

MR Angiography

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Radiology

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1. Discuss Time-of-Flight (TOF) MR angiography and the different aspects of 2D-TOF, 3D-TOF, and multiple overlapping thin slab acquisition (MOTSA) and describe the clinical settings for which each may be of use.
2. Describe Phase Contrast MR angiography, its limitations, and the clinical settings in which it may be used.
3. Explain Contrast Enhanced MR angiography, its benefits in the clinical setting and its potential limitations.

Article

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Introduction

MR angiography (MRA) is a noninvasive technique based on magnetic resonance imaging (MRI), which can produce images of blood vessels similar to those produced by the more invasive catheter angiogram. MRA can be performed with or without injecting intravenous gadolinium (the MR contrast agent) anywhere in the body. The purpose of this paper is to describe the different techniques used for MR angiography. The article is organized according to the MR technique used rather than by the specific blood vessel being imaged. These techniques include time of flight MRA, phase-contrast MRA, and contrast-enhanced MRA.

Time-of-Flight MRA

Time-of-flight (TOF) MRA is based on the phenomenon of "flow related enhancement" or "inflow phenomenon". When fully magnetized spins in blood vessels first enter an imaging volume, they produce high signal. On conventional MR images, this can lead to flow artifacts. On MR angiograms, the effect is harnessed to produce images of the vessel, like an angiogram. In most cases, the clinical question will only involve arteries or veins but not both. In such cases, a 90° pulse is applied above or below the imaging volume to "saturate" (or eliminate) the signal from spins entering in the unwanted venous or arterial vessels. This 90° pulse is referred to as a "spatial saturation pulse" or simply a "SatPulse".

TOF MRA can be performed as a 2D or 3D technique. Two-dimensional TOF MRA consists of single slice gradient echoes, sequentially acquired in the direction countercurrent to flowing blood (so the slices are always meeting fresh, unsaturated, fully magnetized spins, head on). Thus, to image the carotid arteries, you would want to start from the top and go down so that the scan is meeting the up-flowing arterial blood head on ([figure 1](#)). If you were interested in imaging the jugular veins, on the other hand, you would start at the bottom and go up to meet the fresh, down-flowing spins, head on. Using this technique will also allow you to image blood that is flowing in the opposite direction from its normal flow, for example, down flowing blood in the vertebral artery in subclavian steal syndrome.

The slice thickness in 2D-TOF techniques is generally on the order of 2 to 4 mm depending on the application (thinner slices being used for the carotids and thicker slices being used for flow in the portal vein or aorta). The saturation pulse in a 2D-TOF

MRA technique typically travels behind the imaging slice to eliminate unwanted arterial or venous blood.

In 3D-TOF MRA, signal is acquired from a 3 to 6 cm thick slab. This both increases the signal-to-noise (S/N) ratio (compared to the 2 to 4 mm thick slice of 2D-TOF) and decreases the effective slice thickness, improving spatial resolution. Because the slab in a 3D-TOF technique is thicker than the slices used for 2D-TOF MRA, the spins may be exposed to more than one RF pulse as they traverse the slab. This may lead to loss of magnetization and saturation, eliminating signal from the vessel (figure 2). Saturation effects are more pronounced with larger flip angles (which release more magnetization per RF excitation) and shorter repetition times (TR) (which expose the spins to more RF exposures per unit time) in the gradient echo acquisition. There are several ways to decrease saturation effects:

1. Decrease the flip angle (which decreases the signal elicited per RF exposure)
2. Increase the TR (which increases the acquisition time)
3. Decrease the slab thickness (which decreases the coverage)
4. Give gadolinium (which shortens the T1 – see contrast-enhanced MRA below).

MOTSA (Multiple Overlapping Thin Slab Acquisition) is a third TOF MRA technique that consists of multiple 2 to 3 cm thick 3D-TOF slabs covering the anatomy of interest. MOTSA combines the best features of 2D- and 3D-TOF MRA, because it has the unlimited coverage of 2D-TOF and the high spatial resolution of 3D-TOF (figure 3). Occasionally, minor saturation effects will remain, producing low intensity bands at the junctions between the slabs known as "Venetian blind artifacts". To minimize these artifacts, the outer portion of the slab (where saturation effects are greatest) is discarded. Since the discarded portion may be 10% to 50% of the slab thickness, MOTSA is less efficient than 3D-TOF MRA. Typically, the smaller the portion of the slab discarded, the greater the venetian blind artifact.

