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Technical Note

Non-Contrast-Enhanced MR Angiography Using 3D ECG-Synchronized Half-Fourier Fast Spin Echo

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A non-contrast-enhanced three-dimensional (3D) magnetic resonance angiography (MRA) technique, which acquires images in a reasonably short scanning time and requires no contrast agent, is developed. An electrocardiographically (ECG) synchronized 3D half-Fourier fast spin-echo (FSE) technique with an appropriate ECG delay time for every slice encoding in 3D terms was used to examine the thoracic and iliac regions in 16 healthy volunteers at both 0.5 and 1.5 T. Prior to each 3D fresh blood imaging (FBI) experiment, an ECG preparation (ECG-prep) scan was acquired, and an appropriate ECG triggering time was selected for 3D FBI acquisition to optimize visualization of the vessel of interest. In the thoracic and abdominal regions, good-quality 3D MRA images were obtained. Furthermore, the weighted subtraction of two images in different phases provides contrast enhancement between arteries and veins. J. Magn. Reson. Imaging 2000;12: 776–783. © 2000 Wiley-Liss, Inc.

Index terms: non-contrast-enhanced MR angiography; fresh blood imaging (FBI); T2 blurring; pulse sequence; ECG triggering; ECG-prep scan

FOR SLOW BLOOD FLOW in peripheral vessels, several magnetic resonance angiography (MRA) techniques have been reported, such as flow-independent MRA (1), fast spin echo (FSE) (2), 3D half-Fourier FSE (3), and fast maximally interleaved fluid attenuated inversion recovery (FLAIR) with fat suppression (4). As a flow imaging technique, phase contrast (PC) has the benefit of acquiring various flow speeds in arteries and veins using flow encoding. However, 3D PC suffers from a long acquisition time. For fast-flow vessels, 2D and 3D time-of-flight (TOF) techniques have been found to be very useful, especially in cranial MRA. In addition, the use of electrocardiographic (ECG) triggering in 3D TOF has been proposed to enhance the TOF effect and to reduce flow artifacts (5–7). However, when the region of interest is large, the acquisition time becomes quite long, because the slice direction must be perpendicular to the vessel orientation.

Recently, contrast-enhanced MRA techniques, which are rapidly gaining in clinical importance, have been employed to depict vessels with the administration of contrast agent (8). The benefits of the contrast-enhanced MRA technique are a faster acquisition time due to coronal in-plane acquisition, high signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR), flow independence, less sensitive to complex flow, and the ability to perform dynamic studies to differentiate the arterial and venous phases.

Most recently, a non-contrast-enhanced MRA technique, fresh blood imaging (FBI) using coronal in-plane 3D half-Fourier FSE synchronized with ECG gating at every slice encoding, has been developed to acquire images at 0.5 T (9). In this study, we have explored this technique in a variety of regions at both 0.5 and 1.5 T. In addition, this paper presents the detailed theory of the FBI technique and introduces further applications.

THEORY

In general, FSE-related sequences are known to provide "black blood" or flow void images. The FBI technique, non-contrast-enhanced 3D MRA, allows "bright blood" depiction using ECG synchronized half-Fourier FSE without the use of contrast agent.

First, the reduction in the echo train spacing (ETS) length in half-Fourier FSE reduces the single-shot acquisition time, which effectively freezes motion-related artifacts and minimizes susceptibility effects (10). As the ETS becomes shorter, the single-shot acquisition window of half-Fourier FSE becomes shorter. When half-Fourier reconstruction is applied, the single-shot acquisition time for a 256×256 matrix is less than 1 second.

Second, relatively short T2 components show characteristic blurring in the phase-encode direction in FSE-related sequences (11). Because the blood T2 values are relatively short, the signals of blood are blurred

