Projective Imaging of Pulsatile Flow with Magnetic Resonance

Abstract. *Noninvasive angiography with magnetic resonance is demonstrated. Signal arising in all structures except vessels that carry pulsatile flow is eliminated by means of velocity-dependent phase contrast, electrocardiographic gating, and image subtraction. Background structures become in effect transparent, enabling the three-dimensional vascular tree to be imaged by projection to a two-dimensional image plane. Image acquisition and processing are accomplished with entirely conventional two-dimensional Fourier transform magnetic resonance imaging techniques. When imaged at 0.6 tesla, vessels 1 to 2 millimeters in diameter are routinely detected in a 50-centimeter field of view with data acquisition times less than 15 minutes. Studies of normal and pathologic anatomy are illustrated in human subjects.*

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Vascular imaging is indispensable to modern medical care. It consists principally of Roengten angiography and Doppler ultrasonography. Both these techniques have well-known drawbacks (*1*). Arterial catheterization can result in hemorrhage or embolism, and contrast material can cause allergic reaction or tissue anoxia. Ultrasonography is ineffective at important locations, for example the cranium, and seldom provides adequate resolution for making surgical decisions.

Magnetic resonance (MR) imaging is a new, noninvasive imaging modality that is rapidly gaining clinical acceptance. Flow detection with MR spectroscopy has been explored for more than 30 years (*2*). MR imaging of the vascular system has extended these ideas, promising to reveal circulatory dynamics in a high-resolution image. Three approaches to MR flow imaging have developed. The time-of-flight approach systematically interprets proton partial saturation effects caused by flow perpendicular to the image plane (or planes) in conventional (or multislice) tomographic imaging (*3*). Fourier velocity zeugmatography (*4*) uses phase encoding (*5*) to resolve further the protons within each image element (voxel) according to their velocities. Modifications of this basic scheme have sought to abrogate the trade-off between velocity resolution and image acquisition times in physiologically realistic settings (*6*, *7*). Phase-contrast flow imaging combines features of time-of-flight and Fourier methods. Velocity-dependent phase shifts are used to produce modulations of image intensity from which many qualitative features of vascular flow may be deduced (*8*, *9*).

Here we describe projective imaging of pulsatile flow by phase contrast and subtraction (*10*). The prototype of projective imaging with contrast and subtraction is digital subtraction angiography (DSA). In DSA, a pair of identically registered transmission radiographs are recorded for a region before and after intravascular infusion of radiopaque contrast material. These images are then subtracted. Structures whose x-ray absorption is unaffected by the contrast material cancel in the subtraction process. The residue of the subtraction shows the precise distribution of the contrast material within the vascular tree. Our procedure closely imitates this strategy. Its fundamental difference is the replacement of exogenous pharmaceutical contrast by endogenous, physiologic, velocity-dependent phase-contrast and ionizing radiation by the MR signal.

Two projective MR images are acquired for a region. One is gated to cardiac systole and the other to diastole. The imaging pulse sequence was constructed to generate velocity-dependent phase shifts of approximately 1 cycle m^{-1} sec^{-1} (phase per velocity). The pulse sequence that accomplishes this is a conventional two-dimensional Fourier transform (2DFT) pulse sequence with a 9-msec echo time (t_e) (Fig. 1) (*7*, *11*). This choice of phase shift results in the relative preservation of blood proton signal at diastolic flow velocities (<0.1m sec^{-1}) and relative ablation of signal at systolic velocities (<0.5 to 1.5 m sec^{-1}). By this means, the systolic and diastolic image intensities differ in each location by the diastolic blood signal. The subtraction image (diastole minus systole) is the flow image, that is, a map of the pulsatile flow.

