Clinical Cardiology Frontiers

Functional Aspects of Cardiovascular Nuclear Magnetic Resonance Imaging
Techniques and Application

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For many years, attempts have been made to avoid cardiac catheterization by evaluating cardiovascular function using external recording devices including apex cardiology, ballistocardiography, the ECG, impedance cardiology, etc. At first the contribution of the noninvasive techniques was small, but the balance has been profoundly altered, first by the appearance of M-mode echocardiography, followed by two-dimensional, Doppler, and color flow echocardiography, nuclear medicine techniques, x-ray computed tomography, and most recently by nuclear magnetic resonance imaging (NMR, MRI, MR). Invasive investigation has changed, but virtually every cardiac diagnosis previously made by cardiac catheterization can now be made noninvasively with acceptable spatial and temporal resolutions without mortality or morbidity. The exception has been the coronary angiogram. Although there have been images of parts of the coronary arterial tree for several years, these are not comparable in quality with x-ray angiograms. It is doubtful, however, that the place for NMR is to be just another method of demonstrating coronary anatomy requiring interpretation yet revealing no information about flow in the vessel or the state of its wall. Measurements of flow in some coronary vessels during parts of the cardiac cycle have been possible for over 5 years, and robust techniques for visualizing major coronary arteries, measuring flow in parts of them throughout the whole cardiac cycle, are being developed. Coronary arteries have already been visualized using echo planar (EP) "snapshot" techniques, single breath-hold field echo, and spiral imaging. In larger arteries, analysis of the lipid content of plaque has been achieved, and when this is possible in the coronary arteries, prediction of acute thrombosis may be possible.

Special NMR Techniques for Cardiovascular Studies

There are a number of reasons why cardiovascular NMR has only recently begun to display its potential. Most NMR machines are expensive first-generation machines, many using high fields to improve signal to noise at the expense of increased sensitivity to chemical shift and movement artifact. They depend on slow sequences designed to produce high-resolution images of still structures. They were designed to be used by technologists for radiologists, whose work load has not historically included “hands-on” functional cardiac imaging. In the United States, where most NMR machines are sited, cardiologists have limited access to the machines and have underused them in favor of echocardiography, with which they are familiar. ECG gating was initially difficult. A few dedicated cardiac units have developed specialized hardware and software for blood flow measurement and rapid imaging demonstrating the feasibility of acquiring high-resolution anatomic images and accurate functional information. NMR imaging gives excellent spatial resolution and has the ability to acquire tomographic images in any orthogonal or oblique plane or in three dimensions. The NMR signal varies according to the local biochemical environment, therefore there is natural high contrast between different healthy and diseased soft tissues, and the indications for contrast media are rare. Chemical shift imaging and spectroscopy allow analysis of the chemical content of pathological lesions and abnormal metabolism. With the recent development of real-time and subsecond NMR imaging, NMR fluoroscopy, and single breath-hold movies, the application of this new imaging method to the cardiovascular system is likely to expand rapidly. One rate-limiting factor is the availability of machines optimized for cardiovascular studies.

The principles and details of NMR imaging have been well described many times. Most simpler descriptions “gloss over” the difficult and incompletely understood areas and the relevance of classic and quantum physics. The technique involves many interdependent variables, and an understanding of the fundamentals is useful (see “Appendix”).

Magnetic Resonance Velocity Mapping

Spin echo sequences are not suitable for blood flow measurement because moving blood escapes out of the slice during the sequence, and no signal for analysis is available from these “black blood” sequences. The various methods of measuring flow by NMR have been reviewed by Mosteck et al. As yet, only the method of phase-shift velocity mapping has progressed to clinical use. This technique and its clinical applications are considered in detail in this review. The NMR signal possesses both phase and amplitude. The phase of the signal is ignored in conventional images; however, it is possible to encode velocity or acceleration or higher

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orders of motion in the phase of the signal, and a phase image may be constructed to be used as a velocity map or an acceleration map. Phase encoding as a method of measuring flow was first proposed by Moran in 1982, then, almost simultaneously, applications were described by two different groups in 1984. The chosen parameter (velocity, etc.) can be displayed as a gray scale or color coded, using color to represent direction and intensity of velocity.