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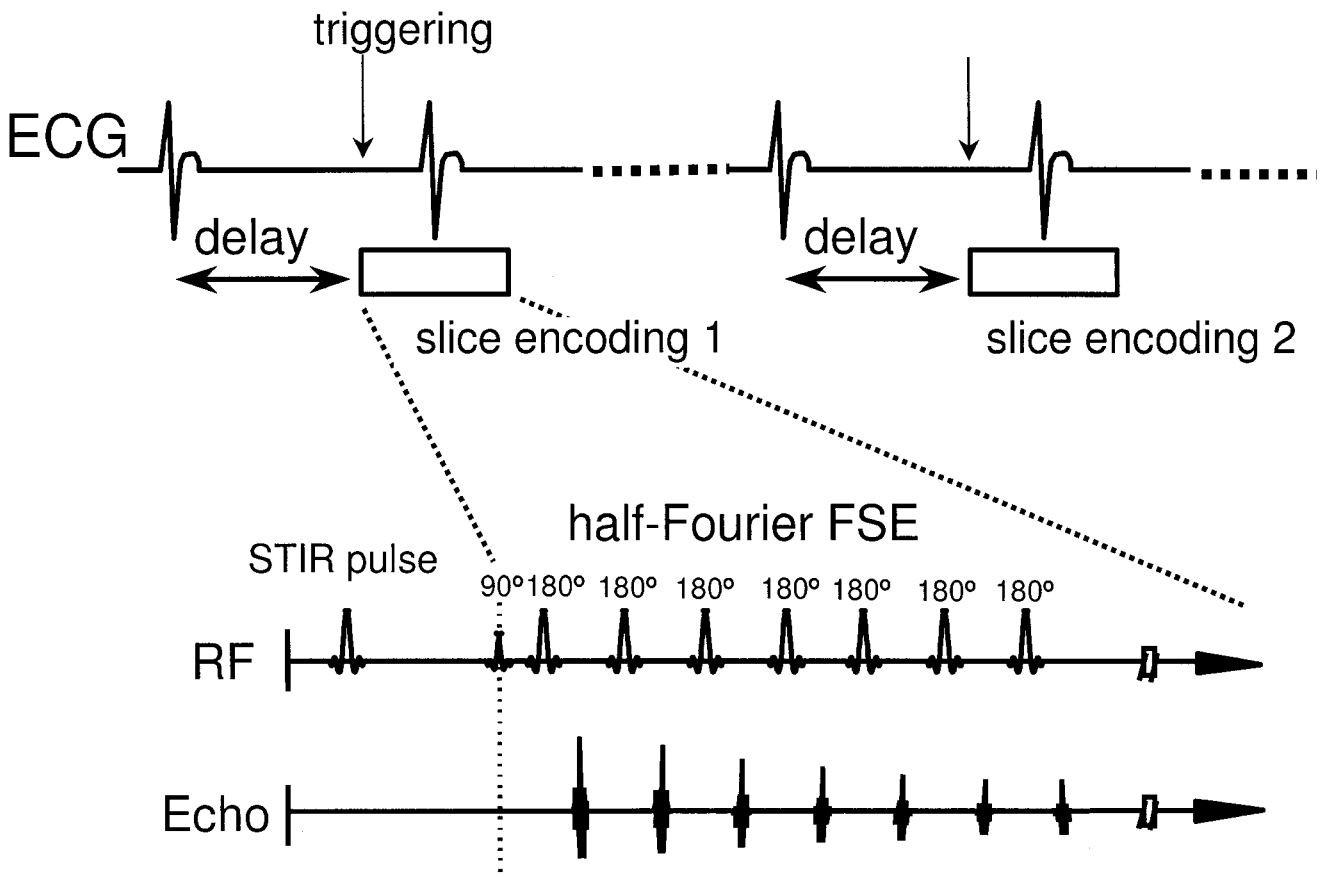


Figure 1. Sequence diagram of ECG-synchronized 3D half-Fourier FSE. The 3D half-Fourier FSE sequence is ECG synchronized for each slice encoding to have the same cardiac phase in every slice partition.

in the phase-encode direction in half-Fourier FSE (12). The blurring in the phase-encode direction means that blood signals are spread over several pixels, or the signal amplitude is reduced due to an increasing full-width at half-maximum (FWHM) in the point spread function (11). However, when the phase-encode direction is placed in the orientation of the blood vessels, signal enhancement is obtained from the overlapped T2 signal blurring between the neighboring pixels, contributing to the "bright blood" signal.

Third, it has been reported that central k-space ordering reduces flow voids in the phase-encode direction in FSE-related sequences (13). Because acquisition is from near the center or low frequencies of the k-space in half-Fourier FSE, less flow dephasing, due to the smaller gradient amplitude, is obtained in the phase-encode direction compared with the read-out direction.

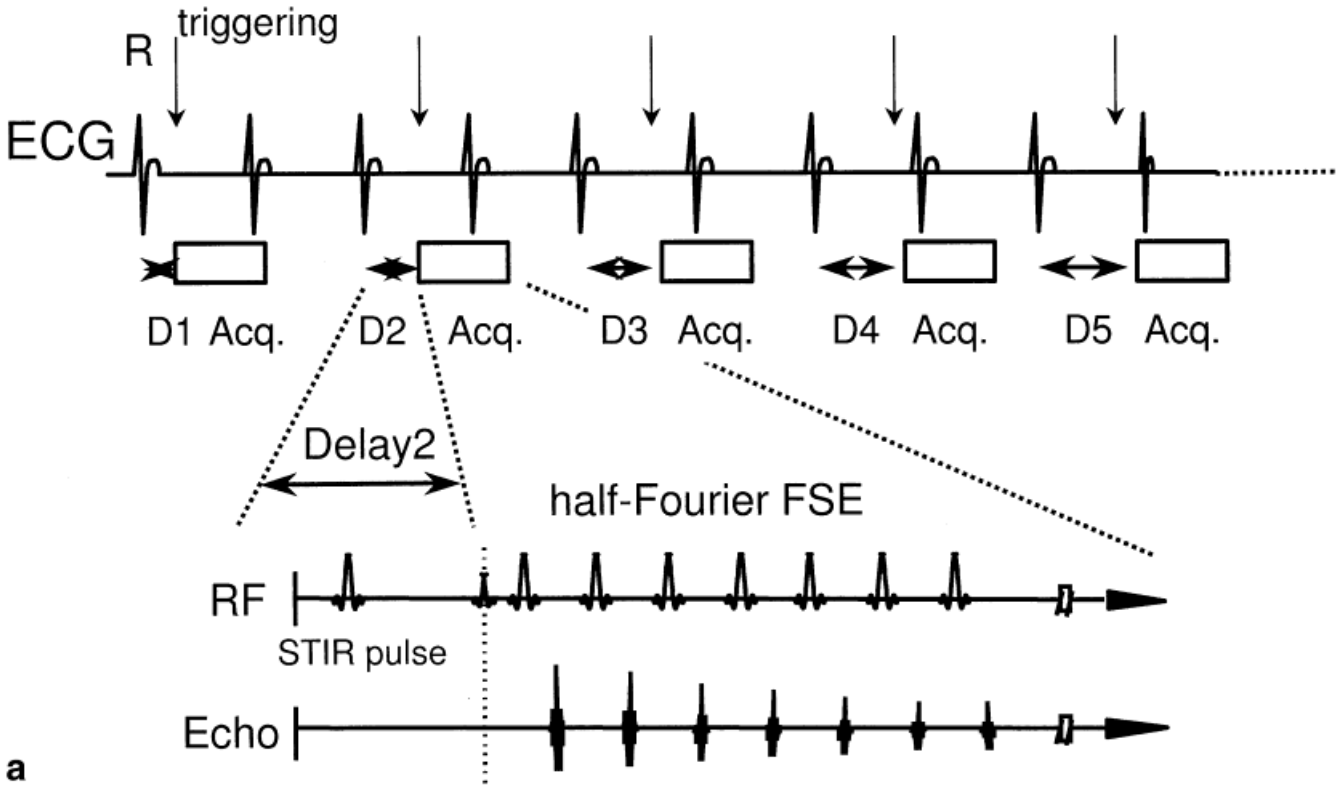
Now, by orienting the phase-encode direction parallel to the vessel of interest, one can take advantage of both signal enhancement of overlapped T2 signal blurring and reduced flow dephasing for fast flow. Taking these points into consideration, the FBI technique using 3D half-Fourier FSE synchronized with ECG gating for each slice encoding has been developed to allow depiction of fast-flow vessels of interest.

