

MR Flow Imaging by Velocity-Compensated/Uncompensated Difference Images

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Abstract: The phase shifts acquired by motion of excited spins along magnetic field gradients can result in decreased signal intensity from blood vessels in conventional magnetic resonance images. The imaging technique can be modified with the use of additional gradient pulses so as to either compensate these phase shifts and increase the signal from the vessels or augment the phase shift and decrease the signal, without altering the signal from stationary tissues. Making a difference image from images made with and without sensitization to motion will cancel out the stationary tissues, leaving an image of the vessels alone. The technique does not require cardiac gating, shows veins as well as arteries, and can be performed in an interleaved manner to avoid registration errors due to patient motion. **Index Terms:** Blood flow, flow dynamics—Magnetic resonance imaging, techniques—Magnetic resonance imaging.

Blood flow effects in magnetic resonance (MR) imaging arise from two principal sources (1): time-of-flight effects due to washout of saturated (2) or excited (3) spins from the imaging region, and phase shifts acquired by excited spins moving along magnetic field gradients (4). The amount of phase shift acquired by a moving spin depends on both the velocity and the sequence of magnetic field gradients it experiences (5-8). When spins with a range of velocities are included in a picture element, the corresponding range of phases will result in a decreased signal intensity.

Macovski proposed the utilization of MR blood flow effects in several approaches to vessel imaging (9), where the signal from stationary tissue is suppressed, leaving an image solely of vessels. Weeden et al. used one of these approaches, namely that of subtraction of image data acquired at two points in the cardiac cycle, to produce vessel images (10). The approach that we have used differs in this way (although it is implied in ref. 11): We can modify a conventional MR imaging pulse sequence with additional magnetic field gradient pulses to compensate for the motion-induced phase shifts so that there will be no net phase shift. Alternatively, we can modify the pulse sequence so that

moving spins will acquire even greater phase shifts, without affecting the signal from stationary spins. Taking the difference of the images thus produced will cancel out the stationary tissues, leaving only the images of vessels containing flowing blood. Alternatively, taking the (complex) difference of the corresponding signals and reconstructing an image from the difference signals will yield an image of the vessels alone.

MATERIALS AND METHODS

The phase shift acquired by an excited spin moving in the presence of a magnetic field gradient will be given by

$$\phi(t) = \int_0^t \gamma \vec{r}(t') \cdot \vec{G}(t') dt \quad (1)$$

where γ is the gyromagnetic ratio, $\vec{r}(t')$ is the position as a function of time, and $\vec{G}(t')$ is the gradient. Any 180° radiofrequency pulses will effectively reverse the current value of the phase. The conditions of zero net phase shift for spins that are stationary, or have constant velocity, constant acceleration, etc., can be separately or jointly satisfied with suitable pulse sequences (12). In particular, suitable balanced pairs of magnetic field gradient pulses are used to control the net phase shifts acquired by motion along the magnetic field gradients used in the imaging sequence. Signals can be ac-

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quired with motion-induced phase shifts either minimized or set to a desired level.

This technique was implemented on a small bore research MR system operating at 1.4 T. The schematic timing diagram of the velocity-compensated/uncompensated (VCUPS) imaging pulse sequence for projection imaging is shown in Fig. 1. There are different effects of motion along different directions in conventional two-dimensional Fourier transformation imaging. Just as with respiration, any motion-induced inconsistencies in phase or intensity between successive phase-encoded data acquisitions resulting from pulsatile flow will result in artifacts propagating along the corresponding phase-encoded direction in the image. Displacement occurring between the times of excitation and position encoding is minimized by keeping the echo time relatively short. There is an additional displacement effect of the primary image along the phase-encoded direction due to phase shifts resulting from motion along this direction; this is compensated for in our imaging technique by the second lobe in the G_x pulse sequence in Fig. 1. An additional balanced gradient pulse pair could be added to the phase-encoding gradient to produce a phase shift due to motion along this direction. Data are acquired alternatively with and without compensation for motion-induced phase shifts. The interleaved data are separated for reconstruction of magnitude images with or without phase compensation. These images are subtracted to yield a difference image showing only the areas of motion (vessels). Alternatively, the paired data can first be subtracted (explicitly including phase) and the difference used to directly reconstruct the vessel image.

RESULTS

Images of a hand obtained with the VCUPS imaging sequences in Fig. 1 are shown in Fig. 2; the resulting difference image is shown in Fig. 3. As another example, the difference image of the head and thorax of a rat obtained with the same sequence with a smaller field of view is shown in Fig. 4.

DISCUSSION

The dominant flow effect on intensity in the uncompensated projection images arises from the motion-induced phase shifts, producing loss of almost all signal from the vessels; the resulting difference images are relatively independent of velocity above a fairly low threshold and thus insensitive to pulsation. As a result, both arteries and veins are well demonstrated, as is seen in the rat images in Fig. 4.

