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Disappearing Iliac Vessels: An MR Phase Cancellation Phenomenon

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Abstract

Magnetic resonance images often fail to distinguish the iliac vessels on one side. Phantom studies and theoretical analysis demonstrated that this phenomenon arises from a loss of the flow void effect. At a certain angle, the additional phase accumulation that results from motion in the section-selection direction exactly cancels the additional phase accumulation from motion in the frequency-encoding direction. Thus, vessels that cut through the imaging plane near this critical angle have substantial signal intensity. This artifact can be eliminated by alternating the direction of the section-selection gradient on sequential phase encodings.

Index terms

Arteries, iliac, 986.1214; Arteries, MR studies, 986.1214; Magnetic resonance (MR), physics; Magnetic resonance (MR), technology; Veins, iliac, 987.1214; Veins, MR studies, 987.1214

One of the great assets of magnetic resonance (MR) imaging is the ability to image the vasculature non-invasively. However, if this property is to be clinically useful, vessels with similar flow must always demonstrate the same signal intensity. Unfortunately, on routine clinical images we often noted differences in signal intensity arising from the midpelvic portion of the external iliac vessels. We confirmed this observation with a retrospective study of 20 patients with no known vascular disease. The signal intensity from left and right external iliac vessels was compared. In all cases different signal was seen in the two vessels, with higher signal intensity always being seen on the right side. In five cases the increased signal intensity on the right side was sufficient to preclude visualization of the vessels from the adjacent psoas muscle (disappearing iliac vessel).

However, by chance a patient was imaged with an inverted section-selection gradient. To our surprise, the asymmetric signal intensity reversed with inversion of the section-selection gradient (Fig. 1), with high signal then being noted on the left side. This confirmed the artifactual nature of the asymmetry in signal intensity.

We studied this phenomenon with a flow phantom that could be obliquely oriented in the imager to simulate the oblique course of the iliac vessels. In certain orientations, the familiar flow void was seen. However, in other oblique positions, substantial signal intensity was seen in the flowing fluid. This confirmed the clinical observation.

A theoretical explanation of this phenomenon was then derived that accurately predicted the angular variation and dependence on section thickness of the intraluminal signal intensity. With this understanding, we then developed a pulse sequence that reduces the effect of this artifact.

Materials and Methods

All studies were performed on a 0.15-T resistive system (Teslacon; Technicare, Cleveland). Multisection spin-echo sequences were used, with a TR of 500–1,000 msec and a TE of 30 msec.

A flow phantom (Fig. 2) was constructed with $\frac{5}{8}$ -inch inside diameter plastic tubing in a Plexiglas frame. Three non-flowing tubes and a U-shaped flow tube were incorporated. All tubes were filled with a blood equivalent solution (150 μ mol of manganese chloride). Pulsatile flow was simulated by intermittent occlusion of a gravity-fed system. The person who occluded the tube provided cardiac gating by timing the occlusions to his arterial pulse. The average flow rate through the U-tube was calculated to be 10 cm/sec.

Transverse sections 10 mm thick were obtained with the flow phantom obliquely oriented to the z-axis of the magnet by the following angles (θ): -36° , -24° , -18° , -12° , 0° , 12° , 24° , and 36° . In the -18° position, the section-selection (G_z) and frequency-encoding (G_x) gradients were inverted by software and hardware modifications. In addition, 5-mm-thick sections were imaged with the phantom in the -18° and -36° positions.

Results

With flow parallel to the z-axis (0°), the familiar flow void was observed in the phantom. The phantom was then angled at -12° , -18° , -24° , and -36° to the z-axis. With 1-cm-thick sections, increasing signal intensity was observed within the flowing tube (Fig. 3), with a maximum between -18° and -24° (Fig. 4). With the phantom angled in the opposite direction (12° , 24° , and 36°), low signal intensity was observed within the flowing tube. With the phantom near the maximal signal intensity position, reversal of either G_x or G_z resulted in a flow void (Fig. 5).

When the section thickness was reduced from 10 to 5 mm, the orientation that demonstrated maximal signal intensity increased to -36° .