Figure 1 shows a sequence diagram of ECG synchronized 3D half-Fourier FSE. A spatially nonselective short-time inversion recovery (STIR) pulse is applied for fat suppression and background signal attenuation. For 3D half-Fourier FSE acquisition, a low-pass phase-encode ordering or a low-frequency sampling was employed with half-Fourier reconstruction to acquire images in a single-shot scan, which was consecutively repeated in the slice-encode direction with encoding gradients, as shown in Fig. 1 (3,9,14). However, fast-flow vessels, such as the ascending and descending aorta, are strongly affected by the cardiac cycle, which effectively changes the speed of flow in different phases of the cardiac period. To determine the appropriate cardiac phase or ECG delay time, an ECG preparation scan (ECG-prep scan), which acquires single slices in multiple phases, has been developed.

MATERIALS AND METHODS

ECG-Prep Scan

Figure 2a shows a schematic diagram of the ECG-prep scan. Various delay times are ECG triggered, followed by 2D half-Fourier FSE to obtain single-slice images in different phases. An appropriate ECG delay time is determined for the vessel of interest from the acquired images, and the selected delay time is later applied in the 3D half-Fourier FSE acquisition synchronized by ECG gating for every slice encoding. For a typical acquisition, the ECG-prep scan acquires several single



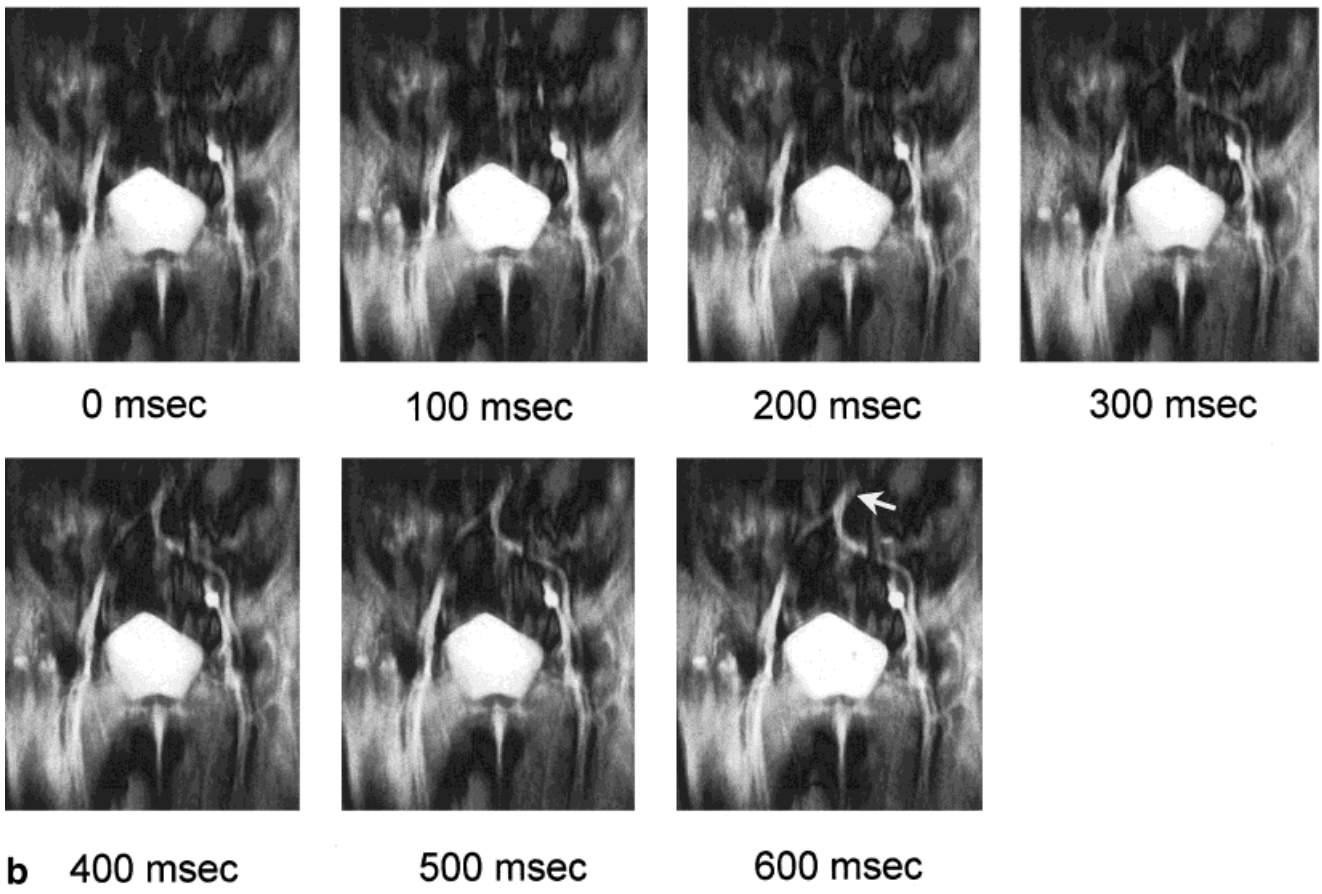


Figure 2. a: Schematic diagram of ECG-prep, single slices in multiple phases. Single-shot images at the same slice position are acquired to determine the appropriate ECG delay time for the vessel of interest. **b:** Typical results of ECG-prep scan acquired in the iliac region at 0.5 T. Each shot takes about 420 msec using 2D half-Fourier FSE [128 (phase-encode) × 256 (read-out) matrix] with various ECG delays as indicated. Note that the descending aorta (arrow) shows "bright blood" or high signal intensity during the diastolic phase but shows low signal or flow voids during the systolic phase. In contrast, the inferior vena cava (IVC) shows relatively high signal intensity throughout the cardiac cycle.

slices at 100-msec intervals starting from zero delay from the R wave. Therefore, the images are acquired with phases of 0, 100, 200, 300, and 400 msec, and so on. Figure 2b shows the results of a typical ECG-prep scan for the iliac region.

Imaging Protocol

