Basics of Magnetic Resonance Imaging

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In clinical diagnostic imaging there has been a rapid evolution of nuclear magnetic resonance techniques. A subset of these has proven particularly powerful when applied to the arena of noninvasive vascular imaging. This article will serve as a primer for practitioners with little prior background in magnetic resonance angiography (MRA). It will develop the first principles of magnetic resonance image creation, and establish familiarity with current imaging applications of the thoracic, abdominal, and peripheral vasculature.

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THE EVOLUTION OF nuclear magnetic resonance techniques to clinical utility in diagnostic imaging took place in the early 1970s. The landmark work of chemist Paul Lauterbur and physicist Sir Peter Mansfield began a rapid progression of continuously improving imaging techniques, which now enjoy widespread application in contemporary clinical practice. These applications are particularly powerful when applied to the arena of noninvasive vascular imaging. This article will serve as a primer for practitioners with scant familiarity with this area and will seek to provide a discussion of the fundamental concepts of noninvasive vascular magnetic resonance (MR) imaging. More detailed readings are suggested in the references.1,2

BACKGROUND

Nuclear magnetic resonance signals are created by certain atomic nuclei when they are excited by radiofrequency (RF) energy in the presence of a strong magnetic field. In 1946, Edward Purcell and Felix Bloch independently developed methods for determining nuclear magnetic precision measurements.3 For this work, they were awarded the Nobel Prize in Physics in 1952. Derivative techniques harnessing these properties to generate spectroscopic data have since been widely employed to elucidate issues of chemical composition in solids and liquids.4 Spurred by the development of computed tomography (CT) in the early 1970s and coupled with advances in the development of new reconstruction algorithms, Drs. Lauterbur and Mansfield hypothesized and created different methods to translate information regarding magnetic spin into cross-sectional images. Many further refinements at various stages have resulted in the acquisition of higher resolution depictions through the improved extraction of signal from tissue over background noise and improvements in the generation of tissue contrast. Additional innovations have permitted the use of intravascular contrast agents to enhance the innate ability of MR imaging to visualize flowing blood in a variety of circumstances.

FIRST PRINCIPLES

Conceptualization of the process by which a useful clinical image is generated necessitates a discussion of three basic areas:
1. Signal creation from the properties of MR
2. Image formation from this signal
3. Generation of tissue contrast

Signal Creation

Some atoms, such as hydrogen, carbon 13, fluorine, sodium, and phosphorous, are inherently susceptible to a magnetic field. That is, these atoms behave like bar magnets, or dipoles, and tend to align themselves along a dominant magnetic field. In doing so, they forgo their otherwise random spatial orientations and become somewhat aligned. The powerful superconducting electromagnet built into the bore of a contemporary MR scanner—approximately 15,000 times the strength of the earth’s magnetic field—provides such a dominant external magnetic field. While all of the atoms listed above can behave in this fashion, clinical MR is essentially the imaging of hydrogen atoms in free water as well as the imaging of hydrogen associated with macromolecules (proteins, lipids, etc.).

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0895-7967/04/1702-0002$30.00/0
doi:10.1053/j.semvascsurg.2004.03.011

These hydrogen atoms—which I will now refer to as protons—do not align themselves perfectly with the dominant external magnetic field and instead rotate in a cone-shaped fashion termed **precession**. The frequency of precession depends upon the strength of the magnetic field as modified by a unique gyromagnetic constant for a given type of nucleus. The precession of protons in this fashion generates a second, smaller magnetic field or magnetization. In its equilibrium state, this tissue magnetization is much weaker than the dominant external magnetic field.

Tissue magnetization does, however, represent a source of potential energy and, therefore, potential signal for image generation when the magnetization of the tissue of interest is intentionally perturbed. Energy can be applied to alter the tissue magnetization in the form of an RF pulse adjusted to the same frequency of precession. The added energy results in augmentation of the angle of precession of the collected protons exposed to the RF pulse. This converts some of the magnetization existing along the axis of the dominant magnetic field (longitudinal magnetization) into measurable magnetization along a perpendicular axis (transverse magnetization). After an RF pulse, the longitudinal magnetization recovers back to its original pre-perturbation state. Induced transverse magnetization recovers at a different rate back to 0, its pre-RF pulse value. At the time of imaging, it is the yet unrecovered transverse magnetization from an incident RF pulse, which induces an electrical, or radio signal that can be perceived by antenna coils within the MR scanner.