A combination of gradient echo imaging with even echo rephasing gives a high signal even from rapidly moving blood and allows accurate flow measurements to be made. A magnetic gradient is applied for a short time in the direction of expected flow, which causes the protons to precess faster in the stronger field and slower in the weaker field. After a pause, the gradient is reversed for the same duration. Still tissue experiences a phase change during the first gradient, which is nullified by the second, whereas moving blood or tissues move into a different phase territory and experience a phase change proportional to velocity. Velocity and higher orders such as acceleration and jerk can be encoded either through the image plane or within it. Many other factors, including inhomogeneity of the applied magnetic field and magnetic susceptibility variations within the patient also lead to phase changes, and to compensate for these, interleaved images are acquired alternating between velocity-encoded and reference images. Subtraction of the two images yields the velocity map. This technique has been validated in vivo using a four-way comparison between magnetic resonance flow measurements in the aorta and pulmonary artery and left and right ventricular stroke volumes in normal subjects. A good correlation has been shown between NMR flow measurements of venous return and aortic flow and between aortic flow measured by NMR and Doppler ultrasound. Comparison of NMR velocity mapping with Doppler ultrasound and cardiac catheterization correlate well. The technique is also shown to be accurate in experimental animals when compared with in vivo flow meter measurements and in vitro when calibrated against flow phantoms. Reduction of the echo time of the gradient echo flow measuring sequence reduces signal loss, which precludes accurate velocity mapping at sites of stenosis, where increased velocity, turbulence, and high orders of motion cause signal loss. The shorter the echo time, the higher the threshold of turbulence intensity at which the signal is lost. Using an echo time of 3.6 msec, flow velocities up to 6 m/sec can be measured reliably covering the range of blood flow velocities found in humans.

The selection of an appropriate velocity window before the acquisition is necessary in order to obtain accurate results. For example, the phase change of 0° to 360° might be used to encode velocities from −4 m/sec to +4 m/sec. If velocities outside this range are encountered, the phenomenon of aliasing occurs, and a velocity of +5 m/sec would appear with a velocity of −3 m/sec. However, the position of the velocity window could be altered after the acquisition to display velocities from −3 m/sec to +5 m/sec or any other velocity window spanning 8 m/sec.

The technique is accurate, relatively simple to use, and can be implemented on standard scanners. Its clinical application has been delayed by the failure of cardiologists to recognize its importance and by manufacturers to make the technique available for widespread use, but some centers are already using the technique successfully.

### Rapid Imaging

The reduction of acquisition times for NMR imaging was first realized by Mansfield with the one-shot EP technique. In this method, a complete two-dimensional image is recovered from a single radiofrequency excitation using rapidly switched magnetic resonance field gradients to form an echo train. The data acquisition time can be reduced to tens of milliseconds and can be used after preparation pulses for anatomic, diffusional, or chemical shift imaging and velocity encoding. EP makes severe demands on the hardware and software of the system, but it is now commercially available and may be fitted to some existing machines, with substantial modification. It is particularly useful in the investigation of organs moving periodically, such as the heart and those moving aperiodically such as the gut. The technique has
also been used for semiquantitative analysis of brain perfusion\textsuperscript{33} and myocardial perfusion using first-pass techniques with the magnetic resonance contrast agent\textsuperscript{34} (see "Myocardial Perfusion"). EP can be also used to produce high-resolution flow velocity mapping.\textsuperscript{35} Spiral echo-planar imaging can produce images with a temporal resolution of 40 msec or less, enabling continuous monitoring of flow throughout the cardiac cycle.\textsuperscript{36,37}

More recently, alternative techniques have been developed that are less demanding on the hardware and software of the system. The best known of these is the subsecond FLASH technique (fast low-angle shot), which forms a train of echoes by using rapidly repeated radiofrequency excitations with a very short echo time.\textsuperscript{38,39} The technique is slower than the EP method, gathering data typically over a few hundred milliseconds, but can usually be implemented on unmodified machines. It is also capable of freezing motion, even of the heart, despite the longer acquisition time, by virtue of the fact that the gross appearance of an image is mainly determined by the data lines in relatively few central phase-encoding steps\textsuperscript{40} taken in the first 50 msec of an image acquired over 300 msec.\textsuperscript{41} This fast imaging technique has been used to measure blood flow by encoding velocity on the phase of the NMR signal\textsuperscript{42} and to measure myocardial perfusion using a first-pass technique with gadolinium-DTPA.\textsuperscript{43} Real-time NMR velocity mapping has been demonstrated using a standard clinical hardware with a novel pulse sequence design to measure flow within
a cylindrical volume (beam) at up to 60 frames per second.44

