

99. MRA Principles and Carotid MRA

As described in Chapter 12, time of flight (TOF) magnetic resonance angiography (MRA) is commonly utilized in the evaluation of the circle of Willis. TOF MRA allows depiction of vasculature without the utilization of gadolinium-chelate contrast agents, relying instead on flow-related enhancement. The appearance of flowing blood is dependent upon the technique employed. 2D spin echo and fast spin echo sequences employ slice selective 90 and 180 radiofrequency pulses, requiring a given proton to be subjected to both pulses in order to produce signal. As such, protons flowing through a given slice with sufficient speed to experience only one pulse produce the appearance of a flow void on FSE images. Such imaging is thus termed a “black blood” technique. Slowly flowing blood, on the other hand, may experience both radiofrequency pulses, thus appearing of high SI.

In distinction, in GRE imaging excitation by only one slice-selective radiofrequency pulse is required, since the echo is refocused using the gradients. This results in flow-related enhancement. The SI of stationary tissues in the imaging volume is suppressed by the multiple repetitive RF pulses, while fresh blood flowing into the imaging volume is of high SI as it has not experienced these pulses. Rephrasing this, for further clarification, in TOF imaging background tissue is suppressed due to the use of slice- (or slab-) selective radiofrequency pulses in short succession, at much shorter intervals than the T1 of typical stationary tissue. When subjected to such pulses, stationary protons within a slice have inadequate time to recover longitudinal magnetization, resulting in a low background SI. In distinction, protons from blood flowing into a slice between slice-selective pulses possess a full complement of longitudinal magnetization, thus appearing of high SI on GRE scans. SI will increase proportionally with increasing flow velocity up until the point where all saturated protons in a slice are completely replaced by unsaturated protons. Thus, images acquired with thicker slices (slabs) may not be optimal for depiction of vessels with slower rates of flow. A further complexity is that laminar flow is disrupted in areas of tortuosity, stenosis, or vessel bifurcations. Reductions in SI thus may occur in these areas in TOF MRA.

In 2D TOF techniques, slices are acquired sequentially one-by-one perpendicular to the direction of flow. When suppression of antiparallel flow from venous structures is desired, additional saturation pulses may be applied adjacent to the slice of interest so as to suppress signal from inflowing venous blood. 2D TOF techniques suffer misregistration artifacts if patient motion occurs between slice acquisitions. Elimination of such artifacts is possible with 3D TOF, in which an imaging slab is excited. Such techniques also allow for improved spatial resolution. In 3D TOF, however, blood travels a larger distance within the imaging

volume, resulting in loss of SI from saturation effects, which is more pronounced with slower flow. Thus, 2D TOF is primarily utilized for imaging of slower flowing venous structures, while 3D TOF is the preferred technique for evaluating intracranial arterial structures. Achieving a balance between saturation effects and background tissue suppression necessitates careful selection of the radiofrequency pulse repetition time (TR) and the flip angles used in 3D TOF MRA. With larger flip angles, longitudinal recovery of background tissue protons takes more time, improving background suppression and potentially image contrast. However, the rate at which signal from flowing blood reaches steady-state suppression is also greater, decreasing depth of penetration (specifically visualization of more distal flow) within a slab. Lower flip angles allow better penetration but at the expense of background suppression, with lower SI of inflowing blood as well. Flip angles may actually be varied across the imaging slab, with lower flip angles being used proximally and larger flip angles distally in an attempt to lessen saturation effects. Considerations with TR are similar. With a shorter time between radiofrequency pulses, background suppression is greater. Magnetization transfer, a complex topic (the reader is referred to *The Physics of Clinical MR Taught Through Images*, 4th edition), may be employed to further decrease background SI (with the caveat of poor suppression of fat SI). TE selection is also an important consideration. Turbulent flow at areas just distal to a stenosis represents a mix of fast and slower moving blood. On MR, turbulence results in phase dispersion, and is manifest as a loss of SI. Effects of such phase dispersions are more pronounced on scans obtained with longer TEs, exaggerating vessel stenoses. The minimum TE that can be achieved with 3D TOF MRA using any specific scanner is substantially less than with 2D, leading to less artifactual exaggeration of vessel stenoses. Reduced voxel sizes also lessen variation in phase within such voxels, further diminishing the effects of turbulent flow on 3D TOF.