The recoveries of longitudinal and transverse magnetization are also termed **relaxation**. Longitudinal relaxation is referred to as spin-lattice relaxation or T1. Simply put, the energy initially transferred to the precessing protons by the applied RF pulse is transferred back to the protons’ surroundings or lattice. T1 actually refers to the time it takes for a tissue being interrogated to recover 63% of its original longitudinal magnetization after an applied RF pulse.

Transverse relaxation, also referred to as spin-spin relaxation or T2, is the time required for 63% of the RF pulse generated transverse magnetization to dissipate. This phenomenon is essentially caused by the loss of the phase coherence, or order amongst the precessing protons in the transverse plane.

In biological tissues, T1 values range from 300 to 1200 milliseconds and T2 values range from 30 to 150 milliseconds. These values are not exactly characteristic for a given tissue type, however, some generalizations can be made. Water has both a long T1 and T2; compared with water, fat has a relatively short T1 and T2. Pathologic tissues often demonstrate higher water content than surrounding normal tissues and therefore imaging sequences can be chosen to emphasize the longer T1 and T2 times one would expect to see. T1 depends on tissue structure. The T1 of water is longer than that of fat because it is more difficult for the relaxing water proton to transfer its energy back to other rapidly moving free water molecules in the lattice or tissue. The T2 of water is longer than that of fat, because rapidly moving water molecules do not remain in one place long enough to allow a local inhomogeneity to develop. An inhomogeneity, or lack of magnetic uniformity within an area, will lead to the loss of phase coherence (order among precessing protons), which will result in the loss of signal.

**Image Formation**

As stated previously, transverse magnetization resulting from perturbations induced by an applied RF pulse generates a signal that is captured by an antenna system housed within the magnet. By applying specially designed sequences of RF pulses, the MR operator can emphasize certain tissue characteristics to be displayed in the resulting image. A simple pulse sequence may only vary the time to pulse repetition (TR) of the successively applied RF pulses to accentuate differences in magnetization and signal intensity of selected tissues.

Similarly, a pulse sequence might seek to emphasize the difference in T2s of body tissues by varying the amount of time allowed after excitation (TE) by an RF pulse, and the collection of a signal. A formal discussion of selectable imaging parameters far exceeds the scope of this introduction. It should, however, be sufficient to say that there exist many ways to exploit the differences between tissues and to therefore generate contrasts useful for imaging.

Interestingly, the signal generated by the relaxation from the applied RF pulse sequence does not contain much information concerning where in the volume of excited tissue it came from. Essentially,
signal is generated from a bulk of interrogated tissue, but exactly localizing the signal to a particular portion of that tissue in order to construct an image is similar to trying to localize where in a group of choral singers various sounds or voices are originating. In fact, unlike CT, where spatial information can be generated by mathematical “back projection” algorithms, MR relies on a much more complicated method to determine the locations of the sources of MR signals.

This system requires the application of supplemental gradient magnetic fields to the MR scanner’s dominant field to create ordered distortions of the magnetic field experienced by the tissue to be studied. A graded distortion in the magnetic field along a particular axis results in a predictable change in the frequency of precession of the protons arranged along the axis of the applied supplemental gradient (recall that the frequency of a proton’s precession is directly proportional to the magnetic field strength it experiences as related by the gyromagnetic constant). To image a particular slice of tissue, RF pulses must be chosen at the specific frequency able to excite only the slice selected by the supplemental gradient added along the desired axis. This is a slice selection gradient.

Similarly, once a desired slice of tissue is selected, another supplemental gradient may be applied in the transverse (x-axis) plane after the RF pulse has been sent in. This supplemental gradient is constructed to vary across the x-axis, from left to right, in a predictable fashion. The gradient will, therefore, affect the frequency of precession of the protons arranged along this axis in a predictable fashion and thereby result in the emission of signals of differing frequencies, which differ from one another in a predictable order. That is, the position of a given proton among a group of protons arrayed along the x-axis can be determined based upon its frequency of precession relative to the others along the applied frequency encoding gradient. Such determinations are achievable through the use of a mathematical process known as Fourier transform analysis, which converts frequency data in a complex wave (the emitted collective radio signal reaching the antenna) into a map of signal strengths for each frequency encountered.

