Magnetic resonance angiography (MRA) has excited the interest of many physicians working in cardiovascular disease because of its ability to noninvasively visualize vascular disease. Its potential to replace conventional x-ray angiography (CA) that uses iodinated contrast has been recognized for many years, and this interest has been stimulated by the current emphasis on cost containment, outpatient evaluation, and minimally invasive diagnosis and therapy. In addition, recent advances in magnetic resonance (MR) technology resulting from fast gradients and use of contrast agents have allowed MRA to make substantial advances in many arterial beds of clinical interest. The goal of this scientific statement is to present the current state of MRA of the extracranial arteries and to suggest current as well as possible future clinical applications for MRA. For the purposes of this statement, MRA is defined as MR techniques that provide cross-sectional or projectional images of normal and diseased arterial anatomy. It does not deal with the equally important area of quantitative flow measurement with MR. The first section deals with the technical basis of MRA. Subsequent sections deal with individual vascular beds in which MRA has shown clinical promise.

MRA: Technical Considerations

The “gold standard” for many manifestations of vascular disease, especially arterial occlusive disease, is CA, an invasive, costly, and potentially hazardous procedure. MRA could represent an alternative, noninvasive approach. Rather than a single technique, MRA actually represents a family of different techniques. As outlined below, contrast between blood and soft tissues is derived from completely different MR mechanisms in the various MR techniques. We will consider the basic principles underlying the MR imaging (MRI) appearance of flowing blood and the techniques used to image blood flow and render angiogram-like MRI scans.

Depending on the imaging technique used, blood may appear bright or dark. On traditional spin-echo MR images, blood vessels usually appear dark. With spin-echo pulse sequences, a pair (90° and 180°) of section-selective radiofrequency (RF) pulses is used to produce an MR signal. If blood flows out of the plane of the section in the time interval between successive RF pulses, the result is an absence of signal, called a flow void. The flow void can be emphasized by use of a thin section or a long echo time. In a fast spin-echo sequence, a long train of echoes is acquired by use of a series of 180° RF pulses; as a result, washout effects are even more substantial than with conventional spin-echo techniques. Other methods for creating a flow void include presaturation, which involves application of an additional RF pulse outside the plane of the section to suppress the signal intensity of inflowing blood; dephasing gradients; and presaturation pulses to nullify the blood signal.

To create bright blood, gradient-echo pulse sequences are used. In a gradient-echo sequence, only a single RF pulse is applied during each sequence repetition, so no signal is lost because of washout effects. Data for images on which blood is bright can be acquired as a series of overlapping thin sections (sequential 2-dimensional [2D]) or as 1 or more thick (3-dimensional [3D]) volumes. Each sequence has advantages, as discussed below. Bright-blood techniques can be subcategorized into time-of-flight (TOF) and phase-contrast techniques. The basis of TOF techniques is that positive flow contrast is generated by inflow effects, whereas the background (stationary tissue) is saturated by the rapid repeated application of RF pulses. Saturation pulses based on flow geometry can be used to eliminate unwanted vessels (e.g., leg veins for lower-extremity arteriography). Use of a segmented gradient-echo sequence with cardiac triggering is helpful to eliminate arterial pulsation artifacts. The basis for phase-contrast MRA is that the flow of blood along a magnetic field gradient causes a shift in the phase of the MR signal. With phase contrast, pairs of images are acquired that have different sensitivities to flow. These are then subtracted to cancel background signal, leaving only signal from flowing blood. Phase contrast also permits flow (velocity) quantification because the phase shift is proportional to the velocity.

The core of MRA is the ability to portray blood vessels in a projective format similar to CA. Currently, projection images are created by postprocessing of images acquired by a 2D or 3D gradient-echo sequence. Although image processing can be postponed until after the patient has left the MR suite, it is best done while the patient is still within the magnet so that additional scans can be obtained if needed. Most
commonly, the images are processed by use of a maximum-intensity projection (MIP) algorithm.\textsuperscript{7,8} With an MIP algorithm, the brightest pixels along a user-defined direction are extracted to create a projection image. Areas with poor flow contrast, including the edges of blood vessels and small vessels with slow flow, may be obscured by overlap with brighter stationary tissue.\textsuperscript{9} Through reduction of the pixel size and suppression of the signal of stationary tissues, the quality of the MIP can be improved substantially. The reduction in pixel size is accomplished by use of a large (eg, 256\times512) acquisition matrix along with a small field of view.

A variety of artifacts caused by phase or magnitude variations in the MR signal afflict MRA. Within a single image volume element (voxel), blood protons flowing at different velocities accumulate a range or dispersion of phase shifts. Complex flow can produce signal loss due to intraview phase dispersion (ie, it occurs during each repetition of the pulse sequence), ghost artifacts from view-to-view signal variations (ie, those that occur over multiple sequence repetitions), and flow-displacement errors relating to the time delay between RF excitation and frequency encoding or between phase and frequency encoding. These effects tend to falsely exaggerate the severity of a stenosis and are most pronounced with 2D MRA. They can also limit flow-based MRA imaging in areas of slow flow, such as aneurysms. A short echo time (TE) minimizes flow-displacement and phase-dispersion artifacts; phase dispersion is further decreased when the voxel size is minimized (eg, by use of thin sections). Small voxels and short TEs are most easily obtained with 3D TOF methods. The biggest drawback of the thick volumes used with 3D is that slow or recirculating flow can become saturated. The advantages of both 2D and 3D techniques are gained with a series of thin-slab 3D acquisitions. The sequential 3D (or multiple overlapping thin-slab acquisition [MOTSA]) technique gives better flow enhancement than single-slab 3D techniques and less dephasing than 2D techniques.\textsuperscript{10} The method has some drawbacks as well. For instance, the nonrectangular profile of the 3D slabs necessitates the use of substantial overlap (up to 50\%) of adjacent slabs, so that total scan time is increased compared with single-slab 3D. Moreover, signal-intensity variations within the individual slabs due to saturation effects cause an annoying "venetian blind" artifact. With sequential 2D or 3D acquisitions, slight patient motion can generate discontinuities in the vessel contour that can be mistaken for focal stenosis or fibromuscular dysplasia.

Recently, an increasingly preferred method for chest and abdominal MRA is the combination of breath-hold 3D gradient-echo sequences with short repetition time (TR) and short TE during administration of a gadolinium chelate, often in a double dose.\textsuperscript{11,12} The contrast agent shortens the T1, or longitudinal relaxation time, of blood (to as low as 50 ms, compared with normal blood T1 of \approx1200 ms), so that the blood appears bright irrespective of flow patterns or velocities. Correct timing of the injection is important to ensure synchronization between the transit of contrast material and scanning. Methods that can be used to achieve correct timing include empirical estimation of transit time, a small test bolus of contrast agent to determine the time delay between injection and arrival of the contrast agent bolus in the target vessel,\textsuperscript{13} automated detection of contrast bolus passage,\textsuperscript{14} and MR fluoroscopy to observe contrast passage.\textsuperscript{15}

Contrast-enhanced MRA (CEMRA) is advantageous in displaying detailed vessel anatomy and in reducing artifacts. MIP images can be reliably rendered in multiple projections. If the patient is incapable of holding his or her breath, then a slower acquisition can be done over several minutes with a longer/slower infusion of contrast agent. However, branch vessel detail may be degraded by respiratory motion. Subtraction of a precontrast from a postcontrast scan eliminates the signal from background tissue (assuming no patient motion). Additionally, it is possible to perform time-resolved 3D MRA studies to better differentiate arteries and veins.\textsuperscript{16}

Recently, testing has begun in humans of intravascular MR contrast agents that are retained within blood vessels and that selectively enhance the blood pool on T1-weighted MR images.\textsuperscript{17–21} AMI-227 (Advanced Magnetics), an ultrasmall iron particle, has been shown to enhance vessels within the abdomen and chest in the steady state.\textsuperscript{17,18} Another iron particle, NC100150 (Nycomed-Amersham), has recently entered human trials.\textsuperscript{19} MS-325 (EPIX Medical, Inc and Mallinckrodt Inc), a gadolinium-chelate that binds noncovalently to plasma albumin, has been shown to enhance the lower extremity and carotid vessels in the steady state as well as to enhance the arteries of the lower extremity during the first pass in a manner comparable to the currently available extracellular gadolinium chelates.\textsuperscript{22} The potential clinical applications of these agents await the results of large-scale multicenter clinical trials.

