

Basic Principles of Phase-contrast, Time-of-flight, and Contrast-enhanced MR Angiography

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The most widely used MR angiographic imaging techniques can be categorized as *phase contrast*, *time-of-flight*, or *contrast-enhanced* methods. The basic physical principles of each of these techniques are described briefly below.

Phase contrast techniques derive contrast between flowing blood and stationary tissues by manipulating the phase of the magnetization, such that the phase of the magnetization from the stationary spins is zero and the phase of the magnetization from the moving spins is non-zero. The phase is a measure of how far the magnetization precess from the time it is tipped into the transverse plane until the time it is detected. The data acquired with phase contrast techniques can be processed to produce phase difference, complex difference, and magnitude images.

In phase difference images, the signal is linearly proportional to the velocity of the spins - faster moving spins give rise to a larger signal, and spins moving in one direction are assigned a bright (white) signal, whereas spins moving in the opposite direction are assigned a dark (black) signal. Thus, the vascular anatomy can be assessed, and the speed and direction of the blood flow can be qualitatively determined, by observing the phase

difference images. In addition, quantitative information regarding the velocity and volume flow rate of the blood can be derived from the phase difference images. To assess flow information, the images must demonstrate the cross section of the vessel so the area can be determined. The product of the area and the average velocity over the vessel yield the volume flow rate.

In the complex difference images also, the signal strength is dependent on the velocity of the spins, but it is not a linear dependence as it is in phase difference processing, nor is the direction of blood flow represented. Therefore, complex difference images are not used to determine quantitative information, but are used for demonstrating the anatomy of the vessels.

Conventional magnitude images also can be constructed from the acquired data because phase contrast techniques do not destroy the magnitude information - they simply alter the phase of the magnetization in a controlled manner.

Phase contrast methods are implemented using two-dimensional or three-dimensional acquisition. Two-dimension acquisition can be completed rapidly and is effective for localizing. Two-dimensional acquisition also can be cardiac gated to provide images of the vessels throughout the cardiac cycle. If the cardiac-gated images are acquired perpendicular to the direction of flow, and phase difference processing is performed, flow information throughout the cardiac cycle can be obtained.

Three-dimensional acquisition is considerably more time consuming than two-dimensional acquisition and so it currently is used less frequently than two-dimensional acquisition. Due to the long scan time associated with three-dimensional acquisition, it currently is not cardiac gated. Benefits of three-dimensional acquisition include an

inherently high signal-to-noise ratio, small voxels, and a shorter echo time (TE) than thin-slice two-dimensional acquisition. Also, three-dimensional data sets can be reprojected or reformatted to permit observation of the vessels from any orientation. Phase difference-processed images reformatted perpendicular to the vessels can be used to obtain information regarding volume flow rate.

Phase contrast methods are sensitive to a range of velocities. To specify this range of velocities, the user chooses a velocity-encoding (Venc) value. Blood velocities higher than the Venc value will be misrepresented in the image, so the user must choose this value carefully. Different velocity encoding values can be used in different scans to highlight different vessels. For example, this is an effective means of producing separate images of the feeding arteries, the draining veins, and the nidus of an arteriovenous malformation, each of which contain blood flowing in different velocity ranges. It also is useful for demonstrating the fast flow in the in-flow jet of a giant aneurysm in a separate image from the slow stagnant flow in the center of the aneurysm.

In order to encode flow in all directions, a flow-encoding gradient must be applied on each of the three gradient axes in separate TR intervals. In addition, a fourth non-flow-encoded acquisition must be acquired. This non-flow-encoded acquisition is subtracted from each of the three flow-encoded acquisitions to eliminate phase accumulation from sources other than velocity. The need to acquire four acquisitions to encode flow in all directions lengthens the scan time. The subtraction results in high contrast between vessels and stationary tissues, permitting large fields-of-view to be acquired without detrimental effects from saturation, as long as a relatively small tip angle (20° - 30°) is used.

Scan time for two-dimensional acquisitions is $4 \times TR \times NSA \times \#PE$, where NSA is the number of signal averages, and #PE is the number of phase-encoding values acquired. The scan time for three-dimensional acquisitions is $4 \times TR \times NSA \times \#PE \times \#SE$, where #SE is the number of slice-encoding values acquired and all other abbreviations are as described above.