Until this point we have only accounted for two dimensions in an otherwise three-dimensional (3D) patient. In order to differentiate two protons at the same location along the x-axis but located at different positions along the y-axis from one another, still another trick is required. A third supplemental magnetic gradient called the phase encoding gradient is briefly turned on and off along the y-axis. Instead of merely altering the frequencies of the protons that are exposed to it and taking advantage of that information in the altered spin signals as we did along the x-axis with the frequency encoding gradient, we now do something else: the short exposure to the magnetic gradient results in some protons precessing more quickly relative to the others based upon their location, relative to the gradient—stronger magnetic fields induce faster spins—so that when the gradient is turned off, the protons are now out phase with one another. Because the gradient has been turned off, the protons along the y-axis all have the same frequency, but now display differing phases based on their locations along the y-axis.

While this all may seem quite a hodge-podge of spin frequencies and phases for various protons scattered throughout the region we are imaging, this complex wave information can be mathematically interpreted. The computer of the MR scanner takes the mixture of different signals that emerge and sorts through the different frequencies of the various protons and the different phases for those protons with the same frequency by means of the Fourier transform. What emerges is an analysis of how much signal (amplitude) of a specific frequency and phase is present in the portion of the patient imaged. The aforementioned tricks—frequency encoding and phase encoding—assign the spatial information; the Fourier transform generated amplitudes are displayed as shades of gray on the final clinical image.

**Tissue Contrast**

Contrast is the exhibition of unlikeness between comparable qualities. In MR imaging, at the most basic level, the presence or absence of protons available to provide imageable signal generates contrast. On an MR of the chest, air within the lungs is proton poor and therefore provides contrast to the protons present within the fat of the ribs’ bone marrow or heart muscle (mainly water). Proton density is therefore a basic form of MR imaging contrast.

There are several other ways to discern differences between tissues. As we have already discussed, imaging sequences of RF pulses can be
constructed to emphasize the $T_1$ properties of various protons: recall that a short $TR$ between RF pulses will result in there being more observable signal in a tissue with a shorter $T_1$ compared with a tissue possessing a longer $T_1$. If we had used a much longer $TR$, there might not be any discernable difference between the amounts of signal that could be elicited from these two tissues. The use of paramagnetic substances—substances that possess unpaired electrons and thereby can concentrate local magnetic forces—can influence (increase) the rate of both the $T_1$ and $T_2$ relaxations of water protons in their proximity. Selective administration of gadolinium (one such agent) to one tissue and not another can be used to increase the difference in $T_1$’s between the two tissues, thereby increasing observed contrast.

Blood flow has interesting effects on MR imaging. What has been described above is predicated upon a static imaging target. Flowing protons, as in blood, do not stay in the slice of patient being imaged long enough to contribute their spins to the RF pulse-elicited signal that generates the final image. Blood vessels contained within a slice of imaged tissue will therefore demonstrate an intraluminal void of signal, termed a *flow void*. It is also possible to use flow to generate signal enhancement. To accomplish this, the slice of tissue to be imaged can be saturated—depleted of its longitudinal magnetization by repeated RF pulsing—thereby allowing the fresh spins of newly entering blood protons to bring a source of new longitudinal magnetization for subsequent imaging. This is termed *flow-related enhancement* and forms the basis for many MR vascular imaging studies performed with a technique known as *time of flight*. So as not to visualize both arterial and venous signals at the same time, the technique is augmented by the placement of saturation bands, which allow the elimination of signal, generated by blood flowing in one direction or the other.

Perhaps the most significant enhancement to contemporary vascular imaging with MR was provided by the use of intravenous gadolinium to provide an additional intravascular signal. As gadolinium induces $T_1$ shortening, it was reasoned that a sufficient concentration of paramagnetic agent in the blood could accelerate the recovery of longitudinal magnetization, that the observed contrast between the signal of even flowing blood and the saturated background tissues would become prominent. Although, initially limited by technical factors inherent to the scanners themselves, contemporary equipment provides stronger and faster gradients, which permit shorter acquisition times. This is important, as shorter acquisition times reduce both the amount of intravenous contrast required and the possibility of motion-induced artifacts.

**APPLICATIONS OF MR ANGIOGRAPHY**

In many centers, MR angiography (MRA) has emerged as the imaging modality of choice in the evaluation of the thoracic and abdominal aorta, and their major branch vessels, including the carotid, renal, and mesenteric arteries. Its acceptance as an accurate tool in the evaluation of peripheral vascular disease continues to grow exponentially.