Coronary MRA is perhaps the most technically challenging area of MRA. To apply MRA techniques to the coronary arteries, additional technical obstacles had to be overcome, including compensation for respiratory and cardiac motion. Standard electrocardiographically (ECG) gated spin-echo and gradient-echo cine images only occasionally show portions of the coronary arteries, and these images are not adequate for detailed evaluation.\textsuperscript{23} Even a 1-second acquisition is too long to freeze cardiac motion. Instead, the acquisition can be “segmented” into blocks of phase-encoding steps, which are then interleaved to create an image.\textsuperscript{24} Segmentation within a single breath-hold of 10 to 20 seconds can reduce the time for data acquisition within each cardiac cycle to 100 ms or less, which is adequate to minimize cardiac motion artifacts, particularly in mid diastole or end systole. This strategy is known as segmented gradient-echo. Because the proximal coronary arteries are embedded in epicardial fat, chemical shift-selective fat saturation pulses are applied to reduce the fat signal. With this combination of techniques, the proximal portions of the left main, left anterior descending (LAD), and right coronary arteries (RCA) are routinely seen.

Recently, navigator echo methods have been applied to eliminate the need for breath-holding. Navigator echo is a motion-compensation technique that relies on tracking of an MR-detectable interface by the MR scanner. In the case of coronary MRA, the diaphragm-lung interface is typically used. With navigator gating for coronary MRA, MRI data are only accepted for image reconstruction when the navigator
echo indicates that the diaphragm or lung is within a certain operator-defined range (usually 3 to 5 mm). In this respect, navigator echoes perform a function similar to the bellows used for respiratory gating but provide more consistent results. Because many data are rejected, acquisition times are typically increased by a factor of 2 or more, depending on the patient's breathing patterns. However, if a fast imaging sequence such as segmented gradient echo is used, scan times are still reasonable, on the order of a few minutes at most. Image sharpness is as good as with breath-holding.25 Because navigator echoes permit signal averaging, the signal-to-noise ratio is increased, and higher in-plane spatial resolution can be obtained (eg, 0.5 mm). Moreover, navigator echoes ensure a consistent cardiac position from image to image, thereby minimizing misregistration artifact if one wishes to process a projection image.

The navigator gating technique discards data that are acquired when the diaphragm is incorrectly positioned and thus increases scan time. Adaptive correction of image location by use of real-time navigator measurement of diaphragm position is a potential method for further reducing slice registration errors.26,27 This method dynamically repositions the slice to follow the movement of the diaphragm and thus keeps the slice in a constant relation to the tissue of interest despite its motion. This technique, called navigator correction, can be combined with navigator gating to improve the efficiency of data acquisition. Navigator methods appear promising for both 2D and 3D acquisitions. Nevertheless, achievement of adequate spatial resolution, especially of small vessels, remains a challenge for MRA. Resolution below 0.5 mm may be required to achieve comparability with CA, especially in smaller (<3 mm) vessels. Motion makes this goal even more difficult to achieve, both by lengthening the scan time to compensate for respiratory motion (through navigators) and by narrowing the scanning window to a portion of the cardiac cycle. In the heart, beat-to-beat variations in cardiac position may exceed 1 mm, making submillimeter resolution even more challenging.

**MRA of the Carotid Artery**

MRA of the carotid bifurcation may be accomplished by several different MRI techniques, including 2D TOF,28 3D TOF,29 and CEMRA.30 When the merits of these sequences are compared, several generalizations can be made. 2D TOF provides a strong vascular signal, even when the arterial velocity is low. 2D TOF should be used to differentiate near and complete internal carotid artery (ICA) occlusion. 3D TOF provides superior, submillimeter resolution but at the expense of sensitivity to flow. The weak vascular signal of 3D TOF in slow-flow situations may be improved by the use of MOTSA.31 The original slices from a 3D TOF acquisition may allow one to see some features of plaque directly. CEMRA is quick and robust and is not impaired by slow-flow situations. CEMRA is a good choice for those patients who cannot maintain a position for prolonged periods. Unfortunately, it can only be effectively performed on MRI systems with enhanced gradient hardware.

There is currently no agreement as to which of these techniques is best. In part, this results from the fact that MRI instruments of different age and from different vendors have slightly different imaging characteristics. However, most experts use the MOTSA sequence for their primary interpretation. CEMRA is a very promising technique, but its diagnostic role awaits further validation. The aortic arch is best imaged by CEMRA, which has proven effective for the study of stenoses, aneurysms, and dissections.32,33 The carotid siphons and intracranial arteries are best imaged with MOTSA.

As a result of time constraints, only the cervical carotid artery is studied usually. A full study of the brain, intracranial vessels, bifurcation, and aortic arch requires one to position the patient 3 times (in the head, neck, and body coils) and necessitates that the patient spend more than 1 hour in the magnet. Such a comprehensive examination should be reserved for those with suspected tandem pathology on the basis of a prior duplex ultrasound (DUS) or MRA examination.

A typical study might begin with a low-resolution 2D TOF to locate the bifurcation. Sections should be acquired as high as the base of the skull so that patency of the ICA can be positively established in the carotid canal of the temporal bone. Lack of flow at this level implies an occlusion at the bifurcation, because the ICA lacks branches between these points. If a vertebral artery or other vessel is absent on the 2D images, the sequence should be repeated without presaturation bands to exclude reversal of flow. Then, a MOTSA acquisition can be acquired of the bifurcation or at any other level identified as stenotic on the 2D series. A MOTSA acquisition of the bifurcation takes ≈10 minutes, whereas 2D TOF of the entire length of the carotid artery takes ≈6 minutes. Technical factors that may affect MRA are excessive patient motion and the presence of ferromagnetic metal in the neck.

When carotid MRA is interpreted, special care should be taken to avoid overestimation of the stenosis severity in an area of turbulent flow.34 This phenomenon is encountered within the decelerating jet distal to a critical lesion. The tendency to overestimate stenosis severity is greatly reduced if one interprets the study from source images or reformations rather than from calculated projections.35–37 Overestimation is also reduced in 3D acquisitions, presumably because the more gradient structure and submillimeter voxels of 3D acquisition result in less phase dispersion, although many practitioners of 2D TOF have shown an excellent ability to properly grade severe lesions. Finally, the tendency to overestimate stenoses is reduced if one quantitatively measures the vessel diameters formally with a calibrated jeweler’s loupe rather than by “eyeball estimation.”

Table 1 summarizes many of the recent prospective comparisons of MRA and CA for the evaluation of the carotid bifurcation.37–49 The median sensitivity for a high-grade lesion was 93%, whereas the median specificity was 88%. These studies assumed CA to be the gold standard. A problem with this assumption is that many CA errors are considered errors in MRA. Because the reproducibility of CA itself is no better than 94%,48,50,51 one can conclude that the actual sensitivities and specificities may be better than those reported in these studies. In fact, preliminary data suggest that
noninvasive imaging may be more sensitive than CA in some instances.\(^\text{40}\) A comparison of CA, MRA, and DUS, with surgical specimens rather than CA used as the gold standard,\(^\text{52}\) found that DUS and MRA each correlate better with the endarterectomy specimen than does CA. This discrepancy can be attributed to the fact that the smallest diameter is often not appreciated by CA when the stenosis is elliptical or complex in shape. When CA and DUS are compared, it is particularly important to remember that the stenotic lumen is usually not circular, because Doppler velocities are determined by cross-sectional area rather than by diameter.\(^\text{53}\) Particularly important is the fact that the smallest diameter is often not appreciated by CA when the stenosis is elliptical or complex in shape. When CA and DUS are compared, it is particularly important to remember that the stenotic lumen is usually not circular, because Doppler velocities are determined by cross-sectional area rather than by diameter.\(^\text{53}\) Potentially important information about the shape of the plaque, available by MRA but not by CA, was not considered in the validation studies. The role of CA as a modality for assessing carotid stenosis has been established by its use in the large, multicenter clinical trials that have established endarterectomy as a successful stroke-prevention intervention. Multicenter clinical trials like the North American Symptomatic Carotid Endarterectomy Trial (NASCET) have never been undertaken with MRA rather than CA used to quantify stenosis. Therefore, MRA and DUS are assumed to be effective only to the degree that they agree with CA diameter measurements.

Kent et al\(^\text{45}\) examined the cost-effectiveness of various imaging strategies for a population of patients with symptomatic carotid artery disease. Use of DUS alone resulted in a quality-adjusted life expectancy (QALE) of 9.619 years. CA alone resulted in a QALE of 9.632 years, with an excess incremental cost-effectiveness ratio of $99 200 per quality-adjusted life year (QALY). The combination of DUS and MRA, followed by CA in the event of disparate results, maximized clinical outcome to 9.639 years at an incremental cost-effectiveness ratio of just $22 400 and was considered the optimum strategy. Obuchowski et al\(^\text{54}\) assumed a 20% prevalence of stenosis requiring surgery in patients who presented with a neck bruit. Three imaging strategies were examined: DUS followed in selected cases by MRA, DUS followed in selected cases by CA, and MRA alone. The QALE of the 3 strategies was virtually identical, whereas the incremental cost per QALY was $292 228, $747 000, and $770 000, respectively. DUS alone was not examined. These data again argue for the use of a combined noninvasive approach.\(^\text{55}\)

When asymptomatic patients are being screened, the advantages of noninvasive imaging are even greater. Kuntz et al\(^\text{56}\) concluded that use of CA in this population actually resulted in a greater incidence of stroke (7.12% in 5 years) than did DUS alone (6.35%), MRA alone (6.17%), or a combination of DUS and MRA followed by CA if necessary (6.34%).