When limited coverage is needed (for example, an evaluation of the circle of Willis in an acute cerebral infarct), a single 3D-TOF slab can be used. When greater coverage is required (for example, an evaluation of carotid artery stenosis or to screen the entire brain for an aneurysm – figures 4 & 5) MOTSA is used. While 2D-TOF MRA can be used to screen the carotids, MOTSA typically has higher spatial resolution and is more accurate at sizing stenoses than 2D-TOF MRA – although neither is as accurate as contrast-enhanced MRA (see below).

Once a series of 2D or 3D slices has been acquired, they can be viewed directly (as 64 or so separate images), scrolled consecutively on a workstation, or displayed in angiographic format. For the last option, a maximum intensity projection (MIP) algorithm is generally used. The MIP algorithm basically tells the computer to connect the bright dots of inflowing blood on the 64 source images looking from a particular angle or projection. We typically rotate the data about a horizontal and or vertical axis, acquiring 12 images every 15 degrees for a total of 180 degrees. (The back 180 degrees is the mirror image of the front 180 degrees so it is not necessary to repeat it.) When rotating vertically, we typically cut the data set in half so the right and left sides do not overlap. This is called a "targeted MIP".

With increasing gradient strength and advanced image processing, shorter repetition times are possible, making 1024² MRA technically feasible. Intracranial vessels appear much larger with 300 μm pixels than they do at lower resolution. This partially reflects the capture of pixels at the periphery of the artery, where the greatest phase dispersion occurs during laminar flow. With larger pixels, this intravoxel dephasing results in nonvisualization. With higher resolution (smaller pixels), the phase differences from one edge of the pixel to the other are smaller, leading to less phase cancellation. Such improved visualization of true arterial diameter at 250 μm spatial resolution will increase the applications of MR angiography to include vasculitis, spasm following subarachnoid hemorrhage, and detection of even smaller aneurysms.

One of the potential problems with the maximum intensity projection algorithm is that it includes anything that is bright on the T1-weighted source images, including subacute hemorrhage (figures 6, 7, & 8), fat, or anything enhancing with gadolinium. It is therefore important to examine both MRA images and conventional images (figure 6) to determine if something is showing up on the MIP images that doesn't represent flowing blood (figure 7). In such cases, it may be desirable to acquire a phase contrast MRA that subtracts out high intensity saturation structures (figure 8) (see below).

The MIP algorithm has an S/N ratio threshold for what is included in the image, so very small vessels or the residual lumen of tight stenoses may not be visualized. For this reason, it is important to review the individual source images before diagnosing a complete stenosis or absence of a vessel from the MIP images.

The source images are also useful for diagnosing the small lenticulostriate collaterals in moyamoya disease. Moyamoya disease

is a poorly understood disease that involves the occlusion of arteries (such as the internal carotid and the middle cerebral arteries). This results in the development of collaterals around the occlusion that on angiography gives the appearance of a puff of smoke or moyamoya.

Phase Contrast MRA

In phase contrast (PC) MRA, signal is based on the phase gained (or lost) as the spins move through a magnetic field gradient. This phase is subtracted from the background phase to determine the portion of the phase that is only due to motion. Background stationary tissue, which is not moving, subtracts out – as does subacute hemorrhage and any other bright signal that would normally be included in a time-of-flight image (figure 8). The amount of phase gain (or loss) has to be less than 180 degrees (or –180 degrees) or "velocity aliasing" occurs (which means that the spins appear to be moving at maximum velocity in one direction in one pixel and then, in the next pixel, they appear to be moving at maximum velocity in the opposite direction). The strength of the phase encoding gradient is determined by setting a parameter called the "encoding velocity" or "Venc". Since phase sensitization has to be performed along the x, y, and z gradients and then each is subtracted from a baseline image (without gradient activation), this sequence typically takes 4 times longer than a TOF technique with the same TR and matrix. For this reason, it is rarely used except when it is necessary to subtract out subacute hemorrhage or fat (for example, from a ruptured dermoid).

Contrast Enhanced MRA

Contrast enhanced (CE) MRA has become the preferred technique for imaging arteries everywhere in the body except for the brain. With modern echo planar imaging (EPI) capable MRI systems, the stronger, faster gradients allow an image to be acquired during the brief time that gadolinium is in the arterial phase before it gets to the veins. CE MRA is essentially a time-of-flight technique without saturation effects. To image the carotids, for example, a coronal 3D-TOF slab is set up to maximize the field-of-view (FOV), covering from the carotid origins through the siphon. Typically, a mask image is acquired and then contrast is injected. In standard k-space acquisition mode, a series of several 10 sec acquisitions are acquired following injection of gadolinium, one of which is bound to be in the arterial phase. The advantage of this technique is that the exact time of arrival of the gadolinium bolus does not need to be known. When the timing is known (by injecting a 2 cc bolus of gadolinium and noting the time it takes to reach the carotid bulb), a more sophisticated acquisition technique allows a higher resolution image to be acquired in approximately 30 to 60 seconds (figure 9). This technique is known as "elliptical-centric" MRA and refers to the pattern of k-space acquisition in the 3D-TOF technique.

