CMR. Clinical examinations have been performed in patients using adenosine, dipyridamole, and dobutamine CMR in a manner similar to stress radionuclide perfusion and stress echocardiographic studies.[14-16] An ability to assimilate technical knowledge, clinical indication, information about contrast agents, and the physiologic basis of pharmacologic stress agents is necessary for safe and accurate diagnosis using CMR technology.

Using the techniques described above, cardiac images can be acquired in a variety of ways. The desired information about the heart during CMR scans can be complicated, but it is usually broken down into distinct categories, as shown in Figure 6. Before the patient enters the scanner, his or her identity

Figure 5. *CMR images can be acquired in several ways. After positioning a patient in the scanner and with knowledge of the overall imaging sequence* **(A)**, *the ECG* **(B)** *can be used to correct for cardiac motion and acquire several kinds of gated images. Images can be obtained in 1 heartbeat using single-shot methods where the K-space is filled in 1 R-R interval* **(C)**, *or data can be acquired piece-wise over several heartbeats using segmented methods* **(D)** *for improved resolution and image quality. CMR, cardiac magnetic resonance; ECG, electrocardiogram; RF, radiofrequency.* www.medreviews.com



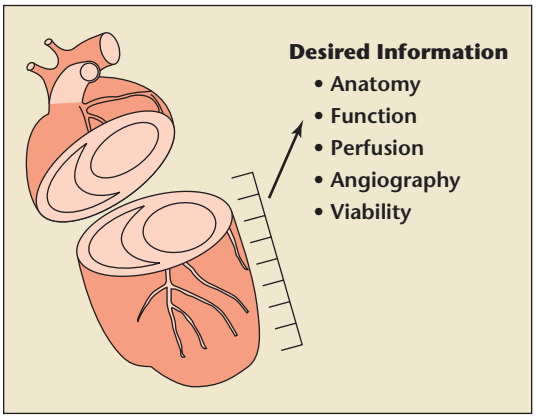


Figure 6. *Depiction of the heart cut in a short-axis cross-section during a cardiac magnetic resonance scan.* www.medreviews.com

should be ascertained, and the absence of contraindications to CMR should be verified. After ECG electrodes and a receiver coil are placed on the patient's chest, he or she should be moved into the scanner, taking care that the heart is advanced to the center. The ECG should be examined and a lead should be chosen where the R-wave is easily separable from the P-wave and T-wave. A typical cardiac examination is depicted by the sequence in Figure 7. Scout images of the heart are initially acquired to determine the location of the heart within the chest. Then, the short- and long-axis views are located, as are any other orientations that may be required. Next, routine examinations usually acquire triggered, breath-held, still-frame, anatomic images at end-diastole that encompass the heart, lungs, diaphragm, and great vessels. Following anatomic evaluation, gated images are acquired throughout the cardiac cycle at sequential short-axis and long-axis slices. Each group of images is played back in a loop that reveals cardiac function; these are referred to as *cine images*. Care is taken to image the myocardium, valves, and any suspicious structures seen on anatomic imaging. Where indicated, cine CMR can be performed with and without pharmacologic stress provided that sufficient facilities and safety procedures exist to do so. Contrast injection often follows cine imaging and can be used for angiographic or perfusion imaging. In the former, images are acquired that maximize intensity between the blood vessel lumen and myocardium; in the latter, single-shot images are acquired every cardiac cycle while a bolus of contrast transits through the myocardium. Contrast-enhanced perfusion data are obtained by rapidly imaging at select slices every heart beat. Perfusion imaging can also be performed with and without pharmacologic stress.