All FBI studies were performed in 16 healthy volunteers (12 men and 4 women; age range 21–45 years) using a 0.5-T (FLEXART, Toshiba, Tokyo, Japan) or a 1.5-T (VISART/EX, Toshiba) clinical imager. Nine subjects (five for 1.5 T and four for 0.5 T) for the thoracic region and seven subjects (four for 1.5 T and three for 0.5 T) for the iliac region were examined. The institutional review boards approved the study, and informed consent was obtained from all subjects. Prior to FBI studies, ECG-prep scans were acquired using 2D half-Fourier FSE, as shown in Fig. 2a. At 0.5 T, ECG-prep scans were performed using the following parameters: TR/TE_{eff} 3 R-R intervals/24.8 or 62.0 msec, ETS 6.2 msec, matrix 128 × 256, TI 140 msec, number of acquisitions (NAQ) 1, slice thickness 40 mm, and field of view (FOV) 38 × 38 cm, for a total acquisition time of about 20 seconds. Similar parameters were used at 1.5 T, with ETS 5.0 msec, TE_{eff} 30 or 60 msec, and TI 190 msec. The single-shot acquisition window for a 128 × 256 (phase-encode, read-out) matrix was about 420 msec with an ETS of 6.2 msec and TE_{eff} of 24.8 msec at 0.5 T (10 mT/m gradient strength) and about 350 msec with an ETS of 5.0 msec and TE_{eff} of 30 msec at 1.5 T (25 mT/m gradient strength).

After the ECG-prep scan, the appropriate delay time was selected and applied in 3D FBI acquisition to trigger every slice encoding in a 3D manner, as shown in Fig. 1. The FBI studies used the same TE_{eff} value as that for the ECG-prep scan to obtain similar contrast. Typical parameters for 3D FBI studies at 0.5 T were as follows: TR/TE_{eff} 3 R-R intervals/24.8 or 62.0 msec, 8–20 slice partitions with a slice thickness of 2 –4 mm, ETS 6.2 msec, matrix 256 × 256, FOV 38 × 38 cm, and TI 140 msec, for a total acquisition time of about 80 seconds to 4 minutes. At 1.5 T, similar parameters were employed, with ETS 5.0 msec, TE_{eff} 30 or 60 msec, and TI 190 msec. To suppress fatty tissue and to attenuate background signals, an STIR pulse was applied, with no field inhomogeneity adjustment or shimming required. For the vessels of the thoracic arch, a relatively short TE_{eff} (24.8 msec for 0.5 T and 30 msec for 1.5 T) was applied. For the iliac region, a TE_{eff} of 62 msec at 0.5 T or 60 msec at 1.5 T was used in consideration of the contrast between vessels and background. Except for the iliac region, an intermittent breath-hold technique using acoustic sound from gradient switching for patient feedback (14,15) or respiratory gating with an external belt was employed. Interpolation in the slice direction was performed to improve the apparent resolution, and the 3D FBI data were processed with maximum intensity processing (MIP). To investigate the relationship between the flow speed and the flow signal, PC and FBI at various ECG delay times were acquired in the thoracic region.