Currently, it appears that the workup of suspected carotid artery stenosis should begin with DUS, performed by an experienced and preferably accredited laboratory. Although it has been suggested that the detection of >70% stenosis on DUS alone might be sufficient to recommend carotid endarterectomy, it may be appropriate to confirm this diagnosis with an additional imaging study to minimize the possibility of erroneous management. If so, a technically adequate MRA at an imaging facility with substantial experience in the performance and interpretation of carotid MRA can fulfill this role at the lowest cost and risk to the patient.\(^\text{57,58}\) CA should be considered when the results of DUS and MRA are discrepant, in cases of possible hairline patency (angiographic string sign), or where lesions are so atypical that they can only be understood by use of the high-resolution, selective vascular contrast of a catheter study.\(^\text{59}\) Supplementary imaging is especially advisable when results of DUS are technically limited. Typical limitations that necessitate additional imaging include the presence of a shadowing plaque, a deep course of the ICA, discordant gray-scale and Doppler measurements, and evidence of tandem lesions in cases in which these would alter patient management. Tandem lesions are suggested by the presence of a dampened and delayed waveform\(^\text{60}\) or unusually low diastolic velocities. To detect disease outside the ultrasonic window, MRA should also be considered in patients with atypical symptoms of cerebral ischemia and a DUS showing no disease or only mild disease. In some situations, the evaluation might begin with MRA (for example, when the MRA is performed as part of a brain MRI in patients with neurological symptoms).

**MRA of the Thoracic and Abdominal Aorta**

Standard T1-weighted MR techniques provide detailed anatomic information about the dimensions of the thoracic and abdominal aorta in multiple planes, can identify the presence of intraluminal thrombus within aneurysms or intimal flaps.

---

**TABLE 1. Summary of Carotid MRA Accuracy Studies**

<table>
<thead>
<tr>
<th>Reference</th>
<th>No. of Patients</th>
<th>TOF Sequence</th>
<th>Comparison</th>
<th>Stenosis Threshold</th>
<th>Sensitivity, %</th>
<th>Specificity, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anderson et al(^\text{37})</td>
<td>50</td>
<td>2D/3D</td>
<td>XRA/DUS</td>
<td>70</td>
<td>92</td>
<td>95</td>
</tr>
<tr>
<td>Mittl et al(^\text{40})</td>
<td>38</td>
<td>2D</td>
<td>XRA/DUS</td>
<td>70</td>
<td>92</td>
<td>75</td>
</tr>
<tr>
<td>Young et al(^\text{47})</td>
<td>70</td>
<td>2D/3D</td>
<td>XRA/DUS</td>
<td>70</td>
<td>86</td>
<td>93</td>
</tr>
<tr>
<td>Vanninen et al(^\text{40})</td>
<td>55</td>
<td>3D</td>
<td>XRA</td>
<td>70</td>
<td>93</td>
<td>88</td>
</tr>
<tr>
<td>Vogl et al(^\text{42})</td>
<td>120</td>
<td>3D</td>
<td>XRA</td>
<td>75</td>
<td>96</td>
<td>97</td>
</tr>
<tr>
<td>Kent et al(^\text{45})</td>
<td>81</td>
<td>3D</td>
<td>XRA/DUS</td>
<td>70</td>
<td>98</td>
<td>85</td>
</tr>
<tr>
<td>Nicholas et al(^\text{44})</td>
<td>40</td>
<td>2D/3D</td>
<td>XRA/DUS</td>
<td>70</td>
<td>92</td>
<td>98</td>
</tr>
<tr>
<td>Patel et al(^\text{43})</td>
<td>88</td>
<td>2D/3D</td>
<td>XRA/DUS</td>
<td>70</td>
<td>94</td>
<td>85</td>
</tr>
<tr>
<td>Levi et al(^\text{48})</td>
<td>45</td>
<td>2D/3D</td>
<td>XRA</td>
<td>70</td>
<td>95</td>
<td>77</td>
</tr>
<tr>
<td>Liberopoulos et al(^\text{40})</td>
<td>52</td>
<td>3D</td>
<td>XRA/DUS/surgery</td>
<td>60</td>
<td>100</td>
<td>80</td>
</tr>
<tr>
<td>Link et al(^\text{41})</td>
<td>40</td>
<td>3D</td>
<td>XRA</td>
<td>70</td>
<td>90</td>
<td>92</td>
</tr>
</tbody>
</table>

XRA indicates conventional angiography (x-ray angiography).
associated with aortic dissection, and can locate the main aortic branches, such as the iliac, main renal, and brachiocephalic arteries, in relation to aortic pathology. However, for projectional angiographic imaging and for detection of branch vessels and branch vessel disease, MRA is essential.

Abdominal aortic aneurysms (AAAs) have been an important focus of studies with MRA because of the desirability of replacing CA for preoperative evaluation. AAAs are typically diagnosed by physical examination, ultrasound, or computed tomography (CT), but angiographic imaging is often desired before preoperative evaluation. The important information that angiography must provide includes the upper extent of the aneurysm (including the relationship of the aneurysm to the renal arteries), the number and patency of the renal arteries, the patency of the mesenteric arteries, the inferior extent of the aneurysm (including the distance from the aneurysm to the aortic bifurcation), and aneurysmal or atherosclerotic disease of the iliac arteries. CA has been the standard approach for such preoperative evaluation. However, several studies of MRA for preoperative evaluation of AAAs have been published. Initial studies used noncontrast TOF MRA, often in conjunction with T1-weighted MRI. Although classification of aneurysms as suprarenal or infrarenal was good, detection of accessory renal arteries, renal artery stenosis, and iliac vascular disease was not adequate to justify replacement of CA. Limitations included the poor resolution in the slice-selection direction obtainable with 2D techniques and poor signal associated with slow flow in the ectatic aorta.61–63

More recently, interest has focused on CEMRA for AAA evaluation.64–67 The use of contrast enhancement both enables higher-resolution 3D imaging and provides high blood signal without the need for fast flow. In an initial study of 27 patients,66 CEMRA in combination with spin-echo and precontrast 3D TOF detected 7 of 9 accessory renal arteries, 8 of 9 renal artery stenoses, 4 of 4 celiac stenoses, and all iliac aneurysmal and stenotic disease. Interestingly, MRA correlated with surgical findings as well as CA for defining the proximal extent of the aneurysm. In a subsequent publication by the same group,67 MRA accurately defined the proximal extent of aneurysm in 87% of 38 patients. Sensitivities for iliofemoral occlusive and aneurysmal disease were 83% and 79%, respectively. Sensitivities for renal artery stenosis and accessory renal arteries were each 71%. In terms of surgical planning, MRA correctly predicted the cross-clamp site in 87%, the proximal anastomotic site in 95%, the need for renal revascularization in 91%, and the use of a bifurcated aortic prosthesis in 75%.

In a study of 43 subjects64 that used CEMRA in combination with noncontrast techniques, MRA correctly defined the maximum aneurysm diameter, as well as its proximal and distal extent, in all patients. For detection of aortic branch artery stenosis involving the celiac, superior mesenteric, renal, or iliac arteries, sensitivity was 94%, with specificity of 98%. Potential limitations of the CEMRA technique included the inability to define the severity of branch vessel stenosis and inadequate visualization of the inferior mesenteric artery. The latter can often be adequately evaluated at the time of surgery. The results suggest that performance of CEMRA is adequate to allow the treatment decision to be made between aortic tube graft and aorto-bifemoral graft. With the imminent introduction of aortic stent-graft devices, additional clinical study will be needed to determine whether MRA can substitute for CA in the decision-making process with regard to their use. Current resolution appears adequate to screen the renal arteries for disease. However, additional study is indicated before MRA can be relied on for evaluation of the mesenteric circulation. Recent improvements in scanner speed that allow contrast-enhanced aortography to be performed in <30 seconds, or within a breath-hold interval,13,68–70 and new blood-pool contrast agents currently under investigation21 may allow additional improvements in this area.