Its advantages over traditional, catheter-based arteriography include the avoidance of nephrotoxic contrast media and ionizing radiation, the relative lack of invasiveness, the ability to image the vessel wall and surrounding structures, the multiplanar and 3D image display, and lower cost. MRA imaging provides essential information for preoperative planning of surgical and endovascular procedures. Following noninvasive MRA, patients can be scheduled directly for intervention. By eliminating the catheter-dependent diagnostic portion, which would otherwise precede intervention, such imaging provides more efficient use of operating room and angiography suite resources. Use of iodinated, potentially nephrotoxic contrast material may also be reduced.

The primary imaging protocol utilized in contemporary MRA is 3D contrast-enhanced MRA (3D CE MRA) with occasional use of two-dimensional (2D) time-of-flight (2D TOF). 3D volumetric imaging is preferred because it provides “angiogram”-like images, can image a large volume in a single acquisition and is much less susceptible to artifacts. Its main limitation is the absolute crucial element of timing for the intravenous infusion of gadolinium contrast. The data obtained from a single acquisition can be manipulated to demonstrate an image in any desired projection (Fig 1), as opposed to conventional angiography, which requires a separate acquisition, or run, for each projection.
In order to expedite patient imaging and optimize patient care, CT is the technique of choice in the evaluation of acute conditions such as trauma, acute aortic dissection and thoracic aortic aneurysm rupture. 3D CE MRA has emerged as the imaging study of choice in the evaluation of most other conditions. The acquisition is typically performed in the oblique sagittal plane to allow for the fewest number of sections and to parallel the course of the aortic arch. Breath-holding and electrocardiogram (ECG) gating are essential for optimal imaging of the ascending aorta to minimize motion-related artifacts. The descending aorta can be adequately imaged without a breath-hold. In addition to 3D CE MRA, cine imaging should be considered to assess flow dynamics, particularly in the evaluation of aortic dissection. So-called “black blood” spin echo, T₂-weighted, and delayed postcontrast T₁ and/or gradient echo imaging sequences should be employed to evaluate the aortic wall and perivascular structures.

THORACIC AORTA

In comparison to conventional angiography, Prince et al demonstrated 100% accuracy of 3D CE MRA in the morphologic evaluation of aortic dissection and aortic aneurysm. Complete assessment of aortic dissection includes depiction of:

1. Entry point—to differentiate Stanford type A (ascending aorta involvement), which requires immediate surgical intervention, from type B (descending aorta only) dissections.
2. Distal extent and re-entry points of the dissection.
3. True lumen compromise.
4. Branch vessel involvement, including evaluation of lumen of origin, and presence of compromise (Fig 2).
5. Presence of end-organ ischemia.
6. Overall diameter of the aorta.
7. Follow-up imaging to monitor for extension or enlargement of the dissection over time.

MR imaging will also differentiate classical aortic dissection from intramural hematoma and atherosclerotic penetrating ulcer.

Thoracic Aortic Aneurysm

The presence of aortic aneurysm and its relationship to branch vessels is readily depicted on 3D CE MRA, particularly with the use of multiplanar reformatted imaging. Based upon these images, the aneurysm can be classified by its location, shape, or etiology. However, MRA does not reveal any information pertaining to the aortic wall or the presence of mural thrombus. Therefore, it is essential to acquire cross-sectional, in particular, post-contrast images to obtain accurate aortic diameter measurements. These measurements are essential
Fig 2. 3D CE MRA of an abdominal aortic dissection. Oblique coronal (A) and sagittal (B) maximum intensity projection images demonstrate a complex aortic dissection. The false lumen is partially thrombosed and contains webs. The oblique sagittal view (B) demonstrates the origins of the celiac and superior mesenteric arteries without any evidence of compromise. Although the renal arteries appear patent on these views, it is necessary to perform more complex reconstruction to optimally visualize their origins. Oblique axial (C and E) and coronal (D and F) multiplanar reconstruction images of the right (C and D) and left (E and F) renal arteries demonstrate the absence of compromise. The same information would have required multiple conventional angiographic acquisitions using a significant volume of iodinated contrast.
for determining the need for surgical intervention and in preoperative planning for endovascular repair. The presence of enhancement of the aortic wall and surrounding soft tissues suggests the presence of a mycotic aneurysm or aortitis.