Time-of-flight techniques derive contrast between flowing blood and stationary tissues by manipulating the magnitude of the magnetization, such that the magnitude of the magnetization from the moving spins is large and the magnitude of the magnetization from the stationary spins is small. This leads to a large signal from moving blood spins and a diminished signal from stationary tissue spins.

In MR, the signal from spins decreases with exposure to an increasing number of excitation pulses, until eventually a saturation value is reached. In time-of-flight imaging, the goal is to subject the flowing spins to only a very few excitation pulses, and to subject stationary spins to a large number of excitation pulses, thereby achieving a signal difference between blood and stationary tissues. This can be achieved by imaging planes, or thin slabs, oriented perpendicular to the main direction of flow. When this is done, the moving spins enter the slice fully magnetized, experience only a few excitation pulses, and then flow out of the slice. This ensures that the signal from the blood will be relatively large because the blood is continuously refreshed during image acquisition, and it, therefore, never experiences enough excitation pulses to become saturated. The stationary tissues, however, remain in the slice, or slab, throughout image acquisition, and so they give rise to a diminished signal because the magnetization from them is saturated due to the constant exposure to the excitation pulses.

The number of excitation pulse experienced by moving spins as they traverse the imaging slice is dependent on the thickness of the slice, the velocity of the blood, the orientation of the vessel, and the TR of the imaging sequence. In general, thinner slices, faster flowing blood, vessels oriented perpendicular to the slice, and a longer TR lead to increased vascular signal. A long TR, however, also leads to increased signal from stationary tissues, so an intermediate TR must be selected. Increasing the tip angle leads to diminished signal from stationary tissues, but it can also lead to increased saturation of blood that experiences multiple excitation pulses, so the tip angle must be carefully selected. These factors and others must be carefully considered when designing a time-of-flight protocol.

Time-of-flight methods can be implemented using two-dimensional or three-dimensional acquisition. For two-dimensional acquisition, data are acquired from multiple slices stacked contiguously along the vessels of interest. The data from the slices can be reprojected or reformatted to demonstrate long segments of the vessels. With two-dimensional acquisition, the slices are thin (1 – 3 mm), increasing the likelihood that the blood experiences only a very few excitation pulses as it flows through the slice. Thus, a large tip angle (60°) can be used to suppress the signal from the stationary tissues without suppressing the signal from blood. Two-dimensional acquisition can be cardiac gated to reduce signal ghosting in images caused by cardiac pulsatility.

For three-dimensional acquisition, a slab oriented perpendicular to the vessels of interest is imaged and the slab is encoded into thin slices using an encoding method similar to that used for phase encoding. Because a slab is imaged, a small tip angle (30°) must be used so the signal from blood that remains in the slab does not become saturated.

The small tip angle necessary to preserve signal from blood also causes a preservation of signal from stationary tissues. Therefore, when three-dimensional acquisition is employed, other mechanisms must be implemented in order to reduce the signal from stationary tissues, as described below.

One method employed to diminish signal from stationary tissues in three-dimensional time-of-flight is the use of magnetization transfer. With magnetization transfer, an off-resonance pulse is applied at the start of each TR to saturate the magnetization from macromolecules. Signal from macromolecules does not appear in MR images because the T₂ of the signal from these molecules is too short. When the magnetization from these macromolecules becomes saturated, nearby water molecules can transfer their magnetization to the macromolecules, leading to a diminished water signal in tissues that contain macromolecules. Grey and white matter in the brain contain macromolecules whereas blood does not. Thus, magnetization transfer can be used to diminish the signal from the brain, leading to increased contrast in intracranial three-dimensional time-of-flight angiograms. A drawback of this feature is that it leads to an increased TR resulting in a longer scan time, and an increase in the deposited energy.

The contrast in three-dimensional time of flight angiograms can be further increased by choosing an echo time that ensures that magnetization from fat and water are 180° out of phase with one and other at a critical point during signal detection so that the signals from them cancel each other. Fat and water precess at different rates, so at certain times, the magnetization from them is opposed. At a magnetic field strength of 1.5T, fat and water are at an opposed phase at echo times of 2.3 msec and 6.9 msec. Choosing

these echo times is an effective means of suppressing signal from tissues that contain fat and water.