Clinical Applications: The Heart

The Myocardium

**Myocardial function.** NMR images are dimensionally accurate.45 The volume of a chamber can be measured in systole and diastole by summing the areas in contiguous images (Figure 1). Ventricular stroke volumes can be measured from a set of transverse images at end systole and end diastole. Eight pairs of slices 10 mm thick give full coverage of the heart, with an acquisition time of about 30 minutes. The endocardial outline in each image may be drawn and the volume computed in each slice. Volume measurement by this method is independent of cavity shape, unlike other methods in which geometric assumptions usually have to be made, and is ideal for the right ventricle. The technique has been shown to be accurate to approximately 2%.45 When left and right ventricular volumes were compared. However, in the noninfarcted left ventricle, an acceptable rapid method of volume measurement for this chamber is by area-length calculations on oblique images containing the long axis of the ventricle (Figures 2A and 2B).46-48 Muscle volume and hence, mass, can also be measured on similar principles49-53 and is important in patients with left ventricular hypertrophy (Figure 3). For example, it has been possible to show regression of hypertrophy after only 3 months treatment of hypertension.50 Wall thickness in hypertrophic cardiomyopathy is better demonstrated than by echocardiography because the three-dimensional distribution is more easily seen.57 Muscle thinning and aneurysm formation are also readily recognized (Figure 4), and previous infarction can be detected and quantified by the presence and extent of thinning58 and wall motion abnormality measured from diastolic and systolic images by superimposition of endocardial contours.58,59 Ventricular function can be studied using cine display,60 allowing systolic function to be evaluated from wall motion and wall thickening, which in the left ventricle is normally uniform and concentric. Reliable measurements can be made of ventricular volumes between end systole and end diastole. Reversible myocardial ischemia can be shown on cine images after pharmacological stress as an area of abnormal wall motion that is otherwise normal on the nonstressed scan.61,62 Acceleration of blood in the aorta measured by NMR velocity mapping before and after pharmacological stress is useful for assessment of the extent of myocardial ischemia and its reversibility.63 Myocardial tagging, in which narrow presaturation bands are applied to monitor progressive distortion of the myocardium during the course of the cardiac cycle, has been used to study regional wall motion and myocardial strain.64-66 Myocardial motion can be quantified using NMR velocity mapping.67,68 These methods should increase understanding of ventricular filling in healthy and diseased hearts (Figure 2, panels C, D, E). There is a growing interest in the distinction between irreversible myocardial damage and "stunned myocardium" that may recover if properly managed, and NMR might make a valuable contribution.69-76

**Myocardial perfusion.** Bolus tracking techniques77,78 have been used to measure tissue perfusion, notably in the kidney and brain.79 If the curve of concentration versus time can be plotted as a known quantity of a tracer passes through an organ, organ perfusion can be calculated from the area under the curve. Alternatively, if the tracer is wholly extracted by the organ, the principles described by Sapirstein60 enable perfusion to be measured from the amount of tracer trapped by the organ, a principle used in nuclear cardiology using both single-photon and positron emitters.81 Commercially available NMR contrast agents using tracers such as gadolinium-DTPA equilibrate rapidly with the extracellular compartment and are washed out rapidly. Therefore, they are not blood pool tracers and are not fixed in the myocardium. Nevertheless, by treating them mainly as extracted tracers, it is possible to measure myocardial perfusion from the peak myocardial enhancement. The model assumes a linear relation between tracer concentration and signal enhancement. This is only true for gadolinium-DTPA at low concentrations, which appears to be the case in animal experiments and in clinical practice.82-84 Now that echo planar and ultrafast gradi-
ent echo imaging can provide at least one image for each cardiac cycle during the passage of the tracer, measurement of myocardial perfusion with high resolution is possible.85-87

Valvular Disease

Detection of turbulence. Gradient echo sequences show turbulent blood flow as areas of signal loss within the high blood signal.88,89 This is highly dependent on the echo time, field strength, and acquisition parameters. At an echo time of 14 msec, which is typically available on commercial machines used for general cardiac studies, there is minor signal loss around the tips of the aortic and mitral valve cusps in normal subjects. In patients with valve abnormalities, the area of this turbulent signal loss is increased and immediately identifiable (Figure 5). The corollary to the ready detection of turbulence by signal loss is that there is no signal for velocity measurement in the turbulent volume. This is overcome by using a reduced echo time.

Application of the modified Bernoulli equation. A stenotic valve or vessel may be assessed by measuring the flow velocity in the jet of blood passing through a stenosis. In the case of a given flow, an increasingly narrow stenosis leads to an increase in the velocity of flow through the orifice. The relation between the velocity of the jet and the difference in pressure on either side of the stenosis can be approximated by the modified Bernoulli equation, which in its simplest form is ΔP=4V^2, where ΔP=pressure drop across the stenosis (mm Hg) and V=velocity (m/sec). This assumes that the blood velocity upstream of the stenosis was negligible. This calculation can be easily applied to determine the peak instantaneous gradient across the stenosis and to calculate the mean gradient during the flow period. It can be used to determine the pressure differential between two chambers in the case of a regurgitant valve.

Stenotic valves. The first step in the assessment of the velocity of blood in a jet uses a magnitude image with an echo time that clearly identifies the jet direction and allows subsequent velocity mapping to be aligned correctly. Accurate velocity mapping of jets through stenoses requires the use of very short echo times.20 The velocity map may be through-plane, with the jet passing perpendicularly through the chosen imaging plane, or in-plane, when the imaging plane is chosen to encom-

![Figure 5](https://circ.ahajournals.org/)

**Figure 5.** Spin echo (panel A) and gradient echo (panels B and C) images in a transverse plane in a patient with tricuspid regurgitation caused by cardiac sarcoidosis. The right atrium is very enlarged. In the gradient echo images in diastole (panel B) and systole (panel C), the jet of tricuspid regurgitation is clearly seen. 1, Left ventricle; 2, right ventricle; 3, right atrium; 4, descending aorta.