Although flow effects (including phase shifts) have long been recognized in MR, efforts to use them for imaging are relatively recent. The velocity dependence of phase shifts and other flow-related phenomena has been exploited by using conventional imaging with cardiac cycle synchronization ("gating") and reconstructing the images so as to display the phase (13); moving regions stand out from adjacent stationary ones by their different phase. By modifying the imaging sequence, the sensitivity of the phase shift to velocity can be adjusted (14,15). An alternative approach has been to acquire conventional gated MR images of a given location at different phases of the cardiac cycle; when these images are subtracted the resulting difference image will ideally show only vessels with

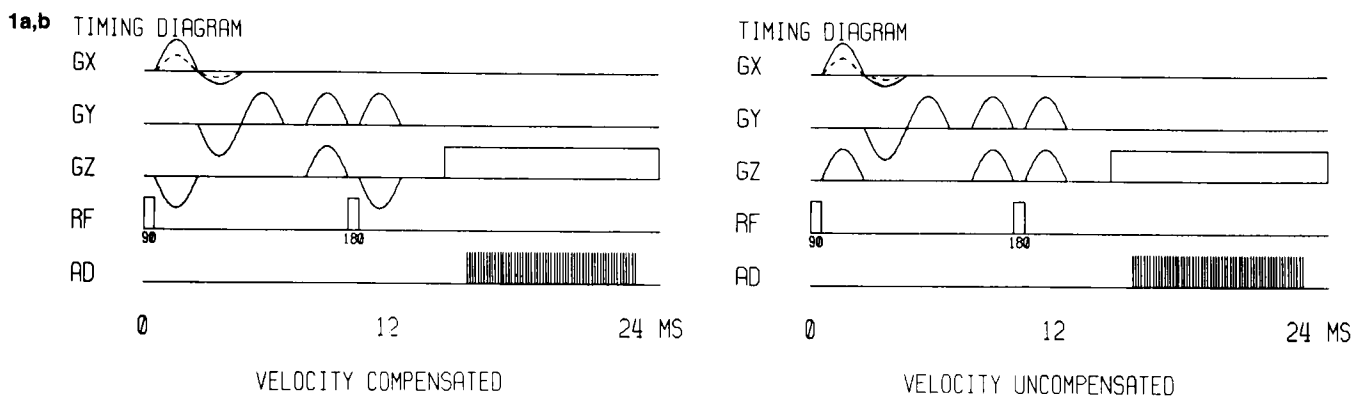


FIG. 1. Pulse sequence timing diagram of velocity-compensated/uncompensated projection imaging technique. The radiofrequency (RF) pulses are shown as nonselective although they could be made selective. **a:** Velocity-compensated pulse sequence. Excitation by 90 and 180° RF pulses produces spin echo detected and digitized as indicated by AD. The G_x gradient provides frequency encoding of position. G_x provides phase encoding of position; broken lines denote change in strength in successive signal acquisitions. G_y suppresses the effects of imperfection in the 180° pulse. **b:** Velocity-uncompensated pulse sequence. Polarity of velocity compensating pulses in G_x has been reversed to sensitize to motion along this direction.

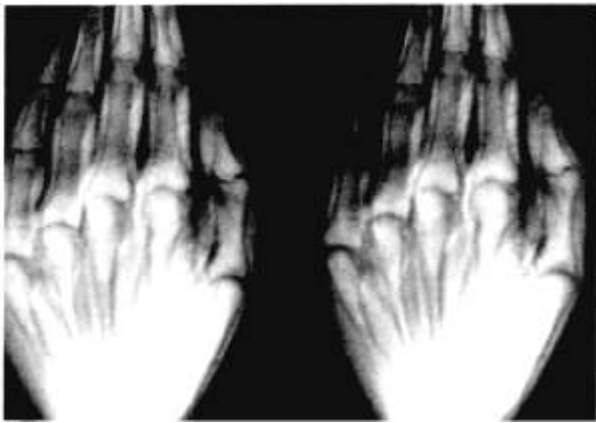


FIG. 2. Projection images of a human hand obtained without (left) and with (right) compensation for velocity-induced phase shifts as in Fig. 1.

pulsatile flow (10,16) in the absence of other motion between images.

The technique reported here differs from those above in that in addition to modifying the conventional imaging sequence to control the sensitivity to flow, taking the difference of two images (or two signals) acquired with different flow sensitivity permits canceling out stationary tissues and imaging only vessels. Computer simulations of a similar technique were presented in (17). Cardiac



FIG. 3. Difference image produced from images in Fig. 2 shows cancellation of stationary tissue, leaving blood vessels. (Eddy current effects produce slight registration artifacts at the ends of the field.)



FIG. 4. Projection image of head and thorax of rat obtained with same pulse sequence as in Fig. 3 (with smaller field of view). Note that both arteries and veins are seen, for example, the carotids (c) and jugulars (j).

gating is unnecessary and veins can be imaged as well as arteries. By acquiring the velocity-compensated and -uncompensated data in an interleaved manner, registration errors due to subject motion are minimized.

A potential problem with this technique arises from possible residual eddy currents from the magnetic field gradient pulses, which can lead to difficulty in fully subtracting stationary tissues. This accounts for the faint background image in the examples shown above.

Although a given set of velocity-uncompensated data will sensitize to components of velocity along only one direction, similar data can be acquired for two additional orthogonal directions and the results combined to sensitize to all velocities.

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