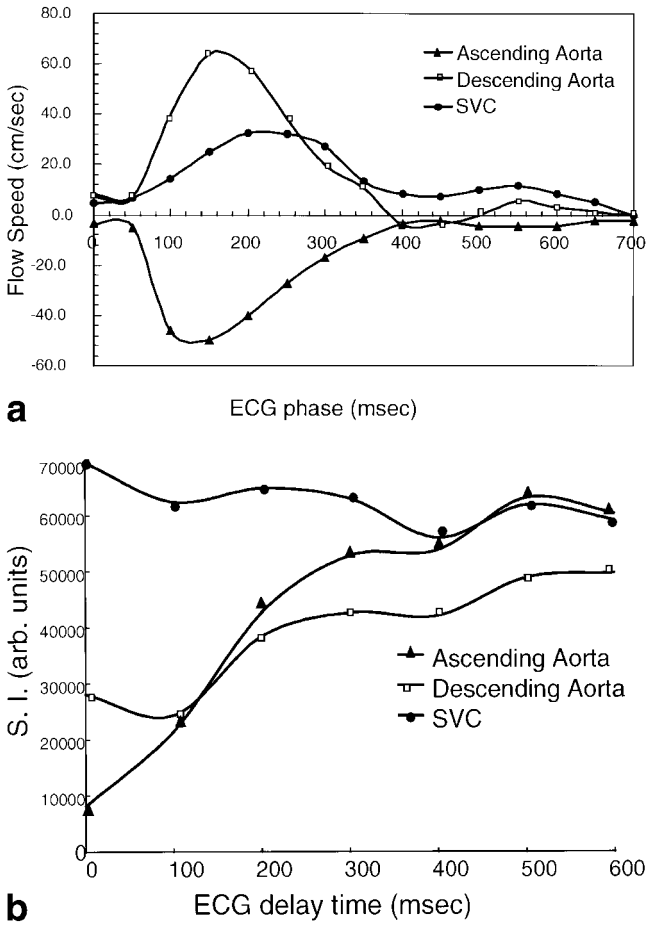


Figure 3. Relationship between flow speed and signal intensity at 1.5 T. **a:** Flow speeds in the ascending aorta, descending aorta, and superior vena cava (SVC) measured using PC. **b:** Signal intensities of the vessels measured using 3D half-Fourier FSE triggered at various ECG delays. When the flow speed is fast during systole, low signal or flow voids are observed. However, the descending aorta and ascending aorta, showing steady moderate flow during diastole, are depicted as "bright blood."

RESULTS

Figure 2b shows the results of a typical ECG-prep scan for the iliac region. Various ECG delay times provide images in different phases with different blood signal intensity. The early systolic phases (0–200 msec) show low signal intensity or "black blood" for the descending aorta, while the diastolic phases (400–600 msec) show higher signal intensity. Thus, the ECG triggering time is an important factor influencing the blood signal intensity in the vessel of interest. In all 16 consecutive subjects, a similar image quality was obtained in terms of the delineation of aortic arch, iliac, and pulmonary vessels.

Figure 3a shows the results for flow speed in the ascending and descending aorta and superior vena cava (SVC) measured using PC in various phases of the cardiac cycle, and Fig. 3b shows the signal intensities measured at different ECG delay times using 3D FBI. The ascending aorta and the descending aorta exhibit fast flow speeds in opposite directions during systole

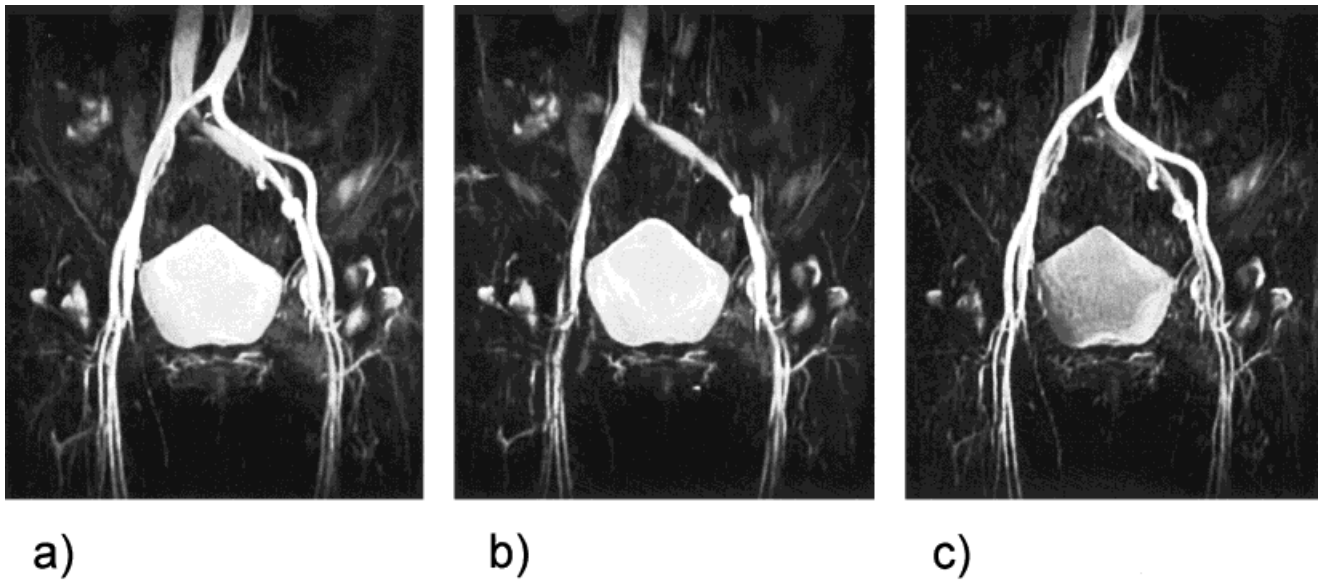


Figure 4. Typical weighted subtraction of the source images of 3D data acquired during the diastolic and systolic phases. The iliac images were acquired in the superior-inferior phase-encode direction using 3D half-Fourier FSE with a TR of 3 R-R intervals, a single shot per slice encoding, a TE_{eff} of 62 msec, and an ETS of 6.2 msec at 0.5 T. **a:** MIP image acquired with an ECG delay of 600 msec, showing both the descending aorta and inferior vena cava (IVC). **b:** MIP image acquired with an ECG delay of 100 msec, showing only the IVC with high signal intensity. **c:** MIP image of weighted subtraction of full intensity of image (**a**) minus half intensity of image (**b**) [1.0 (a) − 0.5 (b)], of the source images obtained with ECG delays of 600 msec and 100 msec, resulting in enhancement of contrast between arteries and veins.

and relatively steady slow flow during diastole, while the SVC shows moderate flow speed throughout the cardiac cycle. Note the correlation between the signal intensity in the vessels during various phases measured by FBI and the flow speed by PC. The signal intensity in the ascending aorta and the descending aorta is lower during systole, when the flow speed is fast (Fig. 3a); however, the signal intensity is high in both regions during diastole, when the flow speed is slow. On the other hand, the SVC shows a steady high signal during all phases due to the moderate flow speed throughout the cardiac cycle. The flow speed measured by PC agreed well with the FBI results.