![Figure 6](https://circ.ahajournals.org/)

**Figure 6.** Gradient echo image (panel A) and a corresponding velocity map (panel B) in a patient with aortic stenosis. The unusual oblique plane was necessary to orientate the abnormal jet direction vertically for velocity encoding. Velocity profile displayed in the center of the jet recorded a peak velocity of 3.3 m/sec (44 mm Hg). 1, Left ventricle; 2, ascending aorta; 3, pulmonary artery; 4, right atrium.
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FIGURE 7. Gradient echo image showing the left anterior descending (arrow) and left circumflex coronary arteries (arrows) of a healthy volunteer (panel A). The equivalent velocity map (encoded from right to left) shows flow (in black) in the left anterior descending artery (panel B). 1, Ascending aorta; 2, main pulmonary artery; 3, left atrium; 4, descending aorta; 5, superior vena cava.

pass the length of the jet (Figure 6). In-plane imaging yields a greater number of pixels for analysis of velocity, but if the jet is small, in-plane imaging is less reliable because of partial volume effects and movement of the jet out of the imaging plane. It is preferable to acquire data in both planes.

Regurgitant valves. The size of the signal loss has been used to provide a semiquantitative estimate of regurgitation through valves. However, each center must calibrate the technique for their own equipment because of the dependence on echo times, field strength, and acquisition parameters. The extent of turbulence can either be expressed by grading the size of the jet or better as an absolute volume. Color-flow Doppler echocardiography is also capable of determining the size of turbulent flow, but volume measurements are unreliable because of the inability to acquire contiguous parallel images of known thickness. It is difficult to separate the turbulent volumes when dual valve disease exists (such as mitral stenosis and aortic regurgitation). NMR can be used to determine both the stroke volumes of the ventricles. The stroke volume ratio between right and left ventricles approximates unity over an imaging period of minutes, and any discrepancy reflects the severity of regurgitation blood flow. This technique only assesses the severity of isolated valve lesions be-

FIGURE 8. Origin of left main coronary artery from pulmonary trunk: Systolic spin echo image in a transverse slice in the mid right atrioventricular groove showing an enlarged right coronary artery (arrow, panel A). The equivalent velocity map showing caudal diastolic flow in white in the dilated right coronary artery (arrow, panel B). Instantaneous flow volume curve (liters per minute) calculated from the complete cine acquisition and plotted against time in right coronary artery (RCA) showing systolic and diastolic peaks (panel C). 1, Left ventricle; 2, right ventricle; 3, left atrium; 4, right atrium.
cause one side of the heart must be normal for comparison. Likewise, it is not possible to distinguish the individual severity of valvular regurgitation when two valves on one side are affected. A more robust technique to quantify the severity of valvular regurgitation involves the subtraction of flow in a great vessel measured by velocity mapping from the stroke volume of the associated ventricle calculated from the contiguous spin echo technique described above. This allows true isolation of the left and right sides of the heart for assessment of regurgitant fraction and regurgitant volumes. The technique cannot, however, separate regurgitant flow from two regurgitant valves on one side of the heart.

**Coronary Arteries and Coronary Artery Bypass Grafts**

Coronary artery bypass grafts can be assessed using NMR imaging. Spin echo imaging is able to identify patent grafts, but a signal void may be associated with small volumes of blood moving forward and backward rather than adequate forward flow. Flow measurement within the grafts would be clinically valuable but is not always possible because of signal loss caused by sternal suture and clips left after surgery. Metal objects cause larger artifacts in field echo flow measuring sequences than in spin echo imaging sequences.

Coronary artery flow measurements cannot yet be made reliably by conventional NMR imaging, although good results can be obtained in favorable circumstances. Figure 7 shows an in-plane velocity map with flow in the left anterior descending artery and the right coronary artery. Although it is less difficult to acquire an in-plane velocity map, partial volume effects mean that through-plane maps are more accurate. Figure 8B is a velocity map in a patient with an anomalous left coronary artery arising from the pulmonary trunk that led to shunting from the aorta to the pulmonary artery through the right coronary artery around the apex into the low pressure system. The coronary flow profile showed two peaks in systole and diastole (Figure 8C). Flow measurements in coronary arteries of normal size are difficult because of the tortuosity of the vessels and cardiac and respiratory motion. Possibly, these difficulties will be overcome by three-dimensional acquisition and display techniques and by rapid imaging. Recently, rapid breath-hold cardiac scanning techniques have shown improved coronary visualization (also see Manning...
Great Arteries