For the thoracic aorta, evaluation of aortic dissection in addition to aneurysmal disease has been another important focus of thoracic MRA. Conventional spin-echo and gradient-echo cine MRI have been shown to be quite effective modalities for evaluation of thoracic aortic disease.71–75 Hartnell et al,76 using noncontrast MRA, suggested that MRA allowed better demonstration of branch vessel stenoses, intimal flaps, and communications into false aneurysms than standard MRI of the aorta. CEMRA was used for evaluation in a range of disease of the thoracic aorta in another study77; in 30 patients with angiographic or surgical correlation, MRA successfully demonstrated dissection in 8 (including the correct type), coarctation in 3, and aneurysm in 10, as well as identifying 1 aberrant subclavian artery. Sensitivity for branch vessel stenosis in that study was 90%. MRA must be combined with conventional MRI to detect intramural hematoma or extravascular fluid collections.72

Renal MRA

Recent developments in renal MRA have led to a significant improvement in technical success rate and diagnostic accuracy. Specifically, 3D CEMRA methods have been shown to provide a more reliable depiction of renal artery morphology than noncontrast MRA techniques. In addition, preliminary work with adjunctive MRA techniques suggests that it is possible to determine the hemodynamic significance of renal artery disease by use of MRA and MRI. MRA methods will therefore likely play an important clinical role in the evaluation of patients with renovascular disease.

MRA was initially advocated as a method for imaging the renal arteries as early as 1990.78 The early approaches to renal MRA included 2 different methods, TOF and phase-contrast MRA. These methods are capable of evaluating the aorta and proximal renal arteries, but the more distal branches of the renal artery are obscured by motion artifact and the limited spatial resolution of these techniques. Early reports on the accuracy of renal artery MRA used acquisition methods that did not require the administration of intravenous or intra-arterial contrast; therefore, they were entirely noninvasive. However, the diagnostic accuracy reported by early investigators varied widely, with sensitivities for detecting renal artery stenosis ranging from 53% to 100% (Tables 2 and 3). The variable accuracy reported by different investigators is largely due to differences in technique, the presence of imaging artifacts, and limited spatial resolution. In addition, it
is generally understood that the noncontrast methods were associated with a relatively high technical failure rate. Many of the studies reporting the diagnostic accuracy of these nongadolinium methods did not include technically inadequate studies in the determinations of diagnostic accuracy. The early techniques were least successful in those patients with significant renal insufficiency, which is often associated with reduced renal blood flow. Finally, the limited spatial resolution of early nongadolinium methods resulted in a limited ability to detect accessory renal arteries, thereby further compromising the applicability of the technique.

Recent studies using CEMRA methods have demonstrated a higher technical success rate, as well as substantially improved diagnostic accuracy relative to noncontrast MRA methods. The improvement in image quality associated with CEMRA methods is largely due to improved spatial resolution, as well as an overall reduction in the number of artifacts. CEMRA methods have emerged as a robust, quick, clinically feasible technique to examine the renal arteries. CEMRA methods are also associated with fewer artifacts and can be performed in patients with poor renal blood flow or with renal insufficiency.

CEMRA methods combine a fast gradient-echo image with an intravenous infusion of a gadolinium-DTPA (diethylene-triamine penta-acetic acid) contrast agent. The rapid (typically <30 seconds) intravenous infusion of contrast results in an ~30-second window during which the gadolinium-DTPA is concentrated in the aorta and renal arteries. The signal enhancement created by the infusion of gadolinium-DTPA allows the use of a 3D gradient-echo acquisition that can be acquired far more quickly than with noncontrast MRA techniques. An entire set of 3D images is acquired during a single 30-second breath-hold.13,70,79

Several methods have been proposed to account for the differences in the arrival time of the gadolinium contrast at the region of interest. The contrast arrival time is dependent on the patient’s cardiac output, the severity of valvular insufficiency, and the presence of aortic pathology proximal to the renal arteries. Significant variations in the contrast arrival time after an intravenous bolus of gadolinium-DTPA have been demonstrated,13,80 and these differences may be associated with imaging artifacts.8 Therefore, several investigators have addressed techniques to precisely time the image acquisition commensurate with the arrival of contrast in the aorta.13–15 In addition, time-resolved CEMRA imaging techniques have been proposed to image the aorta and renal arteries, as well as to assess delays in renal enhancement due to renal artery stenosis.16 Although these new CEMRA methods have improved image quality, results at academic medical centers may not be uniformly applicable to clinical practice. For example, most of these studies have been performed on 1.5-T MR scanners equipped with faster gradient hardware and software. Therefore, whether these results will be equally successful with older MRI hardware or with low-field scanners remains to be determined.

Several investigators have evaluated the diagnostic accuracies of TOF MRA methods for depicting renal artery stenosis using CA as the reference standard78,81–89 (Table 2). 2D28 and 3D90 TOF approaches have been evaluated. In general, these methods require higher renal blood flow rates to ensure a technically adequate study. This requirement may partly explain the variable technical success and diagnostic accuracy reported with the TOF methods. For example, studies in younger patients with systemic hypertension and normal renal function84 tend to yield better results than studies in patients with hypertension and renal insufficiency.83 In addition, the variable sensitivity reported with these methods is also due in part to variations in the localization of lesions within the renal arteries. For example, studies with a majority of proximal lesions1 tended to show better accuracy than studies with more lesions in the mid and distal renal arteries.14 In general, TOF methods are only useful for evaluating the proximal renal arteries.

The reports of diagnostic accuracy for phase-contrast techniques also vary widely, and the data are summarized in Table 3. 2D and 3D phase-contrast methods have been evaluated,89,91–100 Phase-contrast methods generally show better visualization of the distal renal arteries because the technique is essentially a subtraction method. However, phase-contrast methods are more sensitive to the acquisition

<table>
<thead>
<tr>
<th>Reference</th>
<th>No. of Patients</th>
<th>Sensitivity, %</th>
<th>Specificity, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kim et al28</td>
<td>25</td>
<td>100</td>
<td>92</td>
</tr>
<tr>
<td>Kent et al82</td>
<td>33</td>
<td>100</td>
<td>94</td>
</tr>
<tr>
<td>Yucel et al81</td>
<td>16</td>
<td>100</td>
<td>93</td>
</tr>
<tr>
<td>Debatin et al83</td>
<td>33</td>
<td>53</td>
<td>97</td>
</tr>
<tr>
<td>Servois et al85</td>
<td>48</td>
<td>70</td>
<td>78</td>
</tr>
<tr>
<td>Strotzer et al84</td>
<td>55</td>
<td>100</td>
<td>90</td>
</tr>
<tr>
<td>Hertz et al86</td>
<td>16</td>
<td>91</td>
<td>76</td>
</tr>
<tr>
<td>Fellner et al87</td>
<td>46</td>
<td>100</td>
<td>89</td>
</tr>
<tr>
<td>Holland et al104</td>
<td>63</td>
<td>74</td>
<td>98</td>
</tr>
<tr>
<td>Laissy and Shouman-Claeys88</td>
<td>36</td>
<td>94</td>
<td>98</td>
</tr>
<tr>
<td>Duda et al89</td>
<td>22</td>
<td>73</td>
<td>47</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Reference</th>
<th>No. of Patients</th>
<th>Sensitivity, %</th>
<th>Specificity, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Debatin et al83</td>
<td>33</td>
<td>87</td>
<td>97</td>
</tr>
<tr>
<td>Vock et al81</td>
<td>20</td>
<td>89</td>
<td>100</td>
</tr>
<tr>
<td>Grist82</td>
<td>35</td>
<td>89</td>
<td>95</td>
</tr>
<tr>
<td>Gedroyc et al83</td>
<td>50</td>
<td>83</td>
<td>97</td>
</tr>
<tr>
<td>Gedroyc et al101</td>
<td>74</td>
<td>84</td>
<td>91</td>
</tr>
<tr>
<td>Loubeyre et al84</td>
<td>46</td>
<td>100</td>
<td>90</td>
</tr>
<tr>
<td>Loubeyre et al85</td>
<td>12</td>
<td>100</td>
<td>40</td>
</tr>
<tr>
<td>De Cobelli et al102</td>
<td>50</td>
<td>94</td>
<td>94</td>
</tr>
<tr>
<td>de Haan et al97</td>
<td>38</td>
<td>93</td>
<td>95</td>
</tr>
<tr>
<td>Duda et al99</td>
<td>22</td>
<td>78</td>
<td>76</td>
</tr>
<tr>
<td>Silverman et al100</td>
<td>36</td>
<td>100</td>
<td>93</td>
</tr>
<tr>
<td>Schoenberg et al100</td>
<td>19</td>
<td>100</td>
<td>93</td>
</tr>
<tr>
<td>Wasser et al100</td>
<td>13</td>
<td>92</td>
<td>75</td>
</tr>
</tbody>
</table>
parameters selected by the operator, including the selection of the velocity-encoding value for imaging the renal arteries. Because the velocity of renal artery blood flow varies widely between patients, those approaches that incorporated a rapid-velocity prescan tended to perform better than approaches that used an “educated guess” at the correct velocity-encoding value. Phase-contrast techniques tend to take longer to acquire MRA images than TOF methods. In addition, phase-contrast methods require a longer TE than TOF techniques. The result is a greater degree of signal loss due to turbulent flow at or distal to a stenosis. The presence of this signal loss, however, correlates with the functional significance of the stenosis.