Another mechanism used to achieve higher contrast between flowing blood and stationary tissues in three-dimensional acquisition is the use of a ramped tip angle. A ramped tip angle is used to reduce the saturation of blood signal as the vessels penetrate farther into the slab. The tip angle is ramped such that a small tip angle is applied where the vessels of interest enter the slab, and a large tip angle is applied where the vessels of interest exit the slab. The small tip angle at the entrance of the slab prevents the blood from becoming too saturated as it traverses the slab. The large tip angle at the exit of the slab tips a large component of the almost fully saturated blood into the transverse plane to be sampled just before it leaves the slab, providing a larger signal than would be achieved by using a small tip angle. The large tip angle saturates the blood, but this is inconsequential, because the saturated blood will exit the slab. The ramped tip angle provides a more uniform signal along the vessels as they traverse the slab, provided they follow a fairly direct path through the slab. A drawback of this imaging feature is that vessels that run parallel to the edge of the slab in the region of the large tip angle will become saturated.

Even with these imaging features, contrast between blood and stationary tissues can be small in the three-dimensional time-of-flight acquisition, especially when thick slabs are used to achieve adequate coverage. To achieve greater coverage with reduced saturation effects, multiple thin slabs along the vessel can be imaged. This method combines the thin slice benefits of two-dimensional acquisition with the benefits of three-dimensional acquisition, including an inherently high signal-to-noise ratio, small voxels,

and a short TE. Even this multi-slab method can suffer from saturation from slowly flowing blood at the exit of each slab. This saturation effect is referred to as the slab boundary artifact, or the venetian blind artifact.

With both two-dimensional and three-dimensional acquisition, a spatial saturation pulse can be applied just outside the slice, or slab, at the beginning of each TR to eliminate signal from blood that is going to flow into the imaging slice. This is an effective means of eliminating signal from venous blood that is going to flow into the imaging slice, or slab, and if left unsaturated would interfere with observation of arteries.

Contrast-enhanced techniques derive contrast between blood and stationary tissues by manipulating the magnitude of the magnetization, such that the magnitude of the magnetization from the moving spins is large and the magnitude of the magnetization from the stationary spins is small. Signal differences in contrast-enhanced techniques are achieved not only by employing the appropriate sequence parameters (as is the case in time-of-flight techniques), but also by intravenously injecting a contrast agent into the vascular system to selectively shorten the T1 of the blood. By implementing a T1-weighted imaging sequence during the first pass of the contrast agent, images can be produced that show arteries with striking contrast relative to surrounding stationary tissues and veins.

Dramatically shortening the T1 of blood causes the blood to give rise to a very large signal. The large blood signal is minimally affected by intravoxel dephasing, which is typically caused by complex flow and susceptibility variations. Because the contrast-enhanced techniques are relatively insensitive to signal loss, they provide high quality images with fewer artifacts than the non-contrast-enhanced methods. Because effects of

saturation are minimal, large fields of view can be imaged to demonstrate large vascular areas in a short acquisition time. The short imaging time permits acquisition in a single breath-hold interval, providing high quality images even in areas affected by respiratory motion.

Synchronizing the acquisition with the arrival of the contrast agent is critical to image quality. If the data are acquired before the arrival of the contrast agent, the vessels will not appear in the image. If the data are acquired too late, the arterial signal will be diminished, and the veins and stationary tissues will be enhanced.

Several methods have been developed to ensure proper timing of the acquisition relative to the passage of the contrast agent. In one method, a small test bolus is injected, and then images are rapidly and repeatedly acquired. From these images the arrival time of the contrast agent can be determined and used to calculate when to start the acquisition of the angiogram after injecting the full bolus. Other methods monitor the signal in a volume or an image and begin acquisition of the angiogram when it has been determined that the contrast has arrived. Two- and three-dimensional time-resolved angiographic methods also can be used to continuously image during the passage of the contrast agent. In addition to demonstrating the peak arterial frame, time-resolved methods provide some information regarding the hemodynamics.

Contrast agents that remain in the blood pool for several hours have been developed recently and currently are being evaluated. These intravascular agents can be used to increase the imaging time in order to get greater coverage and greater spatial resolution. The drawback of increased acquisition time is that it permits venous enhancement, which leads to difficulty in evaluating the arteries. An additional benefit of

some of the intravascular agents is that they provide greater vascular signal by causing a more dramatic decrease in the T1 of blood.

When properly implemented, all of the MR angiographic methods can yield diagnostic-quality images. The objective of this presentation is to introduce some of the physical principles that affect image quality, so that they can be understood and utilized to consistently obtain diagnostically useful images.