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Nonenhanced Methods for Lower-Extremity MRA: A Phantom Study Examining the Effects of Stenosis and Pathologic Flow Waveforms at 1.5T

Erik J. Offerman, BA^{1,*}, Philip A. Hodnett, MD^{1,2}, Robert R. Edelman, MD^{1,3}, and Ioannis Koktzoglou, PhD^{1,3}

¹Department of Radiology, NorthShore University HealthSystem, Evanston, Illinois, USA

²Department of Radiology, Northwestern University Feinberg School of Medicine, Chicago, Illinois, USA

³Northwestern University Feinberg School of Medicine, Chicago, Illinois, USA

Abstract

Purpose—To evaluate the signal properties of 2D time of flight (TOF), quiescent-interval single-shot (QISS), ECG-gated 3D fast spin-echo (FBI), and ungated 3D fast spin-echo ghost (Ghost) magnetic resonance angiography (MRA) over a range of flow velocities in a pulsatile flow phantom with a 50% diameter stenosis at 1.5T.

Materials and Methods—Blood-mimicking fluid was pumped at eight peak flow velocities through a stenotic region in triphasic and monophasic waveforms. Vascular signal proximal, within, and distal to the stenosis was measured from the source images of the four MRA methods. Coronal maximum intensity projection images were used to compare image quality.

Results—TOF and QISS signal trends were similar, but QISS exhibited the most consistent signal across velocities. At high velocities (≥ 42.4 cm/s), TOF showed post-stenotic signal loss that was not observed with QISS. FBI and Ghost signals peaked at low velocities (3.9–9.7 cm/s) without flow compensation and at high velocities (≥ 64.6 cm/s) with flow compensation.

Conclusion—FBI and Ghost demonstrated dependence on blood flow velocity and flow compensation. TOF was sensitive to flow artifacts at high velocities. QISS proved most robust for accurately depicting the normal lumen and stenosis under a wide range of flow conditions. Monophasic and triphasic flow did not appreciably affect the signal performance of any method.

Keywords

time of flight; fresh blood imaging; ghost; quiescent-interval single-shot; pulsatile flow phantom; magnetic resonance angiography

Peripheral arterial disease (PAD) has increased prevalence in older populations, smokers, and diabetics, and is associated with systemic atherosclerosis and cardiovascular morbidity (1–4). The ankle-brachial index (ABI) is a useful screening test, but is less reliable in diabetic patients who exhibit arterial calcification (5,6). Moreover, it is generally unable to determine the precise location and severity of vascular stenosis, so that an additional imaging test is needed prior to revascularization. Additional imaging tests include contrast-

enhanced magnetic resonance angiography (CE-MRA), which has proven to be accurate and avoids exposure to ionizing radiation and potentially nephrotoxic iodinated contrast agents (7).

Renal insufficiency is common in patients with PAD (8,9). This poses a problem for CE-MRA since it requires injection of a gadolinium-based contrast agent (Gd), which has been linked to nephrogenic systemic fibrosis (NSF) in patients with renal failure (10,11). The continued clinical demand for MRA in PAD evaluation requires scanning techniques that are safe for renally impaired patients. This has led to investigation into nonenhanced MRA techniques that do not require the use of contrast agents (12).

Several nonenhanced MRA techniques for evaluating the lower extremities have been proposed. Time of flight (TOF) is a method that relies on inflow enhancement (13). However, TOF is slow and requires rapid flow for optimal performance. Electrocardiogram (ECG)-gated 3D partial Fourier fast spin-echo (also termed “fresh blood imaging”) (FBI) is a technique that subtracts a “bright-blood” diastolic acquisition from a “black-blood” systolic acquisition, creating an angiogram with high vessel contrast and little background signal (14). However, FBI requires preparatory scans to define the acquisition windows and optimal spoiler gradient strengths, which can lead to operator variability and prolong study time. Ungated ghost MRA (Ghost) is a technique that uses ghost artifacts in the slice-encoding direction to produce an angiogram (15). Ghost does not require ECG gating, but is limited in that it can only produce a 2D angiogram. A more recent development is quiescent-interval single-shot (QISS) (16), which consists of a balanced steady-state free precession (bSSFP) readout applied during slow diastolic flow. The method uses presaturation to suppress background tissue and venous signal while arterial signal is enhanced by inflow during a quiescent interval that overlaps the period of fast arterial flow.