Noncontrast TOF and phase-contrast methods are limited by the accuracy with which they detect the small accessory renal arteries owing to their inferior spatial resolution. Most reports detected fewer than 50% of accessory renal arteries. The inability to detect these vessels is primarily due to the small field of view used for the TOF and phase-contrast methods, because accessory renal arteries often lie outside the imaging volume when an axial slab is used for imaging.

The development of breath-hold CEMRA has been instrumental for the successful application of this technique for renal MRA. Several investigators have reported techniques for breath-hold CEMRA, with substantially improved spatial resolution and image quality relative to non–breath-hold methods. The technical success rate of breath-hold renal CEMRA was addressed by Wilman et al in a study of 25 consecutive patients. The authors found that visualization of the renal arteries was adequate in 24 patients (96%).

Because the CEMRA methods have been developed more recently, fewer studies are available for analysis; these are summarized in Table 4. In a prospective study comparing the accuracy of breath-hold renal CEMRA relative to digital subtraction x-ray angiography (DSA), Hany et al demonstrated a high accuracy for identifying renal lesions of >50% diameter stenosis (sensitivity 93%, specificity 98%). The authors performed image analysis on 78 main and 11 accessory renal arteries. The CEMRA methods allowed better visualization of the accessory renal arteries relative to non-enhanced techniques, primarily because of improved spatial resolution associated with the breath-hold acquisition, as well as a larger imaging field of view. The authors found that the sensitivity and specificity for detecting these renal artery stenoses were 93% and 98%, respectively. Snidow et al examined 32 patients using breath-hold CEMRA and compared the results with CA. The CEMRA protocol was intended primarily to evaluate the aorta and iliac inflow, with the parameters selected to include a screening study of the renal arteries. These authors found a sensitivity and specificity of 100% and 89%, respectively, for obstructive lesions of the main renal arteries. Examination of the accuracy for depiction of accessory renal artery occlusive disease demonstrated a sensitivity of 100% and a specificity of 62%. The relatively lower specificity in that study may be related to the selection of acquisition parameters, which were primarily chosen to ensure evaluation of the aorta and iliac inflow vessels, with inadequate spatial resolution to depict small accessory renal arteries.

It is widely recognized that CA has several limitations, although it is the accepted reference standard for identifying renal artery stenosis. CA is a projectional method that must rely on oblique imaging planes to profile a stenosis that appears “en face” in the renal artery. This limitation makes it difficult to precisely profile ostial lesions or to delineate lesions associated with an asymmetrical luminal narrowing. In addition, the CA technique provides information predominately about the vessel lumen, poststenotic dilatation, and relative renal perfusion. DSA does not provide an assessment of alterations in the renal artery blood flow or pressure gradients across a renal artery stenosis. Data regarding the pressure gradient and renal vein renin production may be obtained by use of additional catheter placements under fluoroscopic guidance.

With these limitations of CA in mind, Wasser et al used a measurement of the pressure gradient across the renal artery stenosis as a reference standard to compare the diagnostic accuracy of ECG-gated 3D phase-contrast MRA and conventional DSA. The authors used a pressure gradient of 15 mm as the definition of a significant stenosis. DSA detected 10 of 13 stenoses (sensitivity 77%, specificity 92%). MRA demonstrated the loss of distal flow signal intensity in 12 of these stenoses, for a sensitivity of 92% and a specificity of 75%. There was no significant difference between the 2 modalities in grading hemodynamic significance of the stenosis, although the ability to detect differences between the 2 methods was limited by the small subject population.

Schoenberg et al, in a study involving ECG-gated phase-contrast MR evaluation of alterations in renal artery blood-flow waveforms, demonstrated changes in the flow waveforms in patients with significant (>50% diameter) renal artery stenoses on CA. The authors, using techniques analogous to Doppler ultrasound measurement of flow waveforms, demonstrated a sensitivity of 100% and a specificity of 93% for identifying stenoses. The authors relied on a qualitative assessment of the renal artery flow waveform, as well as quantitative assessment of changes in the time to maximal flow after the ECG trigger. These data suggest that physiological waveform alterations may indeed be detected by MR phase-contrast flow measurements.

In an effort to step beyond diagnostic correlation studies comparing MRA with CA or pressure measurements, Prince et al sought to demonstrate that findings on MRA studies could be used to predict the likelihood of success after revascularization, as measured by a reduction in the need for

---

**TABLE 4. MRA for Renal Artery Stenosis: Contrast-Enhanced Techniques**

<table>
<thead>
<tr>
<th>Reference</th>
<th>No. of Patients</th>
<th>Sensitivity, %</th>
<th>Specificity, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kaufman et al</td>
<td>27</td>
<td>89</td>
<td>98</td>
</tr>
<tr>
<td>Holland et al</td>
<td>63</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>Snidow et al</td>
<td>82</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>Hany et al</td>
<td>39</td>
<td>93</td>
<td>98</td>
</tr>
<tr>
<td>Rieumont et al</td>
<td>30</td>
<td>100</td>
<td>71</td>
</tr>
<tr>
<td>Steffens et al</td>
<td>50</td>
<td>98</td>
<td>96</td>
</tr>
</tbody>
</table>

---

50% diameter) renal
antihypertensive medications or an improvement in renal function after revascularization. The investigators performed a retrospective study that combined CEMRA for detection of renal artery stenosis with phase-contrast MRA for prediction of patient outcome after revascularization. They found that signal loss distal to a stenosis on the phase-contrast MRA examinations was significantly more prominent in those patients with hemodynamically significant stenoses. Poststenotic arterial dilatation, reduced renal length, and reduction in parenchymal renal thickness were good predictors of patient outcome after revascularization.

In summary, recent developments in renal MRA have led to a significant improvement in the technical success rate and diagnostic accuracy of the methods. Specifically, 3D breath-hold CEMRA methods have been shown to provide a more reliable depiction of renal artery morphology. The new methods are easier to perform, are capable of providing higher spatial resolution, and are associated with fewer artifacts. Although the older noncontrast (phase-contrast and TOF) methods are less reliable for demonstrating renal artery anatomy, the phase-contrast techniques have several important attributes that may make it feasible to demonstrate the hemodynamic significance of a renal artery stenosis. Recent work in MRA has demonstrated that MRA methods may be used to demonstrate the hemodynamic significance of renal artery stenosis and predict patient outcome in response to revascularization therapy.

Peripheral MRA
Peripheral arterial occlusive disease is a local manifestation of the systemic atherosclerotic process. Patients present with a wide range of symptoms depending on the severity of disease and the degree of peripheral ischemia. Treatment options depend on the underlying arterial anatomy and include percutaneous balloon angioplasty, bypass surgery, and amputation. Patients with focal disease will most likely have percutaneous balloon angioplasty, whereas those with long occluded segments or diffuse disease usually undergo surgery. Regardless of the type of revascularization, a preoperative road map is required before intervention, and for the past 3 decades, CA has served this purpose. The recent development and refinement of peripheral vascular MRA has challenged catheter-based approaches and at some centers has replaced it in the evaluation and surgical planning of patients with symptomatic peripheral vascular disease. Its use in individual institutions depends on the expertise of the MR radiologist performing the examination and the ability of vascular surgeons and interventional radiologists to plan interventions based on the MRA images.

Most of the initial work on peripheral MRA was performed on infrarenal vessels with a local surface coil used as an RF receiver. These vessel segments provide an ideal environment for MRA because they are not tortuous, they are unidirectional, and they have steady, nonturbulent flow. The first reports of imaging peripheral vessels with MRA began to appear in the late 1980s and early 1990s. Although most of these were feasibility studies, peripheral MRA was found to be remarkably accurate at depicting peripheral vascular disease compared with CA. 2D TOF was the pulse sequence most commonly used, sometimes combined with phase-contrast MRA.

In 1991, Mulligan et al compared 2D TOF peripheral MRA with CA and ultrasound in 12 patients. There was 71% agreement between MRA and CA. Other investigators used 2D TOF MRA and had similar results. In 1992, Owen et al studied tibial vessels with MRA and CA in patients with severe peripheral vascular disease. Surprisingly, more arterial segments were seen on MRA than on contrast angiography, with these “angiographically occult” runoff vessels confirmed at surgery. Surgical plans were also changed on the basis of the additional information obtained from MRA. The results of the study by Owen et al provoked much discussion and led to several follow-up investigations. In a study of 51 patients, Carpenter et al compared that in 48% of vessel segments, MRA demonstrated information that was not present on CA. The additional information led to a change in the surgical plan for 22% of the patients. Similar investigations by McCauley et al in 1994, McDermott et al in 1995, and Huber et al in 1997 also demonstrated more vascular segments, as did a multicenter trial performed by the American College of Radiology. In the latter study, 155 patients from 6 hospitals were studied with similar imaging hardware and software. Peripheral MRA and CA were compared with the gold standard of intraoperative x-ray angiography. There was 81% agreement between CA and MRA, whereas MRA was found to be more sensitive in the identification of diseased tibial runoff vessels. This was especially true in the most distal segments of the ankle and foot. The superiority of MRA over CA in imaging pedal vessels is a result of the fundamental differences between the 2 techniques. CA requires opacification of vessels with contrast in order for them to be visualized radiographically, whereas 2D TOF MRA only needs flowing blood. By the appropriate selection of imaging parameters, blood flows < 10 cm/s can be readily detected by MRA. It has been shown that because of the high sensitivity of MRA for visualizing tibial vessels, the addition of this test can be cost-effective.