Arteritis

3D CE MRA is excellent at demonstrating luminal irregularities, beading, stenoses, and occlusions of branch vessels due to arteritis. However, T₂-weighted, fat saturated and delayed postcontrast T₁-weighted images are essential in evaluating the aortic wall for inflammation. The presence of bright signal in the aortic wall on these two imaging sequences, suggests the presence of active aortitis.¹⁰

Postoperative Evaluation

MRA is very useful in the postoperative evaluation of aortic grafts for the presence of anastomotic aneurysm and stenosis. Postcontrast imaging with ECG gating (to resolve cardiac motion at the ascending aorta) is essential in evaluating for graft infection.

Congenital Anomalies

Anatomic variants, including arch anomalies and aortic coarctation, are readily assessed with the combined use of conventional MR imaging and 3D CE MRA with multiplanar reformatting. Cine phase contrast imaging can provide additional functional information.

ABDOMINAL AORTA

Imaging is typically performed in the coronal plane to display the abdominal aorta with the fewest number of sections. The field of view should be set to include the aorta above the celiac trunk origin proximally, and at least the common iliac artery bifurcation distally.

Abdominal Aortic Aneurysm

3D CE MRA combined with T₁-weighted delayed postcontrast imaging, and axial spin or gradient echo imaging has been shown to provide an excellent depiction of overall aneurysm morphology, maximal diameter measurements, proximal and distal extent, and aneurysm relationship to the renal, mesenteric, and iliac arteries (Fig 3). It also demonstrates the presence of concomitant occlusive disease involving the renal, mesenteric, and iliac arteries (Fig 3). Kelekis et al¹² demonstrated 100% accuracy of 3D CE MRA in the evaluation of abdominal aortic aneurysms (AAA).

The deficiency of MRA in preoperative staging of AAA is its inability to demonstrate the presence of vessel wall calcifications and its lower spatial resolution for sizing of the aortic wall for stent-graft placement. Additionally, stent-induced signal voids and metallic medical device safety concerns have significantly limited its use in postoperative stent graft follow-up and surveillance. For these reasons, at this time, CT angiography remains the imaging modality of choice for both pre- and postoperative evaluation of AAA for endovascular stent grafting. Nonmagnetic, nitinol-based stents may be imaged adequately and safely, however, further studies will be necessary before this can be accepted into practice.

Abdominal Aortic Dissection

Evaluation of abdominal aortic dissection is similar to that described above in the section on thoracic aorta (Fig 2).

Aortic Stenosis and Occlusion

In comparison to conventional angiography, 3D CE MRA, has demonstrated sensitivities of 99% to 100% and specificities of 89% to 100% in detecting stenotic and occlusive disease of the abdominal aorta and iliac arteries (Fig 4). It is superior to conventional angiography in the evaluation of complete aortic occlusion. In this situation, conventional angiography requires a brachial or axillary approach and relies upon the delivery of contrast to the periphery via multiple small collateral vessels, and a high volume of contrast, whereas, MRA requires a single, appropriately timed, injection into a peripheral vein (Fig 5).

MRA is limited poststenting because of stent-induced paramagnetic susceptibility artifacts. However, as mentioned above, this may be partially resolved with the use of nitinol or platinum-based stents. Imaging with wider flip-angles and very short TEs, typically used with 3D MRA, may also be helpful.

RENALE ARTERIES

Higher resolution imaging techniques such as the use of larger imaging matrices (512 matrix/ frequency encoding) and higher performance gra-
dient systems (to permit shorter imaging times and, therefore, breath-hold technique) are necessary for diagnostic imaging of the renal arteries. The imaging plane should be in the coronal or oblique coronal plane to minimize the number of sections and acquire images in parallel with the course of the renal arteries. A well-executed study should include evaluation of the major branches of the renal artery. Renal MRA should include delayed postcontrast 3D and/or 2D gradient recalled echo imaging sequences to include evaluation of the renal parenchyma and adrenal glands for mass, infarcts, and other pathology.