Aorta

The aorta, because of its size and relative immobility, is particularly suitable for all types of quantitative velocity imaging, which has provided new insights into its function (Figure 9). Normal systolic flow in the ascending aorta is a plug flow with a skewed velocity profile with higher velocities around the inside of the arch. Throughout diastole, the blood continues to move with simultaneous antegrade and retrograde channels. The reason for this complex pattern is unknown. In normal subjects, the retrograde channel is closely associated with the left coronary sinus, and it is tempting to speculate that it augments flow in the left coronary artery by imparting momentum to the blood that is destined to enter it. In patients with coronary artery disease, the retrograde channel is smaller and may enter any of the coronary sinuses. In aortic valve regurgitation, the magnitude of the retrograde flow is understandably increased (Figure 10), and aortic or pulmonary regurgitation may be quantified from the back-flow of blood in the proximal great vessels assessed by velocity mapping.

Aortic dissection. Extent of aortic dissection is readily detected by NMR imaging and is displayed including involvement of other vessels (Figure 11A). The entry and exit points are more difficult to localize, but there is no doubt that invasive investigation can be avoided with a combination of echocardiography and NMR imaging. The previous inability to detect involvement of the aortic valve is no longer a problem, since aortic regurgitation can be detected and quantified on cine NMR imaging and velocity mapping. Although an adequate assessment of the coronary arteries is not obtained by NMR imaging, this is not always necessary before surgery because most surgeons believe that it is difficult to perform coronary arteriography in patients with dissections involving the ascending aorta, and this is rarely a reason for cardiac catheterization. The relative merits of NMR imaging and computed tomography have been compared, and the two investigations appear equivalent in their sensitivity. NMR imaging has the advantage of oblique planes and does not require contrast injection, but it is difficult to image severely ill patients in the current generation of scanners. The sensitivity and specificity of NMR imaging and transesophageal echocardiography have been studied; both are highly sensitive methods to identify and classify acute and subacute dissections of the thoracic aorta, but NMR imaging has a better specificity (fewer false-positives). A thin intimal flap may not show in spin echo images unless static blood in the false lumen provides natural contrast with the true lumen. If there is any doubt, then the flap will be more easily seen using a gradient echo sequence, and velocity mapping will confirm the diagnosis by demonstrating the differential flow velocities in each lumen (Figure 11B). Aortic compliance. Aortic flow wave velocity and regional aortic compliance can be measured by NMR imaging. The flow wave velocity is calculated from the delay between the appearance of the flow wave in ascending and descending limbs of the thoracic aorta. This parameter is closely related to aortic compliance and may be useful for the detection and monitoring of arterial disease. In addition, aortic compliance is an important component of left ventricular afterload, and reduced compliance will not only increase myocardial oxygen demand but may also reduce oxygen supply through its effect on the normal aortic diastolic pressure and backflow that aids coronary perfusion. Similarly, reduced pulmonary arterial compliance may be a factor in the onset of right heart failure in patients with chronic lung disease. Another area of clinical importance may be after surgical repair of aortic coarctation, where persistent systolic hypertension may be the result of more widespread structural changes in the aorta.

Pulmonary Arteries

The retrosternal position of the central pulmonary arteries makes it difficult to assess pulmonary blood
flow by Doppler echocardiography, especially in the presence of skeletal or lung abnormalities. NMR velocity imaging is not technically constrained and is capable of accurate blood flow measurement in any plane or direction. Pulmonary flow profiles have been less well studied, but NMR velocity mapping has confirmed the early forward systolic peak and the increased reverse diastolic flow in patients with pulmonary hypertension. This abnormal pattern (Figure 12) may be caused by the reflected waves from the distal vasculature, which has a high impedance.

 Patients with a single lung transplant are unique in that the cardiac output is ejected into two pulmonary vascular beds with different characteristics, and in these patients, the differential blood flow depends on the relative resistance in each lung. Velocity mapping is useful for assessment of total and differential pulmonary blood flow, which may be useful for monitoring these patients. The ratio of blood flow in the transplanted and the native lungs was 3:1, and the flow profile in the artery of the transplanted lung showed forward flow during systole and most of diastole, whereas that of the native lung showed a narrow early systolic peak and a reverse flow in most of diastole (Figure 13).