Figure 7. *Commonly acquired images and the work flow in routine CMR examinations. CMR, cardiac magnetic resonance.*
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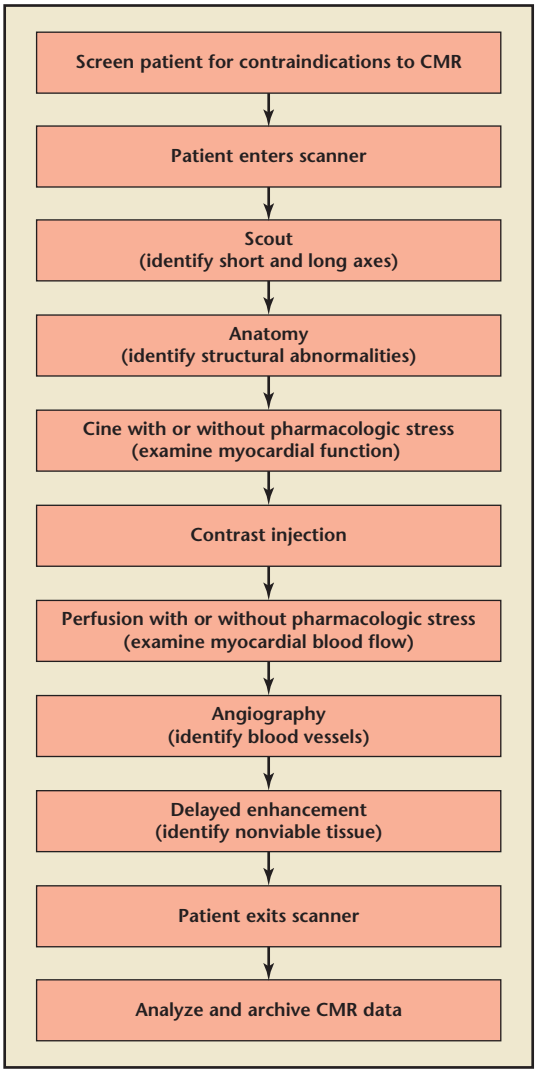

Finally, delayed images are also acquired to identify areas of irreversibly injured myocardium.

CMR examinations are often tailored to the clinical indication for which the patient was referred. For example, a study to determine the presence and amount of myocardial scar may be different from a study to examine whether dobutamine stress causes induction of wall motion abnormalities. Also, CMR studies for aortic angiography, valvular function, and pulmonary vein anatomy may emphasize different parts of the outline shown in Figure 7, and some clinical studies may omit parts of that algorithm completely for efficiency. Additional examinations can be considered based on the clinical indication for which the patient was referred for CMR, and centers often develop familiarity over time regarding which images need to be acquired for particular indications.

After CMR images are acquired, data are often archived, viewed on another workstation, interpreted, and reported. CMR data sets can be extremely voluminous because of the large number of images that are acquired and, indeed, storage can be a major issue. Methods of interpretation also vary by center and may require specialized post-processing systems. Cardiac examinations are usually interpreted with the aid of a heart model, and often a 17-segment model of the left ventricle is employed.[17] In addition, important findings can be present in noncardiac structures, and knowledge of potential findings in the pericardium, mediastinum, thorax, lungs, and diaphragm is essential.

In general, CMR requires a facility that has a scanner with appropriate hardware and software and a well-trained staff. Efficient scanning is often a result of collaboration among physicians, physicists, and technologists. Centers performing CMR should consider that cardiac patients are often prone to complications (arrhythmias, pulmonary edema, ischemia, etc) as well as to general difficulties during the scan (orthopnea, chest discomfort, shortness of breath, claustrophobia, etc). Staff should be trained in the appropriate clinical management of these conditions and have resources to move patients to a higher level of care as needed.

Clinical CMR

The utility of CMR in evaluation and clinical management of patients is controversial and challenging to define precisely. There are compelling data in the literature that suggest the technique provides important and relevant information that affects treatment decisions. Guidelines have been proposed that outline indications for appropriate use of CMR.[4] New indications are emerging as additional data become apparent. Clinical applications of CMR flow naturally from the sequence in which images are acquired, as shown in Figure 7, and are highlighted below.

Interpretation of CMR data usually begins with evaluation of basic anatomy. Often, continuous axial samples of the thorax are used. CMR provides excellent 3-dimensional visualization of cardiac structures, and such data can be used to assess for anatomical variants and congenital malformations, and for the presence of tumors, among other things.[18] Often, anatomic data are also used to assess sizes of the atria, ventricles, and the heart as a whole, which are parameters that may help guide the remainder of the examination.[19,20] Occasionally, assessments with and without fat-saturation pre-pulses may be performed to look for the presence of pericardial fat, intracardiac adipose tumors, and right ventricular dysplasia.[21,22] Also, clinical circumstances may call for assessment of T_1- or T_2-weighted images. For example, data suggest that CMR can be used to assess the degree of myocardial iron deposition in patients based on alterations of T_2 within the heart muscle.[12]