The mechanics of this contrast depends on the projective imaging format. In general, signal intensity is maximal

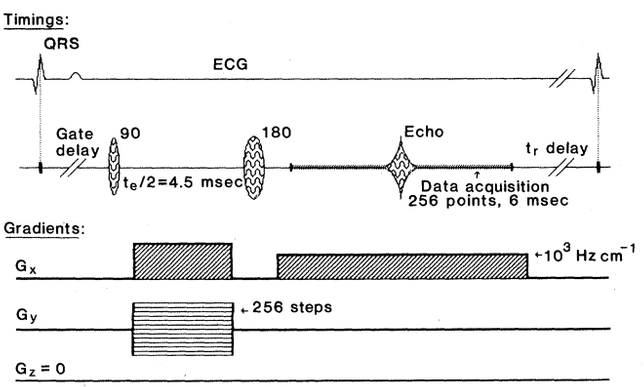

Fig. 1. The imaging pulse sequence, a two-dimensional Fourier transform spin-echo pulse sequence of conventional form. Acquisition is synchronized with the cardiac cycle by ECG gating: pulse sequence repetitions are triggered by the ECG QRS complex, with a delay appropriate for systole or diastole. Velocity-dependent phase shifts are caused by the x gradient pulses, which simultaneously perform spatial (frequency) encoding in x. The y gradient pulse performs spatial phase encoding and causes no velocity-dependent phase shifts: a proton moving in y will be associated with its mean y coordinate for the duration of the pulse. The z gradient is used neither for spatial encoding nor for spatial selection; therefore the experiment is projective along z to the x-y plane.

when the protons in each image element have a common phase (are coherent) and declines as the phases disperse (become incoherent). In each projective voxel, the proton signal has two constituents: a larger coherent part originating in the stationary background material, and a smaller part originating in the blood protons whose coherence depends on the flow velocity. At diastolic flow velocities (<0.1 m sec^{-1}), the imaging pulse sequence produces phase shifts of less than 0.1 cycle. In this case, the blood proton signal behaves coherently and adds to the background signal. In systole, peak velocities between 0.5 and 1.5 m sec^{-1} generate phase shifts of 0.5 to 1.5 cycles. Each projective voxel, however, intersects any vessel along a chord of points, so that all velocities between zero and the maximum velocity are sampled by partial volume averaging. This range of velocities creates signals that are incoherent and make little or no contribution to the net signal for the voxel.

Imaging was performed in a commercial 0.6-T (25-MHz, proton) superconducting imaging system (Technicare). Images resolved $(256)^2$ pixels over a $(50\text{-cm})^2$ field of view, which is approximately $(2\text{ mm})^2$ per pixel. The brief duration of the data acquisition window (6 msec, Fig. 1) necessitated the comparatively large readout x gradient of 0.25 G cm^{-1} (1 kHz cm^{-1}, proton) and the use of a broad-band time-domain signal filter (60 kHz). With two signal averages, each image acquisition requires 512 cardiac cycles, taking about 8 minutes in subjects with a normal pulse.

The systolic gate delay was tailored for each human subject (Figs. 2 through 5). Except in individuals with ventricular dysrhythmias, the electrocardiogram (ECG) QRS complex coincides with arterial diastole. However, the arrival time of peak systolic flow is variable. In normal subjects, arrival times increase predictably with distance from the heart. Disease processes such as aneurysm or occlusion may retard the pulse wave; others, such as nonocclusive atherosclerosis, may accelerate it. The systolic gate delay was selected empirically by performing between one and four brief (1-minute) low-resolution (64 by 256 pixels) "localization" images. Best results were obtained by exploring the probable range of gate delays in 50-msec increments.

The current experiments do not measure flow because of the nonlinear dependence of image contrast on proton motion. Our laboratory has demonstrated a different method of projective imaging that can measure the blood volume crossing each voxel (*9*). It proceeds from an analysis of the full structure of the image data as complex numbers and is based on the use of small phase shifts (<1 radian) to linearize the contrast (*4*).