Little is known about the performance characteristics of these nonenhanced MRA techniques. Given that a study of all techniques mentioned above is not practical in human subjects due to parametric and scan time limitations, we sought to determine the performance characteristics of the TOF, FBI, Ghost, and QISS MRA methods by measuring signal versus velocity using a pulsatile flow phantom. Triphasic and monophasic flow profiles, blood-mimicking fluid, and a tapered stenosis were used to simulate normal and diseased conditions found in vivo.

MATERIALS AND METHODS

Phantom Setup

The phantom setup consisted of a programmable flow pump (CompuFlow 1000MR, Shelly Medical Technologies, Canada) that was used to drive a blood-mimicking fluid through a tubular phantom (inner diameter = 1/4 in.). The blood-mimicking fluid was a solution of 40% glycerol and 60% water (by volume) designed to mimic the viscosity and MR properties of human blood ($T_1 = 850$ msec, $T_2 = 170$ msec) (17). A stenotic region was created by inserting a plastic connector (Delrin, Dupont, Wilmington, DE) in the flow circuit. The connector was manufactured to have a 50% diameter stenosis (75% area stenosis) with respect to the tube and a gradual, tapered onset designed to mimic stenotic conditions in vivo (Fig. 1). The tubular phantom was submerged in a Gd-doped ($T_1 \sim 980$ msec) water bath to reduce field inhomogeneity and mimic the signal from background tissue.

MRI

MRI was performed on a 32-channel 1.5T scanner (Avanto, Siemens Healthcare, Erlangen, Germany) using a spine array coil and a body array coil that was strapped over the bath

containing the vascular phantom. Radiofrequency (RF) excitation was performed using the scanner body coil.

Images from the four MR techniques were acquired separately 30 cm proximal to and at the location of the 50% tapered stenosis while the pump generated a waveform that mimicked the triphasic flow pattern found in the femoral artery of a healthy individual (Fig. 2). Images were also acquired at a region of no stenosis using a monophasic flow pattern commonly found distal to arterial stenosis in patients with PAD (Fig. 3). This monophasic flow pattern was modeled after Doppler ultrasonographic measurements of blood flow distal to severe stenosis (18) and exhibited the tardusparvus phenomenon where the flow velocity in the systolic peak and backflow regions is reduced and the duration of the systolic peak is lengthened with a delayed onset (19). Scans were repeated three times each at eight different peak flow velocities with a simulated cardiac period of 820 msec. Eight peak flow velocities were used to simulate the wide range of velocities found in the peripheral vasculature, ranging from fast flow (40.7 ± 10.9 cm/s) in the larger femoral arteries to slow flow (16.8 ± 5.7 cm/s) found in the dorsalis pedis artery (18) and reduced flow (<10 cm/s) found distal to lesions in patients with PAD. Prior to acquisition, each flow profile was determined using a phase contrast acquisition and the peak flow velocity was measured as the maximum velocity at systole. Parameters for the phase contrast acquisition were as follows: field of view 393×135 mm; matrix 512×512 ; receiver bandwidth 391 Hz/Pixel; flip angle 30° ; TR 44.8 msec; TE 7.8 msec; acquisition time 3 minutes and 21 seconds. Velocity encoding was adjusted to optimize signal to noise ratio for each waveform. The nonstenotic data were acquired 30 cm proximal to the stenosis, therefore avoiding pulse wave reflection and turbulence-induced signal loss caused by the stenosis. The imaging parameters for the four methods are summarized in Table 1. FBI and Ghost utilized fast spin-echo readouts with similar echo spacings and train lengths, while QISS was acquired with a bSSFP readout.

Optimization of Parameters for Fast Spin-Echo Based Methods

Multiple preliminary scans were performed to optimize key pulse sequence parameters for FBI and Ghost. The flip angle and echo time were varied for each acquisition under triphasic flow waveforms with four peak flow velocities (3.9 cm/s, 21 cm/s, 42.4 cm/s, and 83 cm/s). Coronal maximum intensity projection (MIP) images were reviewed for accurate and consistent display of the lumen over the four velocities. In triphasic flow, the trigger delays for FBI were set to 0 msec and 260 msec and in monophasic flow they were set to 0 msec and 300 msec so that the readouts coincided with phases of slow and rapid flow (20).