**Great Veins**

The caval veins are also relatively large, and reliable velocity maps and flow measurements can readily be obtained. The normal pattern of flow has two forward peaks in ventricular systole and diastole (Figure 10C), but this pattern is disturbed by disease. Any condition that causes impaired filling of the right ventricle reduces the diastolic peak, a pattern seen in constrictive and restrictive cardiac disease (Figure 14, panels A and B). Tricuspid regurgitation attenuates the systolic peak of caval flow, sometimes to the extent that reverse flow occurs (Figure 14C). This is perhaps less helpful because the severity of regurgitation can be assessed by cine imaging or from a comparison of right and left ventricular stroke volumes. Nevertheless, a normal sys-
Pulmonary cine curves

4, pathological pressure retrograde the disease. providing for the ability toolic flow peak suggests that tricuspid regurgitation, if it is seen, is not significant. NMR is commonly requested for the assessment of pericardial disease; therefore, the ability to measure caval flow is an important adjunct, providing an estimate of the functional significance of the disease. In patients with obstruction of the superior vena cava, absence of flow can be confirmed and retrograde flow in the azygous vein can be measured.120

Pulmonary Veins

Pulmonary venous flow velocity is a reflection of the pressure gradient between the pulmonary veins and the left atrium,121 and changes of this gradient under pathological conditions affect the flow pattern in the pulmonary veins.122 Normal pulmonary venous flow measured by NMR velocity mapping shows two peaks of forward flow, one during ventricular systole and the other in diastole (Figure 2E).21,123 A small back-flow during atrial systole occurs. A similar reverse flow has been recently demonstrated in the pulmonary veins by transesophageal Doppler echocardiography during atrial systole and transmirtal “A” flow peak.124,125 A noncompliant left ventricle produces high left atrial pressure during atrial systole, causing the retrograde flow in the pulmonary veins to become relatively larger than the flow through the mitral valve. An attenuated systolic peak has been demonstrated in patients with mitral valve regurgitation, and the degree of this

FIGURE 13. Spin echo image (panel A) of the pulmonary artery bifurcation of a patient with left lung transplantation. Flow curves of the main (MPA), right (RPA), and left (LPA) pulmonary arteries of the same patient are calculated from the complete cine velocity mapping (panel B). Blood flow in the transplanted left pulmonary artery is qualitatively and quantitatively different from the that in the native right pulmonary artery. 1, Main pulmonary artery; 2, right pulmonary artery; 3, left pulmonary artery; 4, ascending aorta.

FIGURE 14. Spin echo image in a transverse plane at midventricular level in a patient with constrictive pericarditis showing pericardial thickening (arrows, panel A). 1, Left ventricle; 2, right ventricle; 3, right atrium. Panel B: Superior vena caval (SVC) flow curve of the previous patient measured from the complete cine velocity map acquisition throughout the cardiac cycle. The diastolic peak is attenuated, which implies impaired right ventricular filling. Panel C: Superior (SVC) and inferior vena caval (IVC) flow curve in a patient with tricuspid valve regurgitation (shown in Figure 5). The systolic peak is attenuated, and there is retrograde flow in the inferior vena cava in systole.
attenuation correlates well with the severity of regurgitation.19

**Peripheral Vascular Disease and Atheroma**

Atherosclerotic vascular disease is the most common cause of death and disability in the Western world, producing problems either by reducing blood flow because of the development of a high-grade stenosis or catastrophic sudden total occlusion or by the release of emboli from ulcerated plaques. Clinical symptoms occur when the reduction in blood flow is greater than the compensatory vasodilation of distal arterioles. Techniques for detecting atheroma depend either on imaging the diseased vessels or on assessing the effect of stenosis on pressure and flow. NMR has the unique potential to 1) noninvasively assess the morphology of plaques, 2) study the hemodynamic significance of atheroma that has been demonstrated (Figure 15),126-128 3) determine the lipid content of plaques to “stage” them using chemical shift imaging,126,129 and 4) determine suitability of distal arteries before bypass grafting.130 These parameters could be important in the study of the progression of peripheral vascular disease and its response to pharmacological and surgical intervention and in planning treatment of lesions. In addition, the convenience and practicality of NMR velocity mapping for the detection and assessment of a peripheral arteriovenous fistula has been reported.131 Velocity encoding can be used to enhance a flow image enabling automated detection of the vessel wall and visualization of flow.132,133 However, NMR vascular imaging shares with other tomographic techniques the disadvantage of in-

**FIGURE 15.** Multiple atheromatous plaques causing stenoses of both common iliac arteries and the origin of the left internal and external iliac arteries (arrows): Panel A: X-ray angiogram; panel B: spin echo image; panel C: magnetic resonance velocity map in the same plane as panel B showing velocity profiles across abdominal aorta (1) and right (2) and left (3) iliac arteries. There is increased peak velocity in both iliac arteries compared with the aorta. The peak velocity is greater in the left iliac artery, which demonstrates that the stenosis on the left is greater than that on the right.
terrated display of vessels. This problem has been solved in NMR angiography by three-dimensional acquisition and display. An alternative approach for velocity mapping is to acquire information in multiple slices and then to combine these in an image in which only the maximum intensity of each pixel is displayed, irrespective of which slice in which it is seen.