On a cautionary note, it is important to bear in mind three processes that will augment contrast. First, the fraction of velocity that produces phase shifts and contrast varies as the cosine of the angle between the blood flow and the image x axis. Vessels oriented at large angles to x may be unobserved in the flow image because of an undiminished systolic signal. In theory, two flow images may be acquired, separated by a 90° rotation in the x-y plane. Each vessel will have a

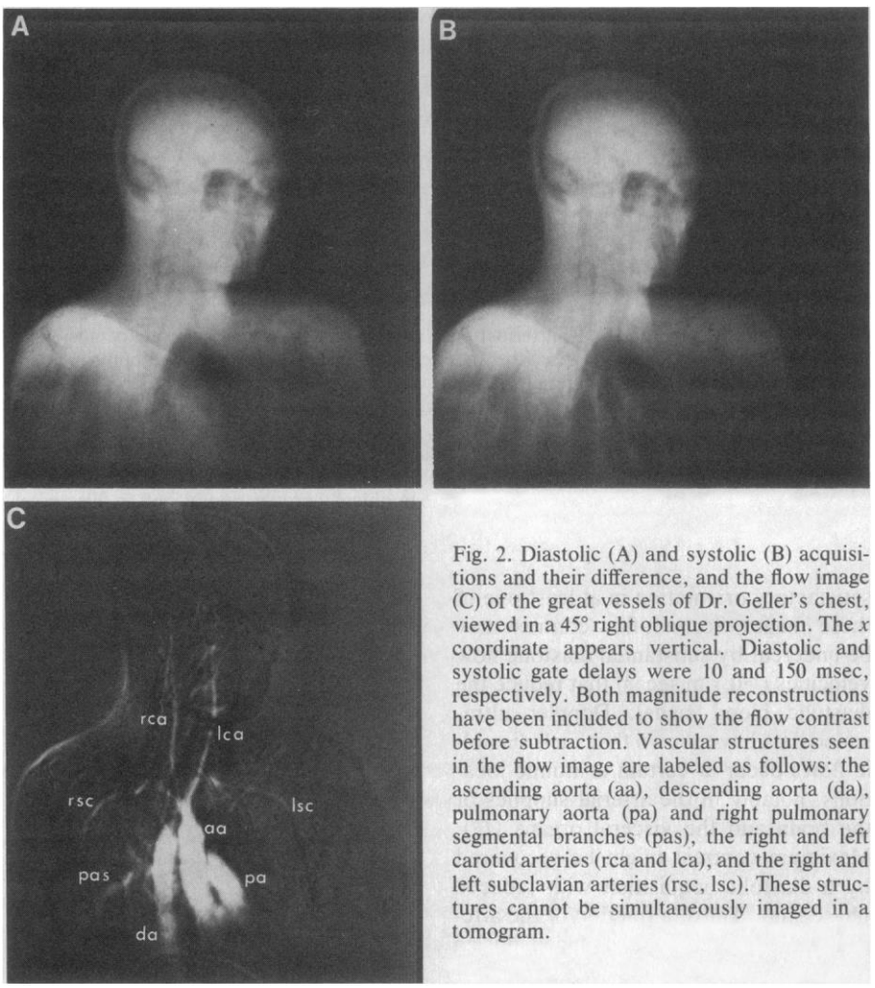

Fig. 2. Diastolic (A) and systolic (B) acquisitions and their difference, and the flow image (C) of the great vessels of Dr. Geller's chest, viewed in a 45° right oblique projection. The x coordinate appears vertical. Diastolic and systolic gate delays were 10 and 150 msec, respectively. Both magnitude reconstructions have been included to show the flow contrast before subtraction. Vascular structures seen in the flow image are labeled as follows: the ascending aorta (aa), descending aorta (da), pulmonary aorta (pa) and right pulmonary segmental branches (pas), the right and left carotid arteries (rca and lca), and the right and left subclavian arteries (rsc, lsc). These structures cannot be simultaneously imaged in a tomogram.