Because of the initial success of 2D TOF MRA in visualizing the infrarenal arteries, many investigators used the same technique to image the abdominal and pelvic vessels. Unfortunately, the results in these investigations varied widely among institutions. Snidow and colleagues compared CA and 2D TOF MRA in 50 patients and only found 52% agreement between techniques. The same group in another comparison study concluded, “For evaluation of symptomatic lower extremity ischemia, 2D TOF MRA cannot be considered a reliable substitute for x-ray angiography.”

Other centers have demonstrated better success using similar techniques. Carpenter et al performed a 2D TOF MRA and CA study in 47 patients and formulated separate interventional plans for each modality. The results of the 2 imaging studies were identical in 41 (87%) of 47 patients. MRA accurately detected patent and occluded arterial segments (sensitivity 99.6%; specificity 100%; positive predictive value 100%; negative predictive value 98.6%). Therapeutic plans based on either MRA or CA were identical for
every patient. MRA imaging times, however, were >2 hours and only included the pelvis and a single extremity. In addition, almost 25% of patients needed a second MRA because of flow-related artifacts. This showed that although it was possible to plan revascularization with 2D TOF MRA, it was not clinically practical.

To overcome the flow-related artifacts and prolonged imaging times that have plagued 2D TOF MRA, several variations in imaging techniques have been examined. Although helpful, these approaches failed to accomplish the goal of a robust technique for an aortogram and runoff examination that could be performed in <2 hours.

Although peripheral MRA has proved accurate in the depiction of focal peripheral vasculature, the examination can be extremely lengthy. To visualize small tibial vessels, surface coils are needed that can image the entire lower extremity during a single series. At present, the examination requires several overlapping series that, when pieced together, produce a vascular road map. The surface coil must be repositioned after each series is performed. Extended-length surface coils have been developed in the hope of speeding examination times without any loss of spatial resolution.

As with renal, aortic, and carotid angiography, CEMRA has proved helpful for peripheral MRA. In 1992, Lossef et al. evaluated the efficacy of gadopentetate dimeglumine (Gd-DTPA) in the lower extremities by use of a flow phantom, 7 healthy volunteers, and 7 patients with peripheral vascular disease. It was determined that the intravenous administration of a bolus of Gd-DTPA improved MRA quality in the phantom, volunteers, and patients. However, in some cases, venous overlap, imaging artifacts, and suboptimal visualization of subtle lesions limited interpretation. In a feasibility study, Prince and colleagues used 3D CEMRA to evaluate the abdominal aorta and the renal, visceral, and iliac arteries in 16 patients. They found that by imaging dynamically during the arterial phase of a 5-minute injection, preferential arterial enhancement could be achieved with minimal enhancement of venous structures and no flow-related artifacts. Additional studies have confirmed the speed and accuracy of peripheral CEMRA. Poon et al. compared the diagnostic accuracy of CEMRA for the measurement of percent stenosis in the iliac arteries rather than using an arbitrary cutoff for significant disease. A recent refinement of CEMRA for peripheral vascular disease has used a stepping table with digital subtraction, comparable to the technique used to perform a peripheral runoff examination with x-ray angiography, although at lower resolution. Those studies showed that whereas stenosis measurements for 2D TOF MRA were significantly different from those obtained on CA, the values obtained with CEMRA were statistically indistinguishable from CA. The use of CEMRA for the inflow (aortoiliac) vessels has become commonplace as part of a complete peripheral MRA examination at many centers.

The timing of arrival of the contrast bolus with imaging is critical. After the arterial first pass of a bolus of MR contrast, there is rapid equilibration followed by renal excretion. If imaging is delayed and the arterial phase is missed, venous and soft tissue enhancement occurs. In addition, sometimes there is a need for several areas to be imaged sequentially, which would require multiple administrations of contrast. For these reasons, subtraction techniques were developed that allow for multiple injections of contrast. These techniques, alone or in combination with moving-table technology, may permit further inroads of CEMRA into TOF MRA of the extremity. Initial results have also been reported for peripheral MRA with blood-pool contrast agents. Dynamic images comparable to conventional agents were obtained at substantially lower doses. Steady-state images demonstrated enhancement of the blood pool for up to 50 minutes, and vessels as small as 1 mm in diameter were seen. Overlap of arteries and veins on MIP images remains an issue for these agents. Possible approaches include the use of multplanar reformating and development of automated segmentation techniques.

After initial feasibility and comparison studies were performed, investigators studied the utility of MRA to plan bypass surgery. This requires imaging from the infrarenal aorta through the plantar arch. In the largest series reported to date, 80 consecutive patients successfully underwent peripheral bypass surgery based solely on a preoperative MRA. Most authors suggest that this should be attempted only at centers with expertise in peripheral MRA and where these techniques have been validated against the gold standard of CA.

MRA has proved useful in a variety of disease states in addition to peripheral vascular disease, including popliteal entrapment and aneurysms. It has also been found useful in identifying patient tibial vessels before fibular graft harvesting.

There has been and continues to be much progress since peripheral MRA was introduced almost a decade ago, and at selected centers with high expertise, MRA has been able to replace CA for patients with symptomatic peripheral occlusive disease. If properly performed and interpreted, MRA can replace diagnostic CA by combining the speed of CEMRA of the iliac and femoral arteries with the high sensitivity of 2D TOF of the tibial arteries. The eventual role of CEMRA with multiple injections, moving-table technology, and/or blood-pool contrast agents remains to be determined.

MRA of the Coronary and Pulmonary Arteries

Of all vascular regions, coronary MRA has been considered the most technically demanding, largely because of the small size of the coronaries (3 to 4 mm in diameter), their tortuous and complex pathway, the abundant signal from surrounding epicardial fat, and the significant motion associated with both respiration and cardiac motion. Tremendous progress has been made since the first clinical reports of human coronary MRA were published in 1991, yet current clinical applications are limited to evaluation of anomalous coronary arteries and assessment of coronary artery bypass graft patency. Virtually all current applications of native coronary MRA use ECG gating to minimize motion artifacts related to cardiac motion. Early attempts at coronary MRA with conventional spin-echo MR were met with limited success. In
1984, Lieberman et al.\textsuperscript{170} were able to visualize portions of the coronaries in only 7 of 23 patients, whereas Paulin and colleagues\textsuperscript{23} identified the origin of the left main coronary artery in all 6 subjects examined and the RCA ostium in 4 of the 6 individuals. No coronary stenoses were identified in either report.

Recent advances in both MR hardware and software have led to the development of imaging strategies that enable visualization of the proximal portion of the major epicardial coronary arteries in the vast majority of subjects. Although not yet standardized, current methods for coronary MRA combine ECG gating with fast imaging sequences and techniques for suppressing signal from surrounding epicardial fat and for myocardial suppression of artifacts related to respiratory motion.

The greatest body of clinical investigation has been performed with segmented gradient-echo sequences, first described in an animal model by Burstein\textsuperscript{171} and subsequently reported in humans by Edelman et al.\textsuperscript{24} This method reduces the number of heartbeats required to complete an MR image and uses an acquisition window of 100 to 150 ms during mid diastole, a period of relative bulk cardiac diastasis but with relatively high coronary blood flow. Typically, 8 interleaved phase-encoding steps are acquired during each cardiac cycle, necessitating 16 successive heartbeats to complete a 128×256 matrix. This allows for the acquisition of a single 2D image during a single 15- to 20-second breath-hold, thereby minimizing respiratory motion artifacts. A slice thickness of 3 to 4 mm and a 240-mm field of view (in-plane spatial resolution of \(\approx 1.9\times0.9\) mm) are often used. Forty or more breath-holds may be needed to complete a coronary MRA study. As with other gradient-echo approaches, rapidly moving laminar blood flow appears as bright signal because of the inflow of unsaturated protons. Focal areas of turbulence, such as those caused by coronary stenoses, appear dark owing to signal loss from dephasing.