Figure 4a and b shows MIP images of the iliac region acquired during diastole and systole, respectively. The diastolic phase (Fig. 4a) shows both the descending aorta and the inferior vena cava (IVC), while the systolic phase (Fig. 4b) shows only the IVC, with the descending aorta depicted as "black blood." Weighted subtraction of full intensity of image (a) minus half intensity of image (b), [1.0 (a) − 0.5 (b)], of the source images followed by MIP processing enhanced the contrast between the artery and vein, as shown in Fig. 4c.

Figure 5a shows the source image, and Fig. 5b shows the MIP image for the upper thoracic region, acquired at 0.5 T during the diastolic phase with a total acquisition time of about 80 seconds without breath-hold. The superior-to-inferior phase-encode direction is applied to depict the carotid arteries running in the superior-inferior direction. The source image (Fig. 5a) depicts the ascending aorta to the common carotid artery as well as the SVC to the internal jugular vein. The upper thoracic vasculature is clearly visualized even at field strength of 0.5 T.

For the fast flow in the aortic arch, two-shot acquisition was used so that each single-shot scan time could be reduced by almost half, although the total scan time became twice as long. Figure 6a shows the source image, and Fig. 6b shows the MIP image of the aortic arch, acquired using two-shot acquisition with the superior-inferior phase-encode direction during diastole on a 1.5-T imager. The total acquisition time was about 3 minutes using the intermittent breath-hold technique.

The pulmonary vessels were examined at 1.5 T, as shown in Fig. 7. The right-left phase-encode direction was applied to depict the pulmonary arteries and veins, and MIP processing was used. The image was acquired using ECG gating during diastole and respiratory gating at expiration with a total acquisition time of about 4 minutes. The pulmonary vasculature is clearly visualized, even out to the small branches, without the use of contrast agent.

DISCUSSION

The following characteristic features contribute greatly to the depiction of blood vessels: short ETS half-Fourier FSE to reduce motion effects, short-T2 blurring effect in the phase-encode direction, and acquisition from near the center or low frequencies of the k-space to compensate for flow dephasing over the read-out direction (12). Thus, the proper selection of the phase-encode direction in half-Fourier FSE provides two benefits: signal enhancement from overlapped T2 blurring signals and reduced dephasing due to acquisition from near the center of the k-space.

From the ECG-prep scan results (Fig. 2b), various ECG triggering times provide images in different phases