Renal Artery Stenosis

Imaging should include the abdominal aorta from above the celiac artery to below the common iliac artery bifurcation (Fig 6). These vessels should be imaged because they may be useful for surgical revascularization, additionally, accessory renal arteries can arise off the common iliac artery. On MRA, as opposed to CT angiography, calcium

Fig 3. 3D CE MRA of a thoracoabdominal aortic aneurysm. Coronal (A) and sagittal (B) maximum intensity projection images demonstrate a complex thoracoabdominal aortic aneurysm with significant luminal irregularity, celiac and superior mesenteric artery stenosis (B) and bilateral common iliac artery irregularity (A). Additional small aneurysms are seen involving the right common and left internal iliac arteries (A). The renal arteries are not visualized (A) in this patient with hemodialysis-dependent end-stage renal disease. The actual diameter of the aorta cannot be ascertained on the MRA images, because these only depict the lumen of the vessel, the so-called "luminogram." Axial cross-sectional image obtained at the same time at the level of the kidneys (C) demonstrates the true cross-sectional diameter of the aneurysm, which is significantly larger than that depicted on the MRA. This is due to the presence of significant mural thrombus. Also note the small size of the kidneys, particularly on the left side (C). Incidentally, note excellent visualization of the portal vein and its branches (A).
Renal function can also be evaluated in several ways: by measuring arterial flow with gated, breath-hold cine phase contrast techniques; by comparing the right and left nephrograms on dynamic perfusion imaging (Fig 7); by evaluating for poststenotic induced signal void on 3D phase contrast sequences; and by assessing excretion of gadolinium with a delayed 3D gradient echo sequence. The diagnosis of renal artery stenosis can be further supported by diminished renal length and parenchymal thickness on the affected side. One should also look for other possible causes of hypertension including adrenal tumors, juxtaglomerular cell tumors, and page kidney.

In one of the larger studies comparing 3D CE MRA to conventional angiography, the sensitivity of 3D CE MRA has been shown to be 96% with a specificity of 92%. Several other studies have demonstrated sensitivities over 95% and sensitivities over 85%.

Most renal artery stenoses are secondary to atherosclerotic disease extending into the vessel from a diseased abdominal aorta (Fig 7). Fibromuscular
dysplasia is another disease of the vessel wall that typically affects younger patients and females. The most typical appearance is that of webs alternating with small aneurysms, the so-called “string of beads” sign that develops in the mid-to-distal renal artery. This may also be seen on a well-executed study.

Renal Transplant Donors

In evaluating potential renal donors, it is essential to identify the number, location and length of the renal arteries, presence of anatomic variations and to confirm the presence of a contralateral normal kidney. This can all be accomplished with a well-executed 3D CE MRA.

Renal Transplants

Imaging is similar to that for native renal arteries with the field of view centered on the pelvis and 2.5 mm or less section thickness. A precontrast axial T$_2$-weighted sequence with fat saturation should be done first to assess for perinephric fluid collections and hydronephrosis. Delayed post-con-
which is typically due to proximal stenosis, the sagittal plane is preferred. MRA is limited by suboptimal imaging of small peripheral branches and collateral pathways. Acute mesenteric ischemia is a surgical emergency that should not be delayed with MRA imaging, except under very unusual circumstances.

**Mesenteric Ischemia**

Chronic mesenteric ischemia typically occurs when there is significant stenosis or occlusion involving at least two of the three main splanchnic arteries (Fig 8). Carlos et al.\(^\text{16}\) demonstrated a sensitivity of 96% and a specificity of 95% in the diagnosis of mesenteric ischemia with 3D CE MRA.

Acute ischemia may be due to in situ thrombosis due to ruptured plaque or may be secondary to embolic occlusion. Other etiologies include poor cardiac output and venous thrombosis (Fig 9). Complete evaluation requires portal venous imaging. Evaluation of the inferior mesenteric artery requires careful review and manipulation of reformatted images. It should be readily visualized using current fast scanner technology.

Other potential uses of mesenteric artery MRA includes evaluation of surgical bypass grafts, visceral artery aneurysms, tumor encasement, anatomic variations of importance in preoperative planning, and pre- and post-liver transplantation vascular assessments.

**PERIPHERAL ARTERIES**

When MRA of the lower extremities was first established, it was performed using 2D TOF imaging without administration of contrast. This method takes advantage of the fresh signal of flowing blood, which had been selectively excited. Imaging is performed in the axial plane. As a result, both arterial and venous blood flowing into the slice to be imaged will appear bright. The differentiation of arterial signal from venous signal was resolved by the application of a saturation pulse on the side of venous flow. The most significant problem with 2D TOF imaging is the relatively long acquisition time required—scanning times could extend as long as 2 hours! The development of 3D CE MRA has nearly eliminated the need for 2D TOF with some exceptions.