**Congenital Heart Disease**

NMR can spare the patients from invasive investigation in many cases. It is possible to detect lesions causing left-to-right shunts, including ventricular and atrial septal defects and patent ductus arteriosus, although specificity for the detection of atrial defects is a matter of debate. Our own experience is that specificity using spin echo imaging in the transverse plane is unsatisfactory, but in practice, the debate will be eclipsed by different techniques such as the use of oblique planes, cine imaging, and velocity mapping. The strength of NMR imaging in the assessment of septal defects is not so much in detecting the lesion (which will usually have been found by echocardiography), but in assessing its functional significance by measurements of flow through the shunt. Abnormalities of the atrioventricular valves (Figure 16) and the great vessels are well visualized (Figure 17A), but again, NMR is of greatest value in assessing functional sequelae such as ventricular contraction, regurgitation, and flow (Figure 17, panels B and C). NMR imaging is particularly suitable for the display of anatomy in cyanotic and complex congenital heart disease (Figure 17). Venous, atrial, ventricular, outlet, and arterial anatomy and connections can all be shown with the acquisition of multiple slices in the three orthogonal planes. The images have a striking clarity that is superior to two-dimensional echocardiography and less dependent on operator skill and experience. However, in practical terms, the examinations are complementary, and NMR imaging should not be regarded as a potential replacement for echocardiography but as a method of completing the anatomic assessment, thereby rendering catheterization and angiography unnecessary. In pulmonary atresia, the presence and state of the central pulmonary arteries and of systemic collaterals to the lungs is important, but this information can be very difficult to obtain, and NMR has reduced the number of invasive investigations in these patients. Perhaps the most useful application of NMR in complex congenital heart disease is in the assessment of patients after surgery. For example, in transposition of the great arteries after Mustard’s operation, right ventricular and tricuspid valve function are important determinants of long-term morbidity, and these can be assessed repeatedly and noninvasively by NMR imaging. Surgical conduits can also be clearly seen and accurate measurements of the lumen made and compared at subsequent studies to detect early evidence of obstruction. NMR is therefore an important method of following up patients after Fontan’s operation and its modifications and all patients with ventriculopulmonary conduits inserted for the correction of Fallot’s tetralogy, pulmonary atresia, double outlet, and truncus arteriosus. NMR images have also been obtained in neonates and infants with congenital heart disease. Sedation is often required, but encouraging results that can lead to a reduction of invasive investigation in these patients have been reported. In the future, the development of rapid imaging techniques will be of considerable help in this area.

Clinical application of NMR velocity mapping has been very successful in patients with congenital heart disease. Intracardiac shunting can be measured in a number of ways. In atrial septal defects, the shunting can be assessed from the right and left ventricular stroke volumes, but this method is not applicable in complex lesions or in ventricular shunts. Flow directly through ventricular defects can be visualized, but the best method has been to measure Qp/Qs directly from aortic and pulmonary flow (Figure 18). This is also very helpful in complex lesions. The possibility of surgery in such patients depends partly upon pulmonary flow, which is difficult to measure by other techniques. From measurements of flow in the aorta, pulmonary artery, and right and left pulmonary arteries, it was possible to calculate flow through each of the defects separately. Other structures of interest for the measurement of flow are surgically created shunts and conduits.

**Ventriculopulmonary Conduits**

The diagnosis of obstruction in extracardiac ventriculopulmonary conduits by noninvasive methods can be difficult. Obstruction may be silent and progress unnoticed. Conduits may be obstructed at the valve or within the tube, as in peel formation within Dacron tubes. It is important to recognize significant obstruction early before right ventricular dysfunction occurs. In 90% of these patients, NMR imaging produces excellent images. There is a correlation between definite obstruction and a conduit diameter of less than 18 mm. Using velocity mapping, the value of NMR imaging is increased and localization and quantification of jet velocity have been accurate when correlated with invasive and noninvasive hemodynamics.

**Aortic Coarctation**

Late complications of surgery for coarctation including restenosis, aneurysm in association with a Dacron patch, and systolic hypertension require careful long-term supervision. Chest radiography is not reliable for the recognition of restenosis, and an aneurysm may extend posteriorly and not be visible on the routine
Figure 17. Spin echo image in a coronal plane in a patient with corrected transposition of the great arteries, mitral and pulmonary valve stenoses, and a ventricular septal defect (panel A). There is a double-inlet left ventricle, giving rise to the pulmonary artery. Small right ventricular outlet chamber gives rise to the aorta, which is to the left of the pulmonary trunk. Panel B: Diastolic (left) and systolic (right) gradient echo images in a plane similar to Figure 19A, showing turbulent jet distal to stenosed mitral (arrow) and pulmonary valves (arrows). Panel C: Gradient echo image in an oblique plane perpendicular to the pulmonary valve stenotic jet with the corresponding velocity map. Panel D: Cranial flow velocity in the ascending aorta and pulmonary artery is seen in black; caudal velocity in the descending aorta and superior vena cava is seen in white. The velocity profile across the stenotic jet shows a peak velocity of 3.5 m/sec (49 mm Hg). 1, Left ventricle; 2, right ventricle; 3, aorta; 4, pulmonary artery; 5, right atrium.