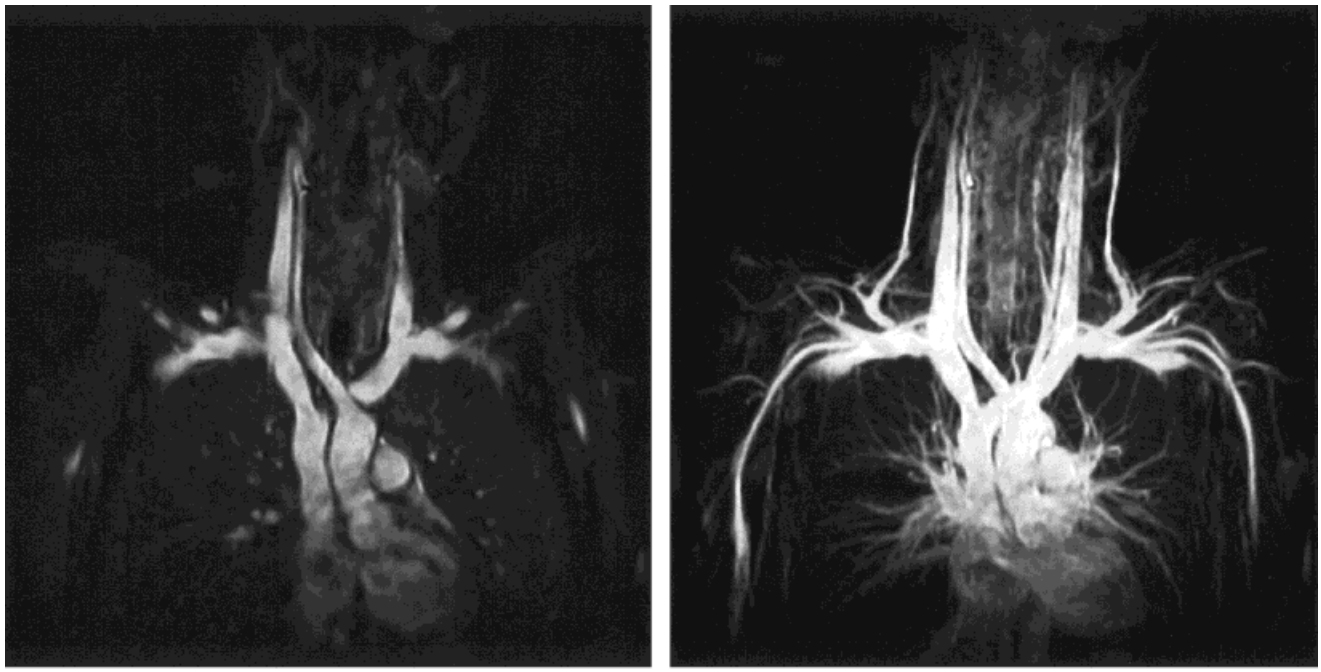


Figure 5. Thoracic region acquired in the superior-inferior phase-encode direction using 3D half-Fourier FSE with an ECG delay of 400 msec at 0.5 T. The acquisition parameters were a TR of 3 R-R intervals, a single shot per slice encoding, a TE_{eff} of 24.8 msec, an ETS of 6.2 msec, a TI of 140 msec, a 256 × 256 matrix, an FOV of 38 × 38 cm, and 8 partitions with a 4-mm slice thickness, for a total acquisition time of about 80 seconds. Finally, interpolation was applied in the slice direction to obtain 16 2-mm-slice source images. Image (**a**) shows the source image of the aortic arch, and image (**b**) shows the MIP image.

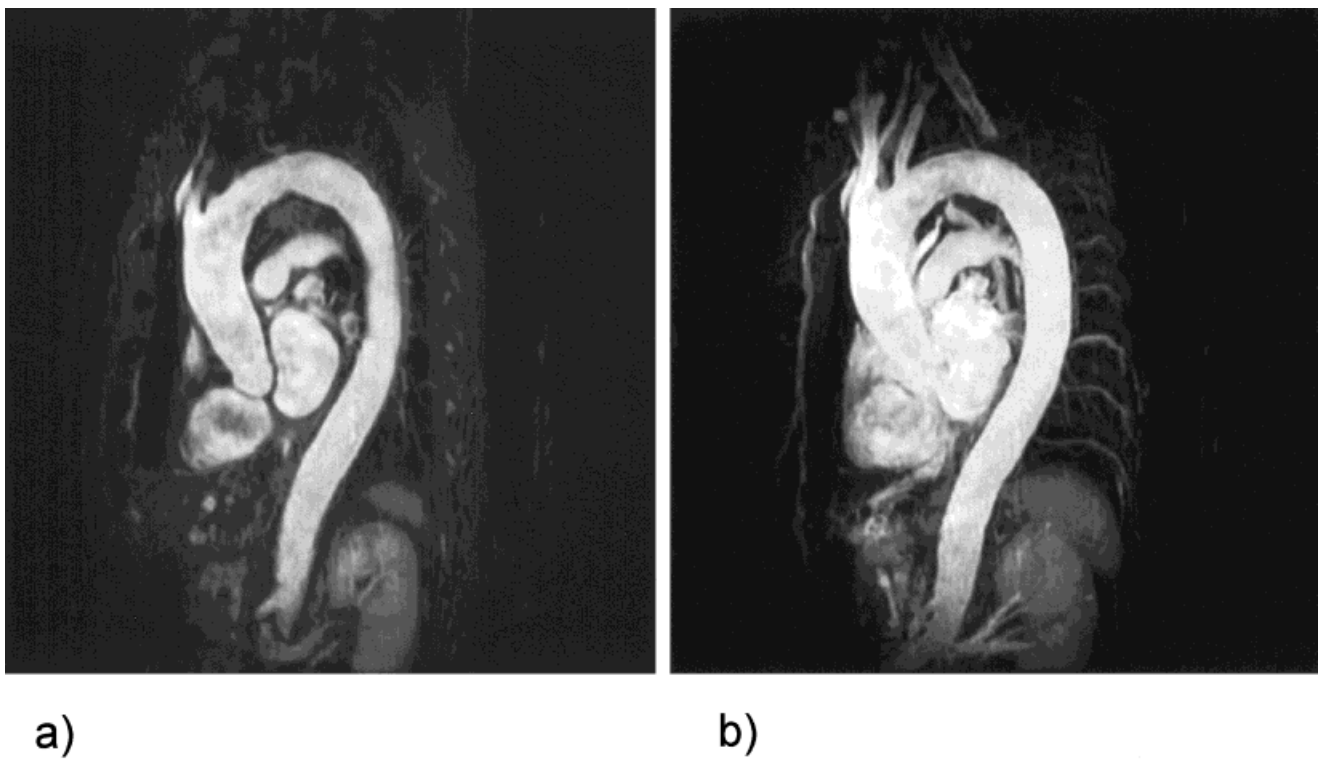


Figure 6. Aortic arch images showing a source image (**a**) and a MIP image (**b**) acquired in the superior-inferior phase-encode direction at 1.5 T using a TR of 3 R-R intervals, 2 shots per slice encoding, a TE_{eff} of 30 msec, an ETS of 5.0 msec, a TI of 190 msec, a matrix of 256 × 256, an FOV of 35 × 35 cm, and 8 partitions with a 3.0-mm slice thickness, for a total acquisition time of about 3 minutes using the intermittent breath-holding technique.