TOF MRA images may be stacked with post-processing to produce maximum intensity projection (MIP) images, which appear similar to images obtainable by conventional angiography. Such projections lead to additional pitfalls, including misregistration artifacts caused by patient motion between sequential slice acquisitions (most prominent with 2D TOF). Even with 3D TOF such artifacts may be seen at slab interfaces (for example three such slabs are commonly acquired for a circle of Willis MRA exam). Also, not all entities appearing as high SI on TOF MRA represent vascular flow. Since TOF MRA is obtained using GRE T1WI, entities with short T1—such as methemoglobin containing clot—can appear as high SI on such images. Review of source images improves identification of such artifacts, including that from motion and metal. Isometric voxel acquisition allows further flexibility due to the ability to reconstruct high spatial resolution projections in any desired

plane. Examples of pathology seen on TOF MRA of the intracranial vasculature are given in Chapter 12.

Due to decreased acquisition times and reduced problems from flow effects including in plane saturation and turbulence, CE-MRA has replaced TOF MRA for most clinical applications, the main exceptions being circle of Willis (3D) and venous thrombosis (2D) evaluations. CE-MRA allows direct visualization of blood by acquisition of images during the first pass of a gadolinium chelate contrast agent (following its intravenous injection). These agents exit the intravascular space rapidly, necessitating early image acquisition. Images are acquired utilizing fast 3D GRE T1WI with both a short TR and TE. The decrease in blood T1 due to the presence of the gadolinium chelate in the blood leads to increased SI of vascular structures on such scans. Digitally subtracted images, whereby post-contrast images are subtracted from a pre-contrast data set are often of value, aiding in suppression of background SI. Image acquisition is typically timed so as to fill the center of k-space (which encodes data related to image contrast) during the time at which arterial contrast is at a peak. The view order of acquisition of data in k-space can be done in many different ways, two such are linear (sequential) ordering and elliptical centric ordering (the latter commonly used in many CE-MRA applications). The choice of k-space data filling however must be matched with the timing and length of the bolus. Accurate determination of circulation time from the site of injection to the area of imaging interest is essential for bolus timing. For example, if the central portion of k-space is acquired too early before maximum arterial SI, ringing artifacts may be generated, whereas venous enhancement and suboptimal arterial SI may be present if images are acquired with too much of a delay. Correct timing of the image acquisition can be achieved by a variety of techniques, including administration of a test bolus, real-time MR fluoroscopy, and acquisition of dynamic scans. In addition to bolus timing considerations, respiratory artifacts may result in ghosting and blurring. Breath holding is generally necessary for carotid (to achieve good visualization of the vessel origins from the aortic arch), thoracic, and abdominal CE-MRA. CE-MRA of the neck is typically performed during breath-holding, with image acquisition occurring 8 to 10 seconds following bolus IV injection (at a constant rate, using a power injector) of a gadolinium chelate. Imaging must be performed rapidly given the fast transit of the contrast agent from the carotid arteries to the jugular veins. MIP projections of CE-MRA images of the neck are displayed in Figure 99.1A-C. On the (A) right side, there is high grade, near complete stenosis of the internal carotid artery at the bulb. (B) Coronal MIP images illustrate two stenoses (white arrows), with an additional high grade stenosis seen in the internal carotid artery on the left, better visualized on (C) additional MIP images. The left vertebral artery is dominant. Axial images reformatted from the original