Data Analysis

Signal within the vascular phantom was measured within public domain software (ImageJ, National Institutes of Health, Bethesda, MD) using region analysis tools. For TOF and QISS, the luminal signal intensity was measured using a circular region-of-interest (ROI) on an axial source image at the center of the stenosis and 15 mm distal. The signal intensity for FBI and Ghost was measured at the same locations except using rectangular ROIs on a full-thickness coronal MIP rather than axial slices (ROI locations are displayed in Fig. 4 as dashed boxes). These measurements were taken over the eight flow velocities in order to determine velocity dependence. Finally, a montage of coronal MIPs with 5 mm thickness centered on the stenotic region was created to qualitatively compare the image quality of the methods, as well as to identify any artifacts. The plastic tubing and the plastic stenotic fitting were over 9 mm in diameter and had no MR signal; thus, it was not possible for partial volume averaging to distort the 5-mm-thick MIP images of the phantom.

RESULTS

During optimization of the fast spin-echo methods, ghost artifacts were observed in the phase-encoding direction at echo times less than 89 msec. At flip angles larger than 90° the intensity of these ghost artifacts increased and the images demonstrated increasing levels of blurring in the phase-encoding direction. These effects worsened at higher flow velocities. Artifacts and blurring were minimized over the widest range of velocities when using an echo time of 89 msec and a flip angle of 90°; these parameters were deemed optimal for both methods and used in the subsequent experiments. Optimized configurations of FBI and Ghost methods were subsequently evaluated with and without flow compensation.

Figure 4 shows a montage of coronal 5-mm-thick MIP images for each technique at the location of stenosis. Except for the Ghost method, which is predicated on the creation and composition of ghost artifacts, the MIP images in no way obscured the presence of pulsatility artifacts. Among the methods, QISS best displayed the stenotic fitting across all peak flow velocities. TOF produced images with similar appearance to QISS, but showed less of the tapered geometry at the proximal and distal ends of the stenotic fitting (white and gray arrows). The TOF method also demonstrated signal loss distal to the stenosis (arrowhead) for flow waveforms with peak velocities ≥ 42.4 cm/s. FBI images acquired without flow compensation depicted most of the stenosis at low velocities. Above 9.7 cm/s no appreciable signal was observed within the stenosis when using uncompensated FBI. Signal within the nonstenotic lumen gradually diminished as the velocity was increased above 21 cm/s. A similar pattern was seen in the Ghost images acquired without flow compensation. At 3.9 cm/s, Ghost displayed stenotic signal which gradually diminished as the velocity increased. At 9.7 cm/s Ghost exhibited the tapered geometry with signal loss at the center of the stenosis. As the flow velocity increased, the area of signal loss with Ghost extended outward from the central region of the stenosis toward the distal and proximal ends of the tube.

With use of flow compensation, the FBI and Ghost methods depicted more of the stenosis at larger flow velocities. Flow compensated FBI displayed signal within the stenotic region beginning at 3.9 cm/s that showed improved geometry yet sporadic signal loss (open arrows) as velocity increased. FBI also displayed signal loss outside of the stenotic region until 40 cm/s and 65 cm/s, where it displayed the poststenotic and prestenotic lumen, respectively. Flow compensated Ghost depicted signal within the stenotic region of the phantom that increased with velocity beginning at 3.9 cm/s. Ghost displayed signal within the nonstenotic segments of the phantom at a smaller velocity than FBI (3.9 cm/s), yet at higher velocities showed a narrower vessel lumen.

The quantitative results of all methods are displayed in Fig. 5. All data points in these plots represent an average over signal measurements from three scans with standard error shown. For each MR technique the data points were normalized to the maximum value found across the eight flow velocities. Normalization was performed to best compare the signal trends of the techniques across all velocities on a single plot.

Figure 5a displays the signal measurements obtained in the full diameter region of the flow phantom 30 cm proximal to the stenosis. These data were used as a control to identify any changes in signal characteristics caused by the stenosis. TOF and QISS followed similar trends, with signals increasing to a maximum at ≈ 40 cm/s and 25 cm/s, respectively. This signal was maintained at higher velocities. QISS signal rose to a maximum at lower velocities than TOF, with 50% and 75% of peak signal obtained at velocities of ≈ 2.5 cm/s and 5 cm/s for QISS versus 6 cm/s and 10 cm/s for TOF. FBI and Ghost signal trends were similar with and without use of flow compensation. Normalized signal values FBI and Ghost

acquired without flow compensation rose to a maximum at 10 cm/s; much sooner than TOF and QISS. However, these signals fell as flow velocity was increased. Normalized signal for uncompensated FBI fell more quickly than that for uncompensated Ghost, where at 60 cm/s it approached zero while Ghost maintained 50% of its maximum signal. Flow compensated FBI and Ghost signals appeared similar and monotonically increased to a maximum at ≈ 80 cm/s.