Analysis of Results

Gradients are used in MR imaging to obtain spatial information in the x-, y-, and z-axes. Section-selection and frequency-encoding gradients are usually balanced gradients, which for stationary spins do not result in net accumulation of phase (ϕ) during the sequence. In contrast, the phase-encoding gradient (G_y) induces phase changes to encode spatial information.

The balanced gradients (G_z , and G_x) are designed so that the phases accumulated between the 90° and 180° pulses are exactly canceled by the phase accumulated between the 180° pulse and the center of the echo (Fig. 6). This is true for spins that do not move during the image sequence. The cancellation will not be correct when there is motion in either G_x or G_z because the rate of phase accumulation varies as the spins move through the gradient (1). Thus, in general, spins moving with a constant velocity in some direction are going to experience a phase shift in each line of data obtained. This constant phase shift does not affect the signal magnitude, but the phase can be exploited to assess flow direction (2).

If, on the other hand, flow is not constant but pulsatile, the additional phase accumulated because of motion will be erratic. Varying phase shifts lead to a misregistration in the phase-encoding direction of signal from the pulsatile blood flow. Consequently, a flow void is seen within the vessel in most orientations. However, in the particular orientation where the additional phase accumulated in G_x just cancels the additional phase accumulated in G_z (Fig. 6), there will be no net phase accumulation from motion for any velocity. Hence, the flow void disappears.

With the help of Figure 7, the above discussion can be expressed mathematically as follows:

$$\begin{aligned}\phi_x &= \frac{\gamma}{2\pi} \left[\int_0^{TE/2} G_x(x_0 + vt\sin\theta)dt - \int_{TE/2}^{TE} G_x(x_0 + vt\sin\theta)dt \right], \\ \phi_z &= \frac{\gamma}{2\pi} \left[\int_0^{TE/2} G_z(z_0 + vt\cos\theta)dt - \int_{TE/2}^{TE} G_z(z_0 + vt\cos\theta)dt \right],\end{aligned}\quad (1)$$

where G_x and G_z are the two gradients, x_0 and z_0 define the location of the moving spins at the start of the sequence, and v is the velocity of flow, as shown in Figure 6. Since the gradients are balanced so that the phase for stationary spins is zero, the terms in x_0 and z_0 can be ignored, and the additional phases ($\Delta\phi_x$ and $\Delta\phi_z$) resulting from motion are as follows:

$$\begin{aligned}\Delta\phi_x &= \frac{\gamma}{2\pi} \left[\int_0^{TE/2} G_x(vt\sin\theta)dt - \int_{TE/2}^{TE} G_x(vt\sin\theta)dt \right], \\ \Delta\phi_z &= \frac{\gamma}{2\pi} \left[\int_0^{TE/2} G_z(vt\cos\theta)dt - \int_{TE/2}^{TE} G_z(vt\cos\theta)dt \right].\end{aligned}\quad (2)$$

Under the particular geometric conditions where these additional phases ($\Delta\phi$) cancel, $\Delta\phi = \Delta\phi_x + \Delta\phi_z = 0$, a relationship governing θ can be established that is independent of velocity. Calculation of the integral for the gradient strengths and durations used in our experimental conditions, including the finite rise and decay times, yields a value for θ of -18° . The solution for θ is sensitive to the accuracy of the gradient balancing and the decay times; thus, the experimental results showing the maximum between -18° and -24° are in excellent agreement with these calculations.

Several direct consequences of this analysis are borne out by the experimental results: (a) reversing the direction of flow does not change the angle for disappearance of the vessel, since the signs of both $\Delta\phi_x$ and $\Delta\phi_z$ change; (b) reversing the G_z changes the sign of $\Delta\phi_z$ in equation (2), resulting in a solution for $\Delta\phi = 0$ at $\theta = +18^\circ$; (c) reversing the G_x changes the sign of $\Delta\phi_x$ in equation (2), also resulting in $\theta = +18^\circ$; and (d) halving the section thickness is achieved by doubling G_z , which changes the integral, resulting in twice the negative angle. These consequences are all borne out by the experimental results reported in the previous section.