Some MR manufacturers also have automated bolus detection software to determine when the gadolinium bolus arrives at the anatomic level of interest. (For example, GE calls this SmartPrep[®] and Siemens calls it CareBolus[®].) This works particularly well for abdominal and runoff angiography (figure 10) but not as well for carotid MRA. Peripheral MRA is also facilitated by the use of a stepping table which first acquires a mask image at three stations moving from inferior to superior and then, after the gadolinium injection, the table chases the bolus from superior to inferior during image acquisition. While this technique works in the vast majority of cases, it is sometimes necessary to perform a 2D-TOF acquisition from inferior to superior through the leg, should there be proximal stenoses and asymmetric delivery of the contrast to the legs (particularly in diabetic patients).

Figures:

1. 2D-TOF Carotid MRA. LAO projection demonstrates normal carotid and vertebral arteries.
2. Saturation Diagram. As spins traverse a 3D-TOF slab, they lose magnetization in proportion to the flip angle (FA). During time TR they recover magnetization according to the equation $1 - \exp(-TR/T1)$. The shorter the TR, the larger the flip angle, and the longer the T1, the sooner saturation (loss of signal) occurs.
3. MOTSA through a normal Circle of Willis.
4. MRA of an Intracranial Aneurysm. A MOTSA demonstrates a cavernous carotid aneurysm.
5. MRA of an Intracranial Aneurysm. For comparison, a lateral projection from a catheter angiogram.
6. Subacute Subdural Hemorrhage. A T1-weighted image shows subacute subdural hematoma with high signal methemoglobin.
7. Subacute Subdural Hemorrhage. A 3D-TOF MRA is degraded by the bright subacute subdural hematoma.
8. Phase Contrast MRA with Subacute Hemorrhage. A PC MRA subtracts out the high signal stationary subdural hematoma.
9. CE MRA of Carotid and Vertebral Arteries. Elliptical-centric k-space acquisition shows right carotid stenosis.
10. CE MRA of the aorto-iliac system demonstrates normal renal arteries and left iliac artery occlusion.