Signal measurements obtained at the location stenosis are presented in Fig. 5b. TOF and QISS, as well as FBI and Ghost, followed similar trends to Fig. 5a; however, all techniques reached their maximum signal values at a lower waveform flow velocity. QISS, uncompensated FBI, and uncompensated Ghost signals peaked at ≈ 5 cm/s (estimated intrastenotic peak velocity = 20 cm/s assuming a 75% reduction of cross-sectional area), while in TOF it occurred later, at 15 cm/s. Uncompensated FBI and Ghost demonstrated a steeper signal drop than seen in Fig. 5a and uncompensated FBI signal approached zero at ≈ 20 cm/s (estimated intrastenotic peak velocity = 80 cm/s), sooner than Ghost. With use of flow compensation, FBI normalized signal rose and peaked sooner than Ghost. Normalized signal for flow compensated Ghost surpassed that for flow compensated FBI at 45 cm/s (estimated intrastenotic peak velocity = 180 cm/s) and was larger at higher flow velocities.

Figure 5c shows signal measurements obtained 15 mm distal to the stenosis. All MR techniques demonstrated a slower rise in signal when compared with Fig. 5b, yet slightly faster when compared with Fig. 5a. Compared with the other methods, normalized signal for the QISS technique peaked at the lowest velocity, whereas signal for TOF reached a maximum at 20 cm/s and decreased to 40% of peak signal at higher flow velocities. Similar to the results shown in Fig. 5a, normalized signals for uncompensated FBI and Ghost were maximal at a peak flow velocity of ≈ 10 cm/s and decreased at higher velocities; normalized signal for FBI also decreased at a faster rate than for Ghost. Flow compensated FBI and Ghost signal trends were also similar to those in Fig. 5a; both trends monotonically increased to a maximum at 80 cm/s. Uncompensated FBI and Ghost signal trends demonstrated a slower fall than shown in Fig. 5b when approaching high flow velocities.

Normalized signal measurements obtained in the full-diameter phantom with monophasic flow are presented in Fig. 5d, and were almost identical to the results obtained with triphasic flow in Fig. 5a. These results spanned a shorter peak flow velocity range than Fig. 5a–c due to the reduced peak systolic velocity in the monophasic waveforms.

DISCUSSION

Although several nonenhanced MRA approaches have been proposed, little has been described about their performance characteristics when imaging flow velocities and patterns found in the lower extremities. Here, four nonenhanced MRA techniques were evaluated in a flow phantom with 50% diameter stenosis under triphasic and monophasic flow conditions. Signal characteristics were found to vary substantially between techniques. In particular, large differences in signal trends and image quality were found between the transient steady-state methods of TOF and QISS and the fast spin-echo methods of FBI and Ghost.

TOF showed consistent intrastenotic signal similar to QISS, yet demonstrated poststenotic signal loss for waveforms with peak flow velocities ≥ 42.4 cm/s. Signal loss distal to the stenosis during TOF imaging has been reported and is caused by dephasing related to poststenotic turbulence from deceleration of the fluid as it exits the stenosis (21). The extent of dephasing is related to the level of turbulence, which increases with velocity (22). The latter was confirmed in the TOF montage in Fig. 4, which showed the area of poststenotic

signal loss increasing with velocity. This type of signal loss was not observed in QISS, presumably due to its use of a shorter echo time (1.3 msec vs. 7.0 msec) and echo train duration (323 msec vs. 526 msec). Signal loss due to phase dispersion is known to increase with the echo time (23,24).

Key performance differences between the TOF and QISS methods likely stem from differences in their imaging readouts. TOF employs a spoiled gradient-echo readout while QISS uses a bSSFP readout. This difference is highlighted in Fig. 4, where TOF exhibited signal loss at the edges of the distal end of the stenosis (gray arrow) which was not seen in QISS. We attribute this loss of signal to saturation of slow or recirculating flow near the edge of the lumen as well as to dephasing within vortical flow patterns that form near the edges of the tube at the distal end of the stenosis. Vortical flow occurs within tapered narrowings, as reported previously (25,26). Signal loss artifact was not seen in the QISS image due to its use of a bSSFP readout that refocuses and does not spoil the transverse magnetization. Underestimation of vessel diameter and overestimation of stenosis has also been reported with spoiled gradient-echo readouts (27), which manifested in Fig. 4 as narrower lumen width in both the stenotic and nonstenotic regions when compared to QISS. This effect was also evident in the source images.