It is of interest to note how components of velocity in the phase-encoding direction (G_y) contribute to the foregoing discussion. First, for equal components of velocity in the x and y direction and the maximum phase-encoding gradient G_y , the additional phase $\Delta\phi_y$ is only 6% of $\Delta\phi_x$. Second, because G_y changes with each phase-encoding step, $\Delta\phi_y$ also changes, even under the conditions of constant velocity. Thus, no angle in the y direction yields a phase error

that could cancel phase errors in either x or z. Therefore, velocity components in y lead to small phase errors and subsequent loss of signal but do not exhibit any phase cancellation phenomena.

Discussion

Reliable demonstration of a flow void in vessels is crucial if MR is to be clinically useful in the diagnosis of vascular thrombosis, compression of vessels, or differentiation of nodes from vessels. We have demonstrated both clinically and in phantom studies an artifact that causes high signal intensity in obliquely flowing vessels of a particular orientation. This artifact arises as a result of the interaction between the section-selection gradient and the frequency-encoding gradient. When flow is pulsatile and not synchronized with the imaging repetition time, random phase shifts occur in both the x and z directions. If these phase shifts do not cancel, the phase encoding is scrambled and the intensity from the vessel is splayed out in the phase-encoding direction, resulting in the familiar flow void. However, at a particular angle of the vessel there is fortuitous matching of phase shifts induced by the two gradients, with the sum of these equal to zero. This is independent of both the velocity and the direction of flow. However, as it is dependent on the orientation of the vessel, cancellation occurs at -18° but not at $+18^\circ$. This is the reason the flow void is seen only on one side.

The particular angle at which the flow void in a vessel disappears depends on the actual pattern of gradients. For example, decrease in the section thickness accomplished by increasing the strength of the section-selection gradient increases the particular angle.

The artifact can be eliminated by alternating the direction of the section-selection gradient on sequential phase encodings. This is simulated in Figure 8, where the data lines from two sequences are interleaved. The even G_y data lines are from a sequence with a normal G_z , and the odd G_y lines are from a sequence with an inverted G_z . This synthetic data set was then reconstructed with appropriate phase corrections on a computer (VAX 11/780; Digital Equipment, Maynard, Mass.). The resultant image (Fig. 8) demonstrates a flow void in both left and right iliac arteries, yielding satisfactory visualization.

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Figure 1. MR images of normal pelvis (echo time [TE], 30 msec; repetition time [TR], 500 msec). **(a)** Normal imaging sequence in which the vessels on the patient's left are visible (arrow). **(b)** With section-selection gradient reversed, the vessels are visible on the patient's right side (arrow).

Figure 2.
Flow phantom consisting of three stationary tubes and a U-shaped tube oriented at an angle θ to the imaging plane.

Figure 3.
Images of flow phantom as a function of angle.

Figure 4.

Intensity of signal from stationary and pulsatile flow as a function of angle θ . Dashed lines indicate signal intensity within the flowing tube (inflow and outflow). Solid line indicates nonvarying signal in the stationary tubes. The lack of flow void phenomenon is evident around -18° .

Figure 5. Images of phantom oriented at -18° show that inversion of either the section-selection or read gradients induces the flow void. Left, normal sequence; middle, section gradient inverted; right, read gradient inverted.

Figure 6. Vector diagram demonstrates velocity components in the x ($V_{\sin\theta}$) and z ($V_{\cos\theta}$) directions. Interaction of these velocity components with the gradients causes the phase shifts.

Figure 7.

Timing diagram for the conventional spin-echo sequence. The two regions of oblique shading are equal in area for the static spins, resulting in an accumulative phase of zero at the echo. However, for spins moving in the negative x and positive z directions, the additional phase accumulation from motion (represented by the difference in the shaded areas) is not necessarily zero.

Figure 8. Synthesized image with alternating direction of the section-selection gradient on sequential phase-encoding acquisitions. There is adequate visualization of the iliac vessels on both sides.