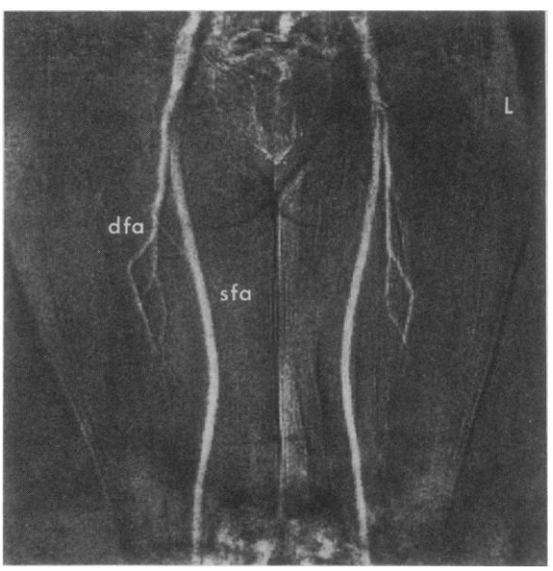

Fig. 3. Flow image of the thighs of a normal subject, projective to the coronal plane. The x coordinate is vertical. Diastolic and systolic gate delays were 10 and 250 msec in this individual (height, 180 cm, HR 76). Labeled arteries are the superficial femoral (sfa) and the deep femoral (dfa).

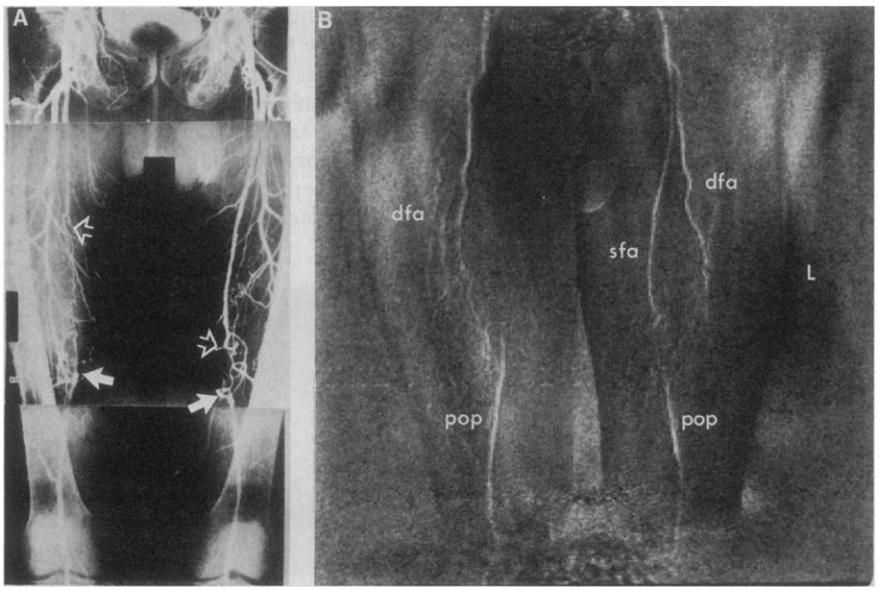

Fig. 4. Roentgen (A) and MR (B) arteriograms showing bilateral atherosclerotic occlusions of the superficial femoral arteries. The Roentgen technique used an intra-arterial bolus injection (75 ml) of iodinated contrast material. MR image acquisitions used gate delays of 10 and 300 msec. Arterial segments are marked as in Fig. 3. Proximal and distal points of occluded segments are marked by open and closed arrows, respectively. Visualization in the MR study of the popliteal arteries (pop) reconstituted by collateral flow implies that they are pulsatile, as confirmed by Doppler ultrasound examination. The poor visualization of the right proximal superficial femoral artery is consistent with angiographically proven poor runoff in this vessel.

satisfactory orientation in at least one of these images. Second, flow contrast may be undercut by substantial diastolic flow velocities (>0.1 m sec^{-1}) that reduce the diastolic blood signal. Doppler ultrasound data have shown that such diastolic flows occur at certain anatomic locations, notably in the arterial supplies of the brain and the visceral organs (*12*). Maximum contrast might be recovered by a diastolic acquisition with reduced phase shift. Third, a fraction of the arterial blood protons are replaced during each interpulse interval (t_r) by unsaturated protons formerly outside the radiofrequency antenna (proton refreshment). The saturated protons' signal is reduced relative to the unsaturated protons' signal by the factor $[1 - exp(-t_r/T_1)]$, where T_1 denotes the longitudinal relaxation time (in this case, in blood). Arterial segments subject to refreshment will have proportionally enhanced intensity in the flow image (*13*).