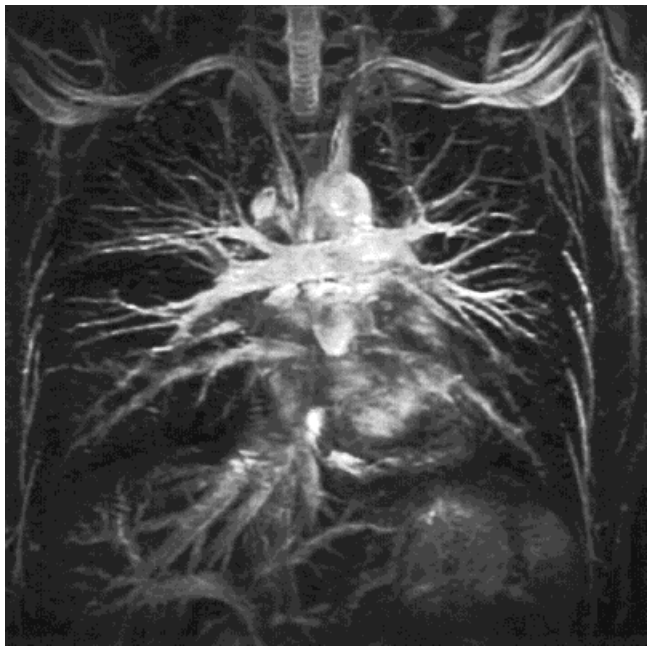


Figure 7. Pulmonary MIP image acquired in the right-left phase-encode direction at 1.5 T using a TR of 4 R-R intervals, a single shot per slice encoding, a TE_{eff} of 60 msec, an ETS of 5.0 msec, a TI of 190 msec, a matrix of 256 × 256, an FOV of 35 × 35 cm, and 16 partitions with a 3-mm slice thickness, for a total acquisition time of about 4 minutes using respiratory gating at expiration. After measuring the ECG delays for the pulmonary vessels using the ECG-prep scan, a delay of 500 msec was applied in 3D acquisition.

with different blood signal intensity. Therefore, it is important to perceive the appropriate triggering time for the vessel of interest preceding the 3D FBI scan.

The results of the flow speed study using PC at various ECG delay times (Fig. 3a) show good correlation with the results of FBI studies of the thoracic vessels (Fig. 3b). Low signal or flow voids are observed when the flow speed in the vessel may be too fast for the ETS of 5 msec in half-Fourier FSE; however, even fast-flow vessels such as the ascending aorta and the descending aorta show relatively slow and steady flow during diastole. This implies that by triggering during the period of slow flow, it is possible to depict such vessels as "bright blood" with half-Fourier FSE. In addition, because the veins are depicted during all ECG phases due to the moderate flow speeds in these vessels despite a delay, weighted subtraction of source images acquired during two different phases allows contrast differentiation of arteries from veins. Therefore, obtaining images at different phases using the ECG-prep scan is important, especially for fast-flow vessels. Note the high signal intensity in the vessel during the late diastolic phase in Fig. 3b. The acquisition window of a single-shot 256 × 256 matrix is about 670 msec, which means that acquisition is performed over an R-R interval depending on the heart rate. However, the near-center or low frequencies of the k-space are sampled in late diastole, reducing flow artifacts.

The FBI technique allows coronal in-plane 3D acquisition and thus enables shorter 3D acquisition. Since the FBI technique involves 3D acquisition synchronized with the ECG delay at every slice encoding, it is possible to perform further processing such as volume rendering and stereo display. An STIR pulse is applied to suppress fatty tissue signals and to attenuate background signals, which does not require field inhomogeneity adjustment or shimming. The contrast characteristics of FBI images make it possible to depict fast-flow vessels during the slow-flow phase, slow-flow vessels, and long-T2 components or structures such as cerebrospinal fluid (CSF), bile, urine, and the thoracic duct (15). Furthermore, applying two-shot acquisition with a reduced acquisition window provides clearer delineation of fast-flow aortic arch vessels (Fig. 6) with reduced flow artifacts, even though the orientation of the arch is not constant. For two-shot acquisition, the k-space is filled with echoes from each shot in an interleaved manner.

The single-shot acquisition window, for a 256 × 256 matrix with a TE_{eff} of 60 msec and an ETS of 5 msec, is 60 msec (TE_{eff} time) to the k-space center and 5.0 msec × 128 (one-half of the 256 phase-encoding steps), for a total of 700 msec. Thus, the total of 140 phase-encoding steps is acquired, which is 128 steps plus 12 steps (TE_{eff} divided by ETS) at the low frequency part of the other half k-space. However, using the two-shot acquisition method with a 256 × 256 matrix, each single shot takes the same 60 msec (TE_{eff}) to the k-space center and 5.0 msec × 64 (one-quarter of the 256 phase-encoding steps), for a total of 380 msec. As each single-shot scan time is reduced by almost half, an acquisition window time of imaging becomes faster. Therefore, as the acquisition window time is reduced, clearer depiction can be achieved for fast-flow vessels. However, the depiction is limited to the phase-encode direction (12). In addition, the differentiation of aortic from venous vessel becomes difficult with slower pulsatile flow, where the flow speed is not dependent on the cardiac phase.

Now, let us summarize the FBI technique by comparing with other non-contrast-enhanced MRA techniques. The FBI technique allows coronal in-plane 3D acquisition, which is not possible with TOF, especially for body MRA, and thus enables shorter 3D acquisition. Compared with ECG-gated TOF or field-echo (FE) type sequences, FBI using a single-shot half-Fourier FSE collects many echoes in a single ECG-gated acquisition to fill up the k-space or to give a single image, which is not the case for ECG-gated TOF or FE that requires multiple ECG gating (5–7). Furthermore, a single stable STIR pulse is applied in FBI, while TOF or FE-type sequences use multiple chemical-shift selective fat suppression pulses per image, which is quite susceptible to field inhomogeneity (6).

In conclusion, the FBI technique is a fast 3D method allowing coronal in-plane acquisition to depict fast-flow vessels without the administration of contrast agent. The merits of the FBI technique are ease of acquisition, no contrast agent requirements, a short acquisition time, high resolution in the in-plane and slice-encode directions, and freedom to repeat the study immediately, unlike contrast-enhanced MRA. This technique shows great promise in many regions at both 0.5 and 1.5 T; however, it may be unsuitable for subjects with

arrhythmia. Further clinical applications and limitations of this technique are discussed elsewhere (16).

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