Figure 1

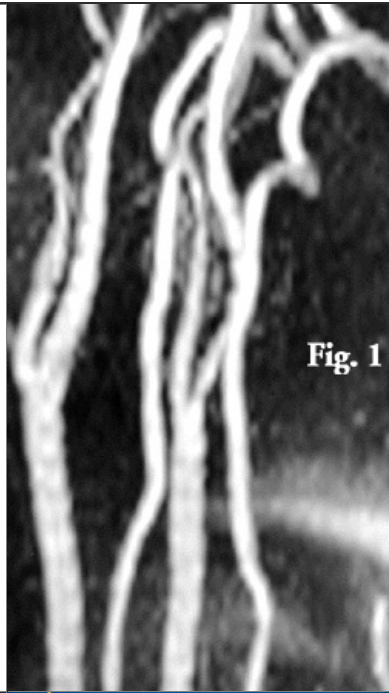


Figure 2

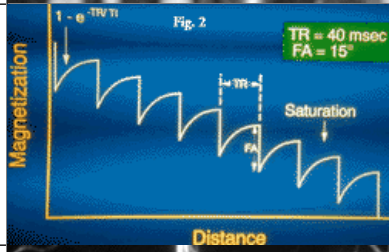


Figure 3

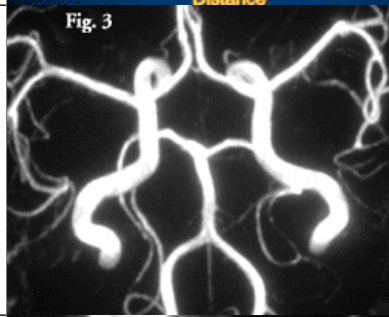


Figure 4

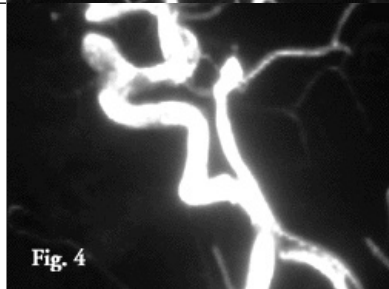
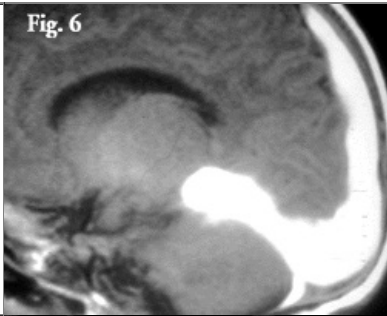
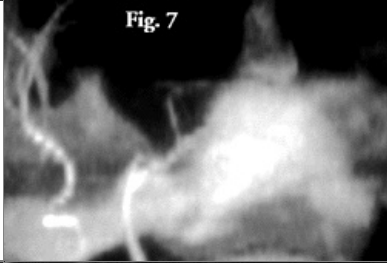
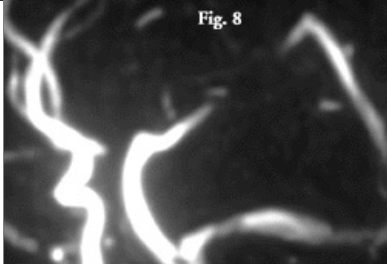
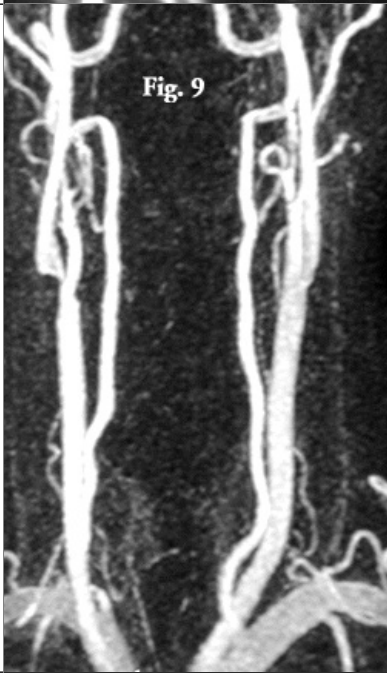
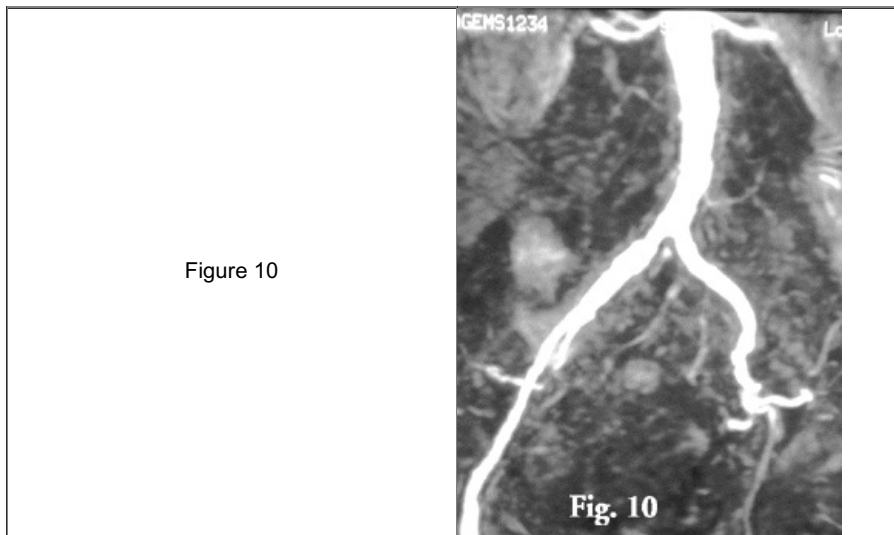


Figure 5



<p>Figure 6</p>	<p>Fig. 6</p> 
<p>Figure 7</p>	<p>Fig. 7</p> 
<p>Figure 8</p>	<p>Fig. 8</p> 
<p>Figure 9</p>	<p>Fig. 9</p> 



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Dr. William Bradley currently is the director of the Magnetic Resonance Imaging Center at Long Beach Memorial Medical Center, in Long Beach, California. He is also a Professor of Radiology at the University of California, Irvine. He actively teaches Magnetic Resonance Imaging to medical students, Radiology residents and fellows in Radiology.

Dr. Bradley has over 100 publications in peer-review journals and is actively involved in research in the field of Magnetic Resonance Imaging. He has presented his research and has given lectures on MRI topics at major conferences around the country as well as internationally, including Europe, Japan, and India.