The greatest challenge for 3D CE MRA imaging today, is the coordination between the timing of contrast injection with imaging of the lower extremities at several, typically three, separate anatomic stations. Two dominant methods have
emerged. One is known as floating table or bolus chase 3D CE MRA. This involves a series of 3D acquisitions beginning at the level of the aorto-iliac arteries with actual chasing of the contrast down the legs at two additional stations, one at the thighs, and the next to include the calves and feet (Fig 10). There are two variations of the bolus chase, one utilizing a slow continuous infusion of contrast, and the other involving a faster bolus injection with more rapid imaging and faster table movement between stations.

The second method, known as time-resolved peripheral MRA, involves a series of three separate, escalating low-dose contrast injections at each station, using image subtraction to eliminate the signal accumulated from the prior injection. Both of these methods can be combined with 2D TOF imaging of the foot and ankle, and possibly also the calf, prior to contrast injection and imaging of the pelvis and thighs. This may be used to compliment the overall study or be helpful as a “backup” in case the timing of the 3D CE MRA portion of the study is poor.

There are several other methods currently under investigation, including one that utilizes a continuous table motion analogous to spiral CT examinations.

Atherosclerotic Occlusive Disease

In many centers, 3D CE MRA has virtually replaced conventional angiography in the evaluation of patients with peripheral vascular occlusive disease (Fig 10). A recent meta-analysis demonstrated sensitivities of peripheral MRA ranging from 92% to 100% and specificities ranging from 91% to 100%.

Bypass Graft Surveillance

3D CE MRA has been increasingly used in bypass graft surveillance (Fig 11). The presence of
Fig 10. 3D CE MRA of the abdomen and both lower extremities using the bolus chase stepping table technique, and a lower extremity phased array coil. Coronal maximum intensity projection (MIP) at the first station—the pelvis (A) demonstrates widely patent infrarenal abdominal aorta, bilateral common, external and internal iliac arteries, and bilateral common femoral arteries. Bilateral profunda femoral arteries and proximal superior femoral arteries are included as well, and are widely patent. There currently is an extension coil that can be placed over the upper abdomen to include imaging of the renal arteries. Coronal MIP at station two—the thighs (B) demonstrates a focal severe stenosis at the adductor canal on the left side (arrow). There is signal loss proximally (also seen at station one) attributable to “edge-of-coil” artifact. This region though is adequately imaged at the first station. This protocol typically calls for overlap between stations. Coronal MIP at station three—the calves (C) demonstrates widely patent three-vessel runoff to the feet, with patency of the posterior tibial artery across the ankle and into the plantar artery. Visualization of some calf veins is present—what we call venous contamination. Despite this, the arteries are well seen, making this a diagnostic exam. Venous contamination can usually be avoided with appropriate timing of the contrast injection in relation to imaging. When it does occur, the arteries can be distinguished from the veins by their appearance, location and density relative to the arteries. When there is cellulitis or tissue loss due to ischemia, precontrast 2D TOF imaging of the feet and possibly the ankles should be considered as a backup, due to early venous drainage related to inflammation.
surgical clips has not been a significant issue with the very short TEs currently used. The ability to acquire a 3D data set, which can be viewed in any projection (Fig 5), is distinct advantage over conventional angiography, which typically requires multiple acquisitions in several projections to appropriately demonstrate anastomotic sites in profile. Bertschinger et al demonstrated 100% accuracy of 3D CE MRA in the surveillance of peripheral artery bypass grafts in comparison to conventional angiography.

Nonatherosclerotic Peripheral Vascular Occlusive Disease

When the distribution of the occlusive process in the lower extremities is clearly asymmetric, particularly if one side is normal or nearly normal and without the typical risk factors for atherosclerotic occlusive disease, other possible etiologies of occlusion should be considered, including distal emboli, popliteal artery entrapment, popliteal artery aneurysm, and cystic adventitial disease. Several of these have distinct MRA findings. Popliteal artery aneurysms should be readily identified on the MRA. Imaging a patient suspected of popliteal artery entrapment should include one acquisition at the knee during plantar-flexion, which should elicit an oblique band-like stenosis in the mid popliteal artery. The very unusual cyst in the vessel wall in the very unusual condition of cystic adventitial disease may be identified on MRA.

Arteriovenous Malformation

Arteriovenous malformations (AVMs) should ideally be imaged with an extremity coil using multiphase technique to demonstrate the feeding arteries, vascular blush, and early draining veins, as well as to distinguish slow from fast flow AVMs (Fig 12).