Radiograph. Imaging by two-dimensional echocardiography is also unreliable because it may be difficult to obtain a good echo window. However, continuous wave Doppler measurements are useful for predicting the presence of a gradient. When restenosis or aneurysm formation are suspected, catheterization and angiography would normally be performed to confirm the findings before reoperation. NMR imaging can provide high-resolution, dimensionally accurate images of aorta and its lumen noninvasively and without administration of contrast agent (Figure 19A). Furthermore, cine NMR velocity mapping allows assessment of the hemodynamic significance of aortic coarctation by measurement coarctation jet velocity and aortic volume flow (Figure 19B).

Limitations of NMR Imaging

NMR imaging is not a universal panacea for imaging in the body. Up to 4% of the normal population are claustrophobic, although most ill patients will tolerate an examination. Implanted pacemakers are a contraindication. Metal objects and implants are a theoretical contraindication; in practice, metal objects in the chest such as sternal wires, metal clips, and prosthetic valves are not ferromagnetic, and at currently available magnetic field strengths, there is no danger of the field exerting significant mechanical forces on them. However, the prosthetic material itself, with its a low proton density, gives little NMR signal, and the metal causes a localized image defect caused by distortion of the magnetic field. The defect is small for spin echo images but larger in gradient echo images, making it difficult to assess velocities and blood flow in the immediate locality of the prosthesis. Other limitations include the long acquisition time, approximately 60 minutes per study, the poor quality of cardiac gated images in patients who have cardiac arrhythmias, and the high cost of the magnetic resonance machine. Each of these limitations is being addressed, with potential developments including rapid imaging, flat bed magnets, and inexpensive solenoid magnets.

Conclusions

Although it is noninvasive, NMR imaging has greater potential than any diagnostic instrument yet conceived. Special techniques to harness its full potential promise
opportunities that will have a profound effect on cardiovascular health care: 1) the replacement of the battery of cardio-diagnostic tests involving radiation and invasion; 2) the ability to study the vessel wall and determine the lipid content of plaques; 3) cost-effective quick studies of the atherogenic process might stimulate drug companies to invest the huge sums to develop innovative compounds to control the disease; 4) screening asymptomatic people for vascular disease in, for example, the aorta or carotid arteries, and by measuring compliance and blood flow leading to a better understanding of the natural history of occlusive vascular disease.

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Appendix

A basic understanding of the physics involved in magnetic resonance imaging is invaluable in appreciating the capabilities of the technique, and all the concepts involved are comprehensible to physicians. Most MR studies are of the hydrogen nuclei (protons) in water. Protons spin on their axes at about a billion billion revolutions per minute, and because they carry a positive charge, they generate a tiny magnetic field like a current flowing round a coil of wire. If the body is placed in a strong magnetic field, the magnetic protons tend to align with it; curiously, unlike compass needles, which align with the north-seeking pole to the north, nearly half the protons lie in a high-energy state with the north-seeking pole facing south. Spinning nuclei are perturbed by thermal energy; the smallest quanta of energy are large in relation to the protons, causing them to precess at a fixed angle of 54°. Just as a child’s top precesses slowly in relation to the rate of spin, the nuclei precess at radio frequencies (rf). The random precessions can be brought into unison by applying an rf pulse at the resonant frequency, putting energy into the system that also increases the number of protons in the high-energy state. The “crowd behavior” of the protons precessing in unison causes the excited tissues to transmit rf signals whose frequency is related to the local field strength. This coherent signal is soon lost, however, because the local magnetic field is inhomogeneous because of imperfections in the magnet and the presence of a
nonuniformly magnetic body within it, interaction with adjacent magnetic molecules, and the movement of iron-containing blood. This degradation of the signal is represented by the time constant $T2^*$ (measured in milliseconds). This is always shorter than signal loss because of dissipation of energy from the system as it returns to the resting state after the rf excitation (represented by the time constant $T1$, which is about 1 second in most tissues). During the NMR experiment, three magnetic gradients are applied first to select the slice, then in the remaining two orthogonal planes to encode spatial information based on the relation of frequency to the local field strength. The jumble of different radiofrequencies are detected by a receiver coil close to the body. The emitted signal is processed mathematically to create an image that is a map of the amplitude of the rf signal emitted by protons in the water in the imaging plane. The sequence of events used to create an image can be varied in many ways. A commonly used imaging sequence is known as the spin echo. The rf pulse is used both to put extra energy into the system and to bring the precessing nuclei into phase to produce a coherent signal. This coherence is soon lost because of the $T2^*$ effect. In the spin echo sequence, a further rf pulse is applied to reverse the precession, and the forces that caused the initial dephasing now work in the opposite direction, reinstating the coherence producing a further signal known as the spin echo. Alternatively, the field can be reversed to bring the nuclei back into phase, gradient, or field echo imaging.

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