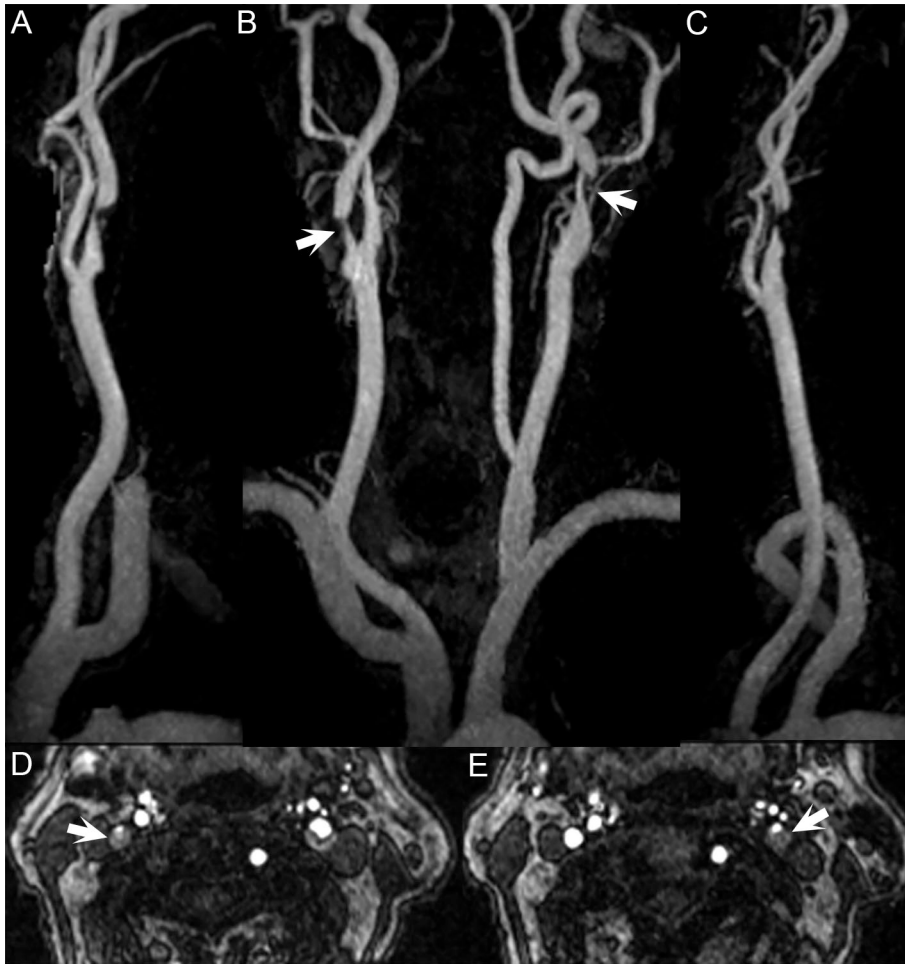


Fig. 99.1

data acquisition, shown in Figure 99.1D,E allow direct visualization of the occlusive atherosclerotic disease present in both the (D, white arrow) right and (E, white arrow) left sides. Evaluation of source images, as in this example, is an important part of scan interpretation, often clarifying findings or artifacts, in addition to providing improved delineation of surrounding soft tissues. Per the NASCET (North American Symptomatic Carotid Endarterectomy Trial) criteria, percentages of stenosis are computed by dividing the luminal diameter at the area of stenosis by the diameter of a normal, more distal portion of the evaluated artery. Both American and European studies demonstrated benefit for carotid endarterectomy in patients with symptomatic stenosis of the internal carotid artery greater than 80%. From an MR imaging perspective, it should be noted that stenoses greater than 95% may result in complete SI loss on CE-MRA. Slow flow distal to such stenoses must be assessed for, as should be collateral flow within the circle of Willis. Incidentally, paragangliomas are typically well visualized as enhancing carotid space masses on CE-MRA, with such an exam suggested for improved lesion visualization and delineation.