In a nonstenotic, pulsatile flow phantom study performed by Edelman et al (16), the QISS technique reached maximum signal at lower velocities than TOF, and its signal remained constant at high velocities under triphasic waveforms. Our results agreed with these findings and further demonstrated that QISS displayed consistent signal at high velocities in the vicinity of a 50% diameter stenosis. The QISS technique best represented the tapered geometry of the stenosis over the widest range of velocities among the compared techniques. This consistency across velocities can likely be attributed to the use of a readout that occurs during the period of slow flow in the cardiac cycle. The use of a diastolic acquisition window avoids flow artifacts seen when imaging during peak flow, and its advantages in the setting of pulsatile flow have been reported (28,29).

TOF and QISS signal measurements obtained under monophasic flow, which spanned the shorter peak flow velocity range of 0 cm/s to 50 cm/s, mimicked the signal trends in the comparable velocity range in the nonstenotic triphasic flow measurements. The similarity of these results suggests the robustness of the evaluated methods under monophasic (diseased) flow conditions. Remarkably, FBI and Ghost acquired with and without flow compensation produced similar results to their nonstenotic measurements under triphasic flow. FBI and Ghost are subtractive techniques that rely on a flow velocity differential between systole and diastole, and this differential is reduced in monophasic flow when compared with triphasic flow.

The FBI method may be applied with the frequency-encoding direction parallel or perpendicular to the direction of flow (30). A frequency-encoding direction parallel to the direction of flow is preferred for depicting blood vessels containing slower flow and a frequency-encoding direction perpendicular to the direction of flow is preferred for imaging vessels containing faster flow. It is likely that the signal peak for FBI would have shifted towards higher velocities had a frequency-encoding direction perpendicular to the direction of flow been used, similar to what was observed with flow compensation. Although not investigated in this report, the use of larger refocusing flip angles, as reported by Storey et al (31), would likely improve depiction of intermediate or high flow velocities.

FBI and Ghost techniques performed similarly and both exhibited trends that contrast with TOF and QISS. This can be attributed to similarities in their acquisition, since both use fast spin-echo readouts with similar echo train durations and RF spacings. Ghost, however,

performed slightly better than FBI at intermediate and high flow velocities by showing less signal loss without flow compensation and more complete geometry of the stenosis with flow compensation when compared to FBI (open arrows in Fig. 4). The improved performance with Ghost likely relates to differences in reconstruction and acquisition methods. FBI subtracts signal intensities from two acquisitions in the image domain, while Ghost is predicated on the complex difference of acquired k -space data to create ghost artifacts. Complex difference processing is substantially advantageous to magnitude subtraction when signals have equal magnitudes but different phases. The advantage of complex subtraction has previously been noted in a flow phantom study by Hoogeveen et al (32), where its use minimized intrastenotic signal loss observed with magnitude subtraction. Further, FBI requires the use of fixed trigger delays which might not be optimal for depicting flow patterns with limited periods of stasis. Conversely, the ungated technique of Ghost, which acquires data at pseudorandom trigger times throughout the cardiac cycle, might be advantageous. Ghost, however, only provides a 2D angiogram, in contrast with FBI, TOF, and QISS which can produce 3D angiograms.

Although FBI and Ghost may be configured to image a large range of flow velocities through use of flow compensation, there existed a range of velocities at the intersection of the uncompensated and flow compensated curves of FBI and Ghost which can be seen in Fig. 5a–d that demonstrated reduced signal intensity. This intersection lies within the physiological flow velocity range of 16 cm/s to 40 cm/s (4 cm/s to 20 cm/s in Fig. 5b). This finding suggests that further alterations of sequence parameters, such as increasing the refocusing flip angle (31) or partial flow compensation (30), may be required to configure FBI and Ghost methods for optimal depiction of intermediate flow velocities.