MR Venography

The venous system is readily imaged using traditional TOF or phase contrast techniques, however, the presence of extremely slow blood flow, very tortuous veins and/or small, tortuous collaterals significantly limits these methods. Additionally, in imaging the upper extremity and central veins, several acquisitions may be necessary because the subclavian veins course perpendicular to the extremities, and the direction of flow in the superior vena cava (SVC) is opposite that of the arms (with the patient’s arms at the side). Even without these issues, TOF MR venography (MRV)
can be very time consuming. 3D contrast-enhanced MRV remediates these issues.

There are essentially two methods of performing 3D contrast-enhanced MRV, direct and indirect. In the direct method, a very dilute mixture (~1:10 to 1:20) of gadolinium is injected via an angiocath(s) placed into the extremity/ies to be imaged using a 3D spoiled gradient echo sequence. Imaging is

Fig 13. 3D CE MR venogram of the central veins. This is a series of coronal source images obtained during the simultaneous bilateral injection of a dilute, 10:1, mixture of saline and gadolinium via angiocaths placed in the antecubital veins. The images progress from anterior to posterior in this series. This patient, with a functioning left arm arteriovenous (AV) fistula for hemodialysis, developed severe swelling of his left arm and the left side of his body. He is a recent recipient of a renal transplant, has an elevated creatinine, and is not a candidate for conventional venography and iodinated contrast. The study demonstrates focal severe stenosis or occlusion of the left innominate vein (arrow). The aortic arch is seen because of rapid venous flow from the AV fistula, and suboptimal coordination between the beginning of the injection and imaging. Nonetheless, this is a diagnostic study, confirmed during treatment with venous angioplasty and stenting the following day, using gadolinium as the contrast agent. The swelling resolved 1-day posttreatment.

Fig 14. 2D TOF and 3D CE MR venogram of the central veins. Coronal maximum intensity projection (MIP) 2D TOF with an inferiorly placed saturation band (A) demonstrates focal occlusion of the upper SVC at the junction of the innominate veins (arrow). Diffuse transverse dark bands represent artifact generated from stacking multiple axial images to create the 3D reconstructed MIP image. This was followed by 3D CE imaging. Coronal MIP (B) and subvolume MIP (C) confirms the findings seen on the TOF sequence, but with fewer artifacts. Note that the TOF image provides better imaging of the internal jugular veins, but no information on the subclavian veins because of “in-plane” flow artifact, an inherent limitation of TOF imaging.
performed during a prolonged injection following a short delay. The indirect method involves delayed, early equilibrium phase coronal MRV during the relatively slow injection of a high volume of gadolinium via a peripheral vein. For both of these methods, 2D TOF or phase contrast imaging can be performed as a complimentary or backup technique.

**Upper Extremity Veins and SVC**

For direct MRV of the arm, precontrast images should be obtained first for subtraction, imaging should be delayed until approximately 30 to 50 cc have been injected and the injection should be continued throughout the imaging to keep the veins distended. In imaging of the central veins, including the SVC, simultaneous bilateral injections should be performed via antecubital IVs (Fig 13). Evaluation of dialysis fistulas can be performed either using the indirect method to image inflow and outflow, or by direct IV access into the fistula itself. Shinde et al\(^{20}\) and Thornton et al\(^{21}\) demonstrated 100% accuracy using 3D CE MRV in the evaluation of the central veins using the indirect method of imaging (Fig 14).

**Lower Extremity Veins and Inferior Vena Cava**

The lower extremity veins and inferior vena cava (IVC) are ideally suited for 2D TOF imaging because, for the most part, their entire course is perpendicular to the axial imaging plane (Fig 15). However, 3D contrast-enhanced MRV provides better resolution. Tourniquets may be used to optimize filling of the deep veins. Simultaneous imaging of both legs can be performed during a single, prolonged, large-volume injection by manually sliding the patient and imaging at three stations. Ruehm et al\(^{22}\) demonstrated sensitivities of 94% to 100% and specificities of 92% to 98% in the evaluation of various disease entities involving lower extremity veins utilizing the direct method of imaging.

**SUMMARY**

Over the two decades from its clinical inception, MR imaging has found widespread application in nearly every area of medical practice. It has transformed medical diagnosis in many ways, with perhaps none more significant than its ability to replace prior “gold standard” invasive diagnostic procedures with safer and highly accurate alternative noninvasive imaging. With contemporary MR scanner technology, the application of MR imaging to the arena of vascular imaging is now enjoying widespread acceptance.

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