### TABLE 5. Current Experience With Coronary MRA in Humans

<table>
<thead>
<tr>
<th>Reference</th>
<th>Technique</th>
<th>Respiratory Suppression</th>
<th>No. of Patients</th>
<th>Vessels Visualized, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Edelman et al.\textsuperscript{174}</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>22</td>
<td>100 (ostia)</td>
</tr>
<tr>
<td>Manning et al.\textsuperscript{180}</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>25</td>
<td>76–100</td>
</tr>
<tr>
<td>Pennell et al.\textsuperscript{182}</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>26</td>
<td>76–95</td>
</tr>
<tr>
<td>Oshinski et al.\textsuperscript{181}</td>
<td>2D seg. GRE</td>
<td>Prospective navigator gating</td>
<td>10</td>
<td>100</td>
</tr>
<tr>
<td>Hofman et al.\textsuperscript{179}</td>
<td>2D seg. GRE, 3D GRE</td>
<td>Retrospective navigator gating</td>
<td>10</td>
<td>80–100</td>
</tr>
<tr>
<td>Paschal et al.\textsuperscript{185}</td>
<td>3D, TOF, MT</td>
<td>None</td>
<td>14</td>
<td>57–100</td>
</tr>
<tr>
<td>Li et al.\textsuperscript{172}</td>
<td>3D GRE, MT</td>
<td>None</td>
<td>14</td>
<td>78–100</td>
</tr>
<tr>
<td>Li et al.\textsuperscript{184}</td>
<td>3D GRE</td>
<td>Retrospective navigator gating</td>
<td>13</td>
<td>100</td>
</tr>
<tr>
<td>Post et al.\textsuperscript{186}</td>
<td>3D GRE</td>
<td>Retrospective navigator gating</td>
<td>20</td>
<td>96</td>
</tr>
<tr>
<td>Wielopolski and de Feyter\textsuperscript{178}</td>
<td>3D seg. EPI</td>
<td>Breath-hold</td>
<td>10</td>
<td>100</td>
</tr>
<tr>
<td>Goldfarb et al.\textsuperscript{183}</td>
<td>3D, contrast</td>
<td>Breath-hold</td>
<td>4</td>
<td>100</td>
</tr>
</tbody>
</table>

Seg. indicates segmented k space; GRE, gradient echo; MT, magnetization transfer; and EPI, echo-planar imaging.

Several other general approaches to coronary MRA have been described, including 3D TOF\textsuperscript{172} methods that offer enhanced signal along with reconstruction advantages for visualization of the tortuous coronary vasculature. However, 3D imaging is prone to blurring,\textsuperscript{173} because scan times usually far exceed that of a single breath-hold. MR subtraction methods that involve selective tagging of blood in the aortic root and suppression of background tissue have also been described.\textsuperscript{174} This approach has great appeal for defining the proximal coronary artery but may be more limiting for more distal regions. Other imaging techniques include spiral acquisitions\textsuperscript{175} and echo-planar imaging.\textsuperscript{176–178}

The identification of the origin and proximal/mid portions of the native coronary arteries is currently possible in the vast majority of individuals. A summary of recent coronary MRA studies reporting on visualization of coronary arteries in healthy volunteers and patients with coronary artery disease is presented in Table 5. Both 2D\textsuperscript{13,179–182} and 3D\textsuperscript{172,178,179,183–186} approaches have been used at multiple centers with similar results. The origin and proximal coronary arteries are visualized in >95% of subjects. In general, more extensive portions of the LAD and RCA are seen.\textsuperscript{180–182} In angiographically normal vessels, proximal coronary MRA diameter corresponds well to CA.\textsuperscript{180,187}

The ability of coronary MRA to reliably identify the native vessels, coupled with the inherent 3D representation of MR data, renders this technique ideally suited for the noninvasive identification and characterization of anomalous coronary arteries. Although they are rare, occurring in only 0.6% to 1.2% of adults referred for CA,\textsuperscript{188} anomalies associated with passage of the anomalous vessel between the aorta and pulmonary artery can result in impaired myocardial perfusion and have been associated with sudden death.\textsuperscript{189} Although the presence of anomalous vessels is well defined by CA, their passage anterior or posterior to the aorta can sometimes be difficult to appreciate. With the use of breath-hold 2D TOF
approaches, accurate identification and the course of anomalous coronary vessels have been well described.190–192 In many studies, coronary MRA has been found to be superior to CA because it has identified prior misclassification (by CA) of anomalous vessels.190,192 These data strongly support a clinical role for coronary MRA as a noninvasive tool for defining the anatomic course in patients with known or suspected anomalous coronary vessels.

Although coronary MRA is sufficiently advanced to allow for definition of anomalous coronary vessels, the data do not support a general role for identification of native vessel stenoses. The spatial resolution of the most widely tested breath-hold coronary MRA approach precludes quantitative coronary MRA. Most series have used the visual identification of a prominent signal “void” as being indicative of a diameter stenosis of \( \geq 50\% \). Although the technique has been shown to accurately identify proximal and mid-coronary stenoses in some centers,193–195 with good correlation with location of stenoses and with length of the coronary lesion slightly overestimated by MR,194 data from other centers have been more variable (Table 6).196–199

Much of the variability in coronary MRA success is likely related to the lack of sequence standardization (data acquisition interval and TE) and the dependence on the breath-hold for respiratory artifact suppression. Frequently, \( \geq 40 \) sustained breath-holds are required to complete a scan, a requirement that often leads to both patient and operator fatigue. Improvements in data acquisition with the use of either fewer breath-holds or free-breathing strategies promise to simplify the approach, improve spatial resolution, and likely improve accuracy. However, a growing obstacle to coronary MRA is the use of intracoronary stents. Although not a contraindication for MRA200 (up to field strengths of 1.5 T), local signal voids induced by the stent preclude assessment of the stented portion of the vessel.201

The relatively stationary position of saphenous vein and internal mammary artery bypass grafts offsets many of the technical challenges of native vessel coronary MRA. Furthermore, their relatively straight and predictable course has allowed imaging of bypass grafts with conventional ECG-gated spin-echo202–205 and gradient-echo 202,206,207 MR techniques. For most of these investigations, contiguous transverse images are obtained at levels corresponding to that expected for the bypass graft. The graft is then characterized as “patent” if the normal signal void (spin echo) or bright signal (gradient echo) of laminar blood flow is seen in at least 2 contiguous anatomic levels in the expected region of the bypass graft. If a signal void is seen at only 1 level, a graft is considered “indeterminate,” and if no signal of laminar blood flow is seen, the graft is considered occluded. Both spin-echo and gradient-echo approaches have a similar accuracy (Table 7). The addition of CEMRA208–210 may further improve accuracy for bypass grafts. In an unusual case,211 coronary MRA was able to correctly identify the bypass graft and confirm patency of the native LAD distal to the graft anastomosis when CA failed to do so.

### Table 6. Summary of Current Coronary MRA Studies for the Detection of Significant Coronary Artery Disease (>50% Diameter Stenosis)

<table>
<thead>
<tr>
<th>Reference</th>
<th>Technique</th>
<th>Respiratory Suppression</th>
<th>No. of Patients</th>
<th>No. (%) Vessels With Stenoses</th>
<th>Sensitivity, %</th>
<th>Specificity, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Manning et al193</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>39</td>
<td>52 (35)</td>
<td>90 (71–100)</td>
<td>92 (78–100)</td>
</tr>
<tr>
<td>Duerinckx and Urman190</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>20</td>
<td>27 (34)</td>
<td>63 (0–73)</td>
<td>(37–82)</td>
</tr>
<tr>
<td>Post et al194</td>
<td>3D GRE</td>
<td>Retrospective navigator gating</td>
<td>20</td>
<td>21 (27)</td>
<td>38 (0–57)</td>
<td>95 (85–100)</td>
</tr>
<tr>
<td>Pennell et al194</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>39</td>
<td>55 (35)</td>
<td>85 (75–100)</td>
<td></td>
</tr>
<tr>
<td>Hu et al198</td>
<td>2D Interleaved</td>
<td>Breath-hold</td>
<td>23</td>
<td>27</td>
<td>63</td>
<td></td>
</tr>
<tr>
<td>Post et al196</td>
<td>2D seg. GRE</td>
<td>Breath-hold</td>
<td>35</td>
<td>35 (28)</td>
<td>63 (0–100)</td>
<td>89 (73–96)</td>
</tr>
<tr>
<td>Muller et al197</td>
<td>3D GRE</td>
<td>Retrospective navigator gating</td>
<td>35</td>
<td>54 (31)</td>
<td>83 (50–100)</td>
<td>94 (85–100)</td>
</tr>
</tbody>
</table>

Seg. indicates segmented k space; GRE, gradient echo.

Sensitivity and specificity values are reported as mean (individual vessel range).