This study had notable limitations. First, only a single cardiac interval of 820 msec was simulated. It is possible that heart rate may affect the performance of the techniques. FBI may exhibit reduced signal at short R-R intervals due to saturation of vascular spins by its use of a coronal imaging slab, whereas TOF and QISS would largely avoid this effect due to their use of axial slices. Ungated Ghost is not synchronized to cardiac interval and would not be affected in the same manner as FBI, but would demonstrate reduced signal in the setting of pseudogating when the cardiac interval approximately equals the sequence repetition time. Second, this study was only performed at 1.5T. It is possible that the methods may perform differently at 3T. In particular, the spoiled gradient-echo and fast spin-echo approaches of TOF, FBI, and Ghost may be advantageous to bSSFP-based QISS at 3T due to off-resonance effects. Further investigation at 3T is required. Lastly, only a limited number of sequence configurations were studied. We chose the best parameters in context of the experimental setup; however, TOF, FBI, Ghost, and QISS can be customized to image a variety of flow conditions.

In conclusion, TOF and QISS exhibited similar signal trends under triphasic and monophasic flow waveforms; however, QISS showed the most consistent signal in the proximity of the stenosis over the range of velocities investigated. QISS also displayed fewer flow-related artifacts than TOF and was less affected by signal loss distal to the stenosis. Similarly configured FBI and Ghost MRA methods depicted the stenosis with variable fidelity and demonstrated a strong dependence on the use of flow compensation. Remarkably, all methods performed similarly under monophasic flow conditions as they did under triphasic flow. Future studies using different stenosis geometries, flow patterns, and magnetic field strengths will be performed in order to attain a more comprehensive understanding of these MRA methods.

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Figure 1. Schematic of the tapered stenotic connector in the sagittal plane. This connector was designed to mimic a 50% diameter stenosis.

Figure 2. Triphasic waveforms simulating flow in the femoral artery at eight peak flow velocities. Flow velocity was measured 10 cm proximal to the stenosis by phase contrast imaging.

Figure 3.
Monophasic waveforms simulating diseased blood flow commonly seen distal to severe stenosis.

Figure 4.

Coronal maximum intensity projection images (5 mm thickness) through the region of stenosis for each MRA method. White and gray arrows identify signal loss at the proximal and distal ends of the stenosis for the TOF method, respectively. The arrowhead identifies poststenotic signal loss exhibited by the TOF method. The open arrows identify intrastenotic signal loss in flow compensated FBI. The dashed boxes depict the locations of the intrastenotic and poststenotic measurement ROIs. FC, flow compensation.

Figure 5.

Normalized signal measurements obtained for each of the four methods (**a**) at a region of no stenosis, (**b**) at the location of the 50% diameter stenosis, and (**c**) 15 mm distal to the center of stenosis under the triphasic flow patterns shown in Fig. 2. (**d**) Normalized signal measurements for each of the four methods at a region of no stenosis under the monophasic flow patterns shown in Fig. 3. Data are the average of three measurements. Error bars denote standard error.

MRI Parameters

Table 1

	Slices	Slice thickness (mm)	FOV (mm)	Matrix	FA (degrees)	TR/ETD (msec)	RF TR/TE (msec)	Slice PF	Phase PF	GRAPPA	Bandwidth (Hz/pixel)	Flow comp	TD (msec)	TA (s)
TOF ^a	63	2	400 × 275	384 × 384	70	1 R-R/526	47.8/7.0	–	6/8	–	114	Yes	–	403
QISS ^{a,b}	63	2	400 × 280	400 × 400	90	1 R-R/323	3.4/1.4	–	5/8	2	694	No	100	53
FBI ^{a,c}	33	2	400 × 279	384 × 384	90	1 R-R/241	3.3/89.1	6/8	5/8	2	868	Yes/No	0 and 260	114
Ghost ^{c,d}	256	2	400 × 279	384 × 384	90	1200/255	3.0/88	6/8	5/8	4	1002	Yes/No	–	115

FOV = Field of View, FA = Flip Angle, TR = Repetition Time, ETD = Echo Train Duration, RF TR = Interpulse Repetition Time, TE = Echo Time, TI = Inversion Time, PF = Partial Fourier, TD = Trigger Delay, TA = Acquisition Time, R-R = Simulated Cardiac Interval; GRAPPA = Generalized Autocalibrating Partially Parallel Acquisition Factor.

^aTOF, QISS, and FBI were all performed with ECG gating using a simulated cardiac interval of 820 msec.

^bThe quiescent inflow time period for QISS was 350 msec.

^cThe FBI and Ghost techniques collected one partition in 2 and 1 shots, respectively.

^dNonselective RF pulses were used with the Ghost acquisition.