Following anatomic assessment from still-frame images, dynamic motion of the atria, valves, and ventricles over time is assessed using cine loops acquired in multiple views. Of particular importance in most CMR studies is detailed characterization of left ventricular wall motion. Assessments of ejection fraction, qualitative and quantitative wall motion, and ventricular volumes can be made. Cine CMR can also provide information on the right ventricle, intracardiac shunts, and valvular dysfunction.[23-25] These evaluations may be aided by acquisition of phase images in a manner similar to Doppler echocardiography. Functional evaluation of the left ventricle can be performed with and without an intravenously administered pharmacologic stress agent, such as dobutamine, to increase myocardial oxygen demand and determine whether inducible wall motion abnormalities exist. Several groups have documented ventricular dysfunction during pharmacologic stress in clinically relevant patient populations.[16,26]

After evaluation of cardiac function, often a gadolinium-based contrast agent is injected to assess blood vessels or myocardial perfusion. Intravenous agents can be used to perform angiography of the cardiac chambers, aorta, pulmonary arteries and veins, and the great vessels. In addition, several groups have described clinical coronary angiography by CMR, although the resolution is a fundamental limitation to this application.[27,28] Alternatively, contrast agents can also be used to acquire information on myocardial

perfusion where a bolus of contrast through the myocardium is observed in a dynamic loop. Similar to cine imaging, perfusion CMR can also be performed with and without a pharmacologic agent, such as adenosine, to effect coronary vasodilation. Several groups have documented stress-induced perfusion defects in viable myocardium that correspond to coronary stenoses.[15,29,30]

Following gadolinium injection, a technique known as *delayed enhancement* or *late gadolinium hyperenhancement imaging* is used to look for areas of irreversible myocardial injury.[31] An excellent review of principles of delayed enhancement imaging using the inversion recovery pulse sequence is provided by Kim and colleagues,[32] and an overview of the data in various disease states is provided by Bucciarelli-Ducci and coworkers.[33] In patients with coronary disease, the presence of late gadolinium enhancement within the myocardium represents myocardial infarction. In patients with no coronary disease, areas of elevated myocardial image intensity may represent myocyte death from causes other than coronary atherosclerosis. Identification of areas of delayed enhancement has been described in nonischemic cardiomyopathies, including hypertrophic obstructive cardiomyopathy, myocarditis, Chagas' cardiomyopathy, amyloidosis, sarcoidosis, Fabry's cardiomyopathy, and others.[34-40] There is certainly a large number of complicated conditions in which cell death may occur, and the reader is encouraged to diligently examine the literature to determine whether a particular condition of interest is associated with delayed enhancement.

Interestingly, groups that perform CMR may use different tests to look for the same clinical disease state. An example is assessment of presence of coronary stenosis. Using results at cardiac catheterization as the gold standard, investigators have employed stress cine, stress perfusion, and coronary angiography CMR to look either indirectly or directly for the presence of a coronary lesion.[15,16,27] Depending on the expertise at a particular site, the utility of CMR may vary or may not exist at all. In addition, it may be necessary to change the mode of CMR acquisition based on patient needs. For example, it may be the routine of a particular center to acquire anatomic views with a single-shot sequence. However, if an abnormality is seen, it may be appropriate to acquire additional images over the same region of interest with higher spatial resolution using a segmented approach. In addition, it may be that patients have frequent ectopy, causing variations in the regularity of their heart, and/or patients may not be able to reliably hold their breath multiple times. In such circumstances, attempts may be made to acquire anatomic information using single-shot CMR and function information using cine CMR that employs a real-time acquisition. In such cases, overall image quality is reduced compared with segmented methods but, usually, single-shot images for anatomy and real-time images for function are diagnostic.[41]

As emphasized by Rehwald and colleagues,[3] the most frequent indications for clinical CMR are stress testing, viability, angiography, and evaluation of congenital abnormalities. Rigorous comparisons of CMR with other emerging techniques (CT, positron emission tomography, 3-dimensional echocardiography, etc) for specific indications have not to date been performed. Rather, most studies have summarized results of selected small patient series and have shared experiences at centers of excellence in advanced cardiac imaging. For example, several lines of evidence suggest that delayed contrast enhancement by CT may be a useful test for scar detection in a manner similar to delayed enhancement CMR.[42,43] Also, the role of CT for coronary evaluation is becoming virtually routine at many clinical centers, suggesting that CMR coronary angiography may be less emphasized in the future.[44] CT may have roles for assessment of perfusion and myocardial function as well. Similar new and exciting data are emerging for positron emission tomography and echocardiography. A concept for the integration of some of these new techniques in the assessment of coronary patients has been recently proposed.[45] CMR has the advantages of versatility, lack of ionizing radiation, and superior soft tissue contrast. Which applications of CMR will be suitable for routine use remains to be determined. It may be that future CMR studies with larger patient populations as well as determination of whether CMR techniques are easily applicable at many centers will help discern the future roles of this imaging method.