### Table 7. Summary of Studies Evaluating Coronary MRA for the Assessment of Graft Patency After Aortocoronary Bypass Grafting

<table>
<thead>
<tr>
<th>Reference</th>
<th>No. of Patients</th>
<th>No. of Grafts</th>
<th>% Patent</th>
<th>Sensitivity, %</th>
<th>Specificity, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Spin echo</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>White et al205</td>
<td>25</td>
<td>72</td>
<td>69</td>
<td>86</td>
<td>72</td>
</tr>
<tr>
<td>Rubinstein et al204</td>
<td>20</td>
<td>47</td>
<td>62</td>
<td>90</td>
<td>72</td>
</tr>
<tr>
<td>Jenkins et al203</td>
<td>16</td>
<td>41</td>
<td>63</td>
<td>89</td>
<td>73</td>
</tr>
<tr>
<td>Galjee et al202</td>
<td>47</td>
<td>84</td>
<td>74</td>
<td>98</td>
<td>85</td>
</tr>
<tr>
<td>Gradient echo</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>White et al206</td>
<td>28</td>
<td>28</td>
<td>52</td>
<td>93</td>
<td>86</td>
</tr>
<tr>
<td>Aurigemma et al207</td>
<td>45</td>
<td>45</td>
<td>73</td>
<td>88</td>
<td>100</td>
</tr>
<tr>
<td>Galjee et al202</td>
<td>47</td>
<td>84</td>
<td>74</td>
<td>98</td>
<td>88</td>
</tr>
</tbody>
</table>
TABLE 8. Summary of Recommendations for MRA

<table>
<thead>
<tr>
<th>Vascular Bed</th>
<th>Utility for Diagnosis</th>
<th>Intervention Planning</th>
<th>Contrast Helpful?</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carotid</td>
<td>For atherosclerotic occlusive disease</td>
<td>Complementary to ultrasound</td>
<td>Under investigation</td>
</tr>
<tr>
<td>Aorta</td>
<td>For a wide range of aortic pathology</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Renal/visceral</td>
<td>For atherosclerotic occlusive disease</td>
<td>Not fully validated</td>
<td>Yes</td>
</tr>
<tr>
<td>Peripheral</td>
<td>For atherosclerotic occlusive disease</td>
<td>Yes; moving table maximizes efficiency</td>
<td>Yes, especially for proximal vessels</td>
</tr>
<tr>
<td>Pulmonary</td>
<td>Under investigation; substitute for CT in patients with contraindications to x-ray contrast</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Coronary</td>
<td>Congenital anomalies; under investigation for atherosclerotic occlusive disease</td>
<td>No</td>
<td>Under investigation</td>
</tr>
</tbody>
</table>

As with intravascular stents, a major obstacle for bypass graft imaging is the local signal loss or artifact associated with implanted metallic objects such as hemostatic clips and ostial graft rings. Artifacts from graft markers or clips may be indistinguishable from signal “voids” related to blood flow through a stenosis in the bypass graft. In addition, grafts with very tight stenoses, in which blood flow would be expected to be low, often result in insufficient contrast to characterize a graft as patent.224 A determination of the ultimate role of MRA in bypass graft evaluation in clinical practice will require further study.

Although current coronary MRA strategies are acceptable for evaluation of anomalous coronary arteries and bypass graft patency, the major impact of coronary MRA will likely be felt when methods are sufficiently robust to allow for evaluation of the native coronary arteries. Current advances in MR hardware and software have focused on 2 different areas: suppression of respiratory motion artifacts and CEMRA.

Although breath-holding has been widely used to minimize respiratory motion artifacts, this approach requires significant patient training and cooperation, as well as operator expertise. Breath-hold variability introduces slice registration errors, and the breath-hold duration limits the use of potential enhancements, such as 3D acquisitions, signal averaging, and higher-resolution coronary MRA. Imaging during free (unrestricted) breathing offers obvious advantages over breath-holding. A novel technique for minimizing respiratory-related motion is with the use of real-time navigator echoes25,27,212–217 that interrogate an interface such as the diaphragm-lung or heart-lung.25 Positional changes of this interface along the axis of the navigator can then be followed through the normal respiratory cycle. Image acquisition then can be gated to respiratory motion: data are acquired (prospective gating) or accepted (retrospective gating) only when the interface is within a preselected window near the end-expiratory position. A more elaborate variation is to prospectively correct the slice position on the basis of navigator data, thereby allowing for the use of wider gating windows with improved time efficiency and image quality.27,218 An alternative method for minimization of respiratory motion artifacts would be to decrease the acquisition time so that an entire coronary MRA data set could be obtained in several brief breath-holds.27,178,212,218 Enhanced gradient functions and use of MR contrast agents (see below) will potentially allow for such rapid imaging.

Intravenous contrast media have had an important impact on aortic, renal, and peripheral MRA11 and promise to provide similar enhancements for coronary MRA. Recently, the use of contrast media has been described for coronary MRA in experimental animals for assessment of bypass graft patency208–210,219 and for visualization of the native coronary arteries in humans.213

Coronary MRA is a noninvasive diagnostic method currently in a state of rapid evolution. Advances in technology will undoubtedly lead to enhanced resolution, improved accuracy, and shorter scan times. It is almost certain that coronary MRA will be a prominent diagnostic clinical tool in the future. In anticipation of this, the development and use of more MR “friendly” intracoronary stents, graft markers, and vascular clips are strongly encouraged.

Pulmonary MRA has also been slow to develop into a clinically useful application. Initial approaches used 2D or 3D TOF techniques, the former with and the latter without breath-holding.22,220–223 Some encouraging results were achieved, but the techniques were not reliable enough to allow widespread clinical use. Recently, the use of CEMRA has been extended to the pulmonary arteries.224–229 It is possible to cover the entire pulmonary tree in a single breath-hold during the injection of contrast. Meaney et al224 reported on 30 patients who underwent both breath-hold pulmonary CEMRA and CA for suspected pulmonary embolism. A spatial resolution of 1.25×2.5×3.0 mm was achieved in that study. For the 3 readers, sensitivity ranged from 75% to 100%, with specificities of 95% to 100%. As scan speeds improve, resolution promises to further increase. Steiner et al225 reported spatial resolution of 1.7×1.7×2.0 mm in a 23-second breath-hold. Recently, the blood pool contrast agent NC100150 has been used to achieve free-breathing pulmonary MRA with navigator echoes in normal subjects.226 With continuing improvements in technology and validation in larger groups of diseased patients, pulmonary MRA may become competitive with the other minimally invasive imaging modalities for pulmonary embolism, including radionuclide ventilation-perfusion scanning and CT angiography.

Conclusions
MRA continues to show rapid evolution. The hallmark of the last few years has been the widespread interest in CEMRA,
which has essentially displaced many of the flow-dependent MR strategies used previously. Current CEMRA uses extracellular-fluid gadolinium contrast agents in first-pass mode. Progress in several areas will determine the evolution of MRA in the near future. Several intravascular contrast agents are in clinical trials, and the role of these agents appears promising but remains to be determined. Hardware and software advances continue, with development of ever-faster scanning, more powerful image processing, and dedicated devices such as cardiac and peripheral coils and movable tables for peripheral MRA.

The current status of MRA can be summarized as follows (Table 8): carotid MRA has established itself as a complement to DUS for preoperative evaluation, markedly lessening the need for CA. MRA has gained widespread acceptance for imaging the aorta. The use of aortography can now be limited to special situations. In the renal arteries, MRA has an established role for main renal artery screening. Additional advances will be required to replace CA for evaluation of intrarenal vessels and nuclear medicine for renal functional evaluation. In peripheral vascular disease, MRA can replace CA now in patients in whom catheter introduction or iodinated contrast poses special risks (e.g., because of contrast allergy, renal failure, or severe inflow disease). With the introduction of CEMRA and the development of movable tables for peripheral MRA.

References


100. Deleted in proof.


103. Deleted in proof.


106. Deleted in proof.


109. Deleted in proof.

110. Deleted in proof.


115. Deleted in proof.

116. Deleted in proof.


**Key Words**: AHA Scientific Statement ■ angiology ■ magnetic resonance imaging ■ imaging ■ arteries
Magnetic Resonance Angiography: Update on Applications for Extracranial Arteries
E. Kent Yucel, Charles M. Anderson, Robert R. Edelman, Thomas M. Grist, Richard A. Baum,
Warren J. Manning, Antonio Culebras and William Pearce

Circulation. 1999;100:2284-2301
doi: 10.1161/01.CIR.100.22.2284

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
Copyright © 1999 American Heart Association, Inc. All rights reserved.
Print ISSN: 0009-7322. Online ISSN: 1524-4539

The online version of this article, along with updated information and services, is located on the
World Wide Web at:
http://circ.ahajournals.org/content/100/22/2284

Permissions: Requests for permissions to reproduce figures, tables, or portions of articles originally published
in Circulation can be obtained via RightsLink, a service of the Copyright Clearance Center, not the Editorial
Office. Once the online version of the published article for which permission is being requested is located,
click Request Permissions in the middle column of the Web page under Services. Further information about
this process is available in the Permissions and Rights Question and Answer document.

Reprints: Information about reprints can be found online at:
http://www.lww.com/reprints

Subscriptions: Information about subscribing to Circulation is online at:
http://circ.ahajournals.org//subscriptions/