The efficiency of vascular imaging in the projective format is self-evident. Neither tomography nor existing three-dimensional imaging techniques can present as detailed anatomy from as large territories in so compact and accessible a form (*14*). Projective imaging is also technically advantageous. MR imaging times grow exponentially with the dimensions of the experiment. Although projective images are sensitive throughout a three-dimensional volume, their data acquisition and processing requirements are those of two-dimensional imaging. A related advantage is the reliance on imaging pulse sequences and reconstruction procedures of completely conventional types. From a clinical standpoint, the noninvasiveness of this method may allow its application outside the compass of current indications for angiography; for example, jeopardized arterial grafts could be given frequent follow-up assessment. Pulsatility, the source of contrast, reflects both anatomy and physiology. Further work must interpret these data in clinical and pathophysiological contexts.

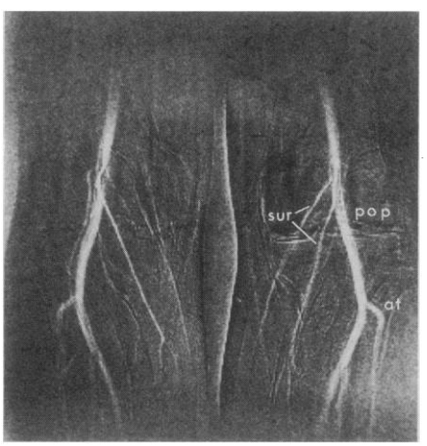

Fig. 5. Surface coil image of the vessels of the knees of a normal subject. Resolution is doubled relative to images above by doubling the gradient strengths in the imaging pulse sequence. This doubles the velocity-dependent phase shifts without conspicuous degradation of contrast. Signal is transmitted by a 50-cm saddle-shaped body radio-frequency coil and is received by a 20-cm square surface coil. Labeled arteries are the popliteal (pop), anterior tibial (at), and sural branches (sur). The diameters of the sural branches are about 1 mm.

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11. Moving protons experience a sequence of magnetic environments and therefore of Larmor frequencies unlike any static proton. These frequency differences accumulate as a phase shift. This phase shift is evident at the center of a Hahn spin echo, when the phases of static protons are restored to their common initial value. When a constant gradient G is applied, then the phase shift ϕ is given by

$$\phi = \frac{1}{4}\ \gamma \mathbf{G} \cdot \mathbf{V}\ t_e^2$$

where γ is the gyromagnetic ratio, V is the proton velocity, and t_e is the echo time [see (*2*)].
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13. In both systolic and diastolic gating, each interpulse interval represents exactly one cardiac cycle. Therefore, the set of arterial segments subject to proton refreshment is the same in both images. The refreshment-dependent modulation of image intensity is intact in each flow image because it "factors through" the image subtraction. Relative to image subtraction, proton refreshment resembles a static variable (such as proton density) more than a dynamic variable (such as phase contrast).
14. For an ingenious use of multislice MR imaging, see J. Hale *et al.*, *Proceedings of the 3rd Annual Meeting of the Society for Magnetic Resonance in Medicine* (Society for Magnetic Resonance in Medicine, Berkeley, 1984), p. 294.

15. This work was supported in part by the Technicare Corporation, Solon, Ohio, and by U.S. Public Health Service grant 1 KO4 CA00848-03, awarded by the National Cancer Institute, Department of Health and Human Services. We thank Drs. David Kramer and David August and Lauren Abbate for thoughtful suggestions and criticism, and Edith Bell for technical assistance.

17 June 1985; accepted 23 September 1985