Readers interested in undergoing training in CMR should consider applying for dedicated fellowships and adding additional course work to their studies. Attendance at major scientific meetings that have particular emphasis on CMR is critical, especially given the evolving nature of the field. Subspecialty training in magnetic resonance of the heart is available at several centers, and criteria have been outlined for Level I, II, and III certification.[46,47] Research efforts are also key to understanding the role for CMR in modern practice. A tremendous amount of investigation is ongoing to determine the precise utility for CMR in cardiac patients and whether CMR has additional indications. At many centers, clinical

practice can be appropriately coupled with research protocols.

Advanced Topics

CMR scanners are becoming more powerful, and initial data indicate that higher field strength systems (eg, 3.0 Tesla) allow for improvements in SNR that can be used to acquire images with higher resolution and/or reduced acquisition times.[48] Similarly, parallel imaging protocols with specialized receive coils have been designed to acquire phase encoding steps more rapidly.[49] Like the hardware, the software is also changing. Pulse sequences are becoming faster, incorporating true 3-dimensional information, and routines that allow for more streamlined scanning are being developed. Postprocessing workstations are also being designed to quickly handle the massive amount of CMR information that needs to be archived, analyzed, and reported.

Safety for patients undergoing CMR is a relevant and ongoing concern. Specifically, movement, heating, and malfunction of indwelling ferromagnetic devices are the subjects of multiple investigations.[8,50,51] Safety guidelines should be followed and reviewed frequently. Also, there is recent concern about nephrogenic systemic fibrosis due to gadolinium contrast agent administration.[52] This and other issues should be thoroughly taken into consideration by all parties when thought is given to conducting CMR studies.[8]

There is also need to establish whether CMR findings have prognostic value in patients. Preliminary conclusions have been obtained in patient series, but findings need to be generalized over larger populations, and more data are necessary.[53,54] Also, whether CMR provides incremental, complementary information in certain circumstances or whether CMR can be used in place of other imaging techniques has yet to be established definitively. In addition, incorporation of standard clinical data as well as information acquired using other imaging modalities is important for proper CMR interpretation and appropriate care of patients.[45,55] The exact algorithm with which to achieve these objectives, however, remains uncertain.

Conclusion

Improving outcomes of patients who suffer from diseases related to the heart by early diagnosis, rapid implementation of treatment, and adequate follow-up is a key goal of clinical medicine. Juxtaposed to the tremendous societal problem of cardiovascular disease is the emergence of imaging techniques with multiple possible cardiac applications, such as CMR. The wide potential applicability of CMR should be coupled with the management and treatment of patients with heart disease in a manner that is clinically appropriate. CMR requires experienced centers and specialized training of practitioners. The quality of CMR images by itself does not necessitate the examination or obviate the use of other methods. Rather, findings on CMR need to be related systematically to the presence or absence of underlying myocardial pathophysiology. A basic knowledge of principles of magnetic resonance with application to the heart and proper appraisal of past and ongoing studies should be embraced in order to correctly interpret clinical studies. Knowledge of fundamental principles that affect images can complement clinical findings and aid in the accurate diagnosis of cardiac conditions. As the field of CMR continues to evolve, the need for re-appraisal of existing evidence supporting its use in cardiac patients will become apparent. ■

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Main Points

- Cardiac magnetic resonance (CMR) allows clinicians to obtain relevant information on anatomy, function, perfusion, and viability of the myocardium.
- The presence of protons (^1H) on water molecules (H_2O) provides the matter that allows for acquisition of data by magnetic resonance.
- The time a patient can hold his or her breath is a fundamental limitation that affects CMR image resolution for most applications.
- Centers performing CMR should consider that cardiac patients are often prone to complications (arrhythmias, pulmonary edema, ischemia, etc) as well as to general difficulties during the scan (orthopnea, chest discomfort, shortness of breath, claustrophobia, etc).
- Findings on CMR need to be related systematically to the presence or absence of underlying myocardial pathophysiology.

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