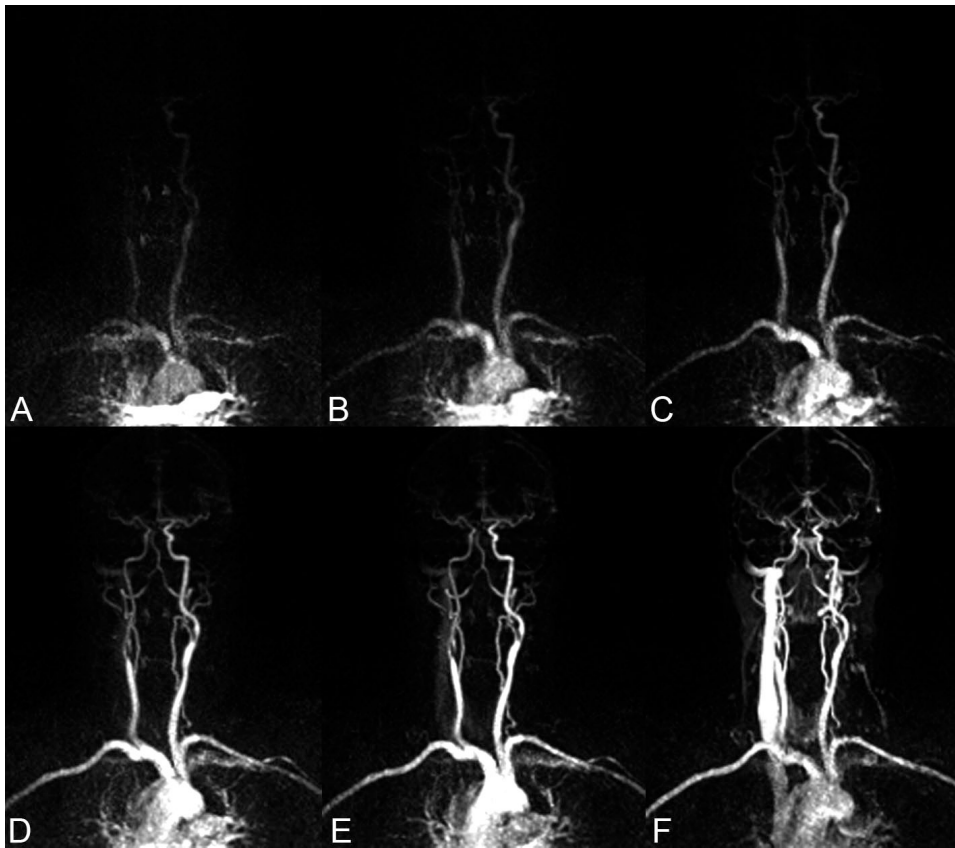


Fig. 99.2

It is also possible to acquire CE-MRA images with high temporal (and to a lesser degree, spatial) resolution utilizing dynamic techniques. The development of view sharing, specifically with techniques such as TWIST (time-resolved angiography with interleaved stochastic trajectories) and TREAT (time-resolved echo-shared angiographic technique) has permitted a marked acceleration in image acquisition. Such techniques are based on the fact that contrast information in MR is primarily derived at the center of k-space. Such techniques rapidly sample the center of k-space multiple times allowing for acquisition of dynamic contrast information, the periphery being sampled less frequently. Because multiple images are acquired at different temporal intervals, test boluses are not needed to time image acquisition. Furthermore, the gadolinium chelate dosage can be decreased. The direction of flow may be determined by such imaging as may separation of arterial and venous components of flow. Such sequential images obtained with TWIST technique are displayed in Figure 99.2A-F. The contrast bolus progressively fills the brachiocephalic artery in Figs. A-C with progressive filling of the remainder of the great vessels in (D-E) later images. The (F) final image demonstrates complete opacification of the circle of Willis and right internal jugular vein. Delayed filling of the right internal carotid artery,

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demonstrates the hemodynamic significance of the high-grade right internal carotid artery stenosis. Dynamic CE-MRA is also useful in the evaluation of subclavian steal, the technique readily identifying reversal of vertebral artery flow to supply the subclavian artery distal to an area of stenosis or occlusion. Such reversal may go undetected on conventional CE-MRA, although phase-contrast MR can similarly detect flow directionality.

CE-MRA can also aid in delineation of other less common pathologies of the great vessels including traumatic carotid or vertebral artery dissections. In these cases, the false lumen fails to fill with contrast on CE-MRA. In comparison to atherosclerotic narrowing, narrowing of the vessel due to a dissection typically involves a longer segment, and spares the carotid bulb.

In regard to atherosclerotic disease, ulcerated plaque is identified on CE-MRA by the presence of focal outpouching within mural thrombus and plaque, while fibromuscular dysplasia exhibits a characteristic “string of beads” appearance. Vasculitis is manifest as on CE-MRA as alternating stenoses and dilatations in the involved arteries. In addition to various congenital vascular abnormalities seen commonly involving the circle of Willis, vascular anomalies of the great vessels are often seen on neck CE-MRA.