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Accelerated Whole-heart Magnetic Resonance Angiography Using a Variable-Density Poisson-Disc Undersampling Pattern and Compressed Sensing Reconstruction

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Abstract

Purpose: To accelerate whole-heart three-dimension magnetic resonance angiography (MRA) by employing a variable-density Poisson-disc undersampling pattern and a compressed sensing (CS) reconstruction algorithm, and compare the results to sensitivity encoding (SENSE).

Methods: For whole-heart MRA, a prospective variable-density Poisson-disc k-space undersampling pattern was developed in which 1–2% of central part of k-space was fully sampled, and sampling in the remainder decreased exponentially toward the periphery. The undersampled data was then estimated using CS reconstruction. In patients, images using this sequence with an undersampling rate of ≈ 6 were compared with those using a SENSE rate of 2 ($n=15$) and a SENSE rate of 6 ($n=13$).

Results: Compared to SENSE rate 2, CS rate 6 images had similar objective border sharpness, significantly lower subjective image quality scores at all 4 locations (all $p < 0.01$), and shorter scan times ($p < 0.05$). Compared to SENSE rate 6, CS rate 6 had similar objective border sharpness at all 4 locations, significantly better subjective image quality scores at 3 of 4 locations (all $p < 0.01$), and similar scan times ($p=0.24$).

Conclusion: Compared to SENSE with a comparable acceleration rate, a variable-density Poisson-disc undersampling pattern and CS reconstruction achieved better subjective image quality and similar border sharpness.

Keywords

Whole-heart magnetic resonance angiography; respiratory motion; self-navigator; variable-density undersampling; compressed sensing

INTRODUCTION

Whole-heart three-dimension magnetic resonance angiography (3D-MRA) with electrocardiogram (ECG) and respiratory gating provides a high-quality depiction of the heart and chest vasculature (1). It is widely available and extensively used in clinical practice for coronary angiography (2) and anatomic assessment of congenital heart disease (3). The respiratory gating allows for a free-breathing acquisition, making it applicable to patients who are too young or too ill to hold their breath. In addition, because the acquisition is not confined to the time of a breath-hold, high spatial resolution can be attained.

We previously reported a modification of whole-heart 3D-MRA which utilized 1) a novel prospective respiratory “self-gating” approach (Heart-NAV) that channels the gating signal through the conventional respiratory navigator processing chain, and 2) an intravascular contrast agent coupled with an inversion recovery (IR) preparation pulse (4). The intravascular contrast agent and IR pulse timed to null myocardium were employed to improve the signal-to-noise and contrast-to-noise ratios (5). Use of Heart-NAV eliminated the image artifact produced by the navigator restore pulse that is needed when an IR pulse is used in conjunction with a conventional diaphragmatic respiratory navigator. Moreover, Heart-NAV has the potential to improve respiratory motion compensation because it tracks the heart rather than the diaphragm position.

Despite these enhancements, whole-heart 3D-MRA acquired at high spatial resolution still suffers from a relatively long scan time—approximately 15 minutes for 1.2 mm isotropic resolution (6). During a long image acquisition, the patient’s body position, heart rate, and breathing pattern may change leading to reduced image quality or an incomplete scan (7). Therefore, in this study, we accelerate the Heart-NAV 3D-MRA IR sequence by a factor of 6 using a variable-density Poisson-disc random undersampling pattern and, given the sparsity of the data, employ a compressed sensing (CS) algorithm to reconstruct the images (8,9). This method was compared with a more conventional acceleration approach, sensitivity encoding (SENSE) (10), performed with an undersampling factor of 2, a typical value in the clinical setting, and 6 to match the acquisition time of our proposed technique. With an eye toward routine clinical implementation, we also compared the CS reconstruction time using a graphics processing unit (GPU) versus a central processing unit (CPU).

METHODS

Whole-heart 3D-MRA Sequence

We have previously described in detail the contrast-enhanced whole-heart 3D-MRA IR sequence with Heart-NAV respiratory motion compensation used throughout this study (Figure 1) (4). Briefly, a 3D steady-state free precession (SSFP) sequence is used to acquire the MRA data in a sagittal orientation with frequency encoding in the superior-inferior

direction. To minimize cardiac motion artifacts, the data is divided into multiple segments, each of which is run during the quiescent period of the heart set by the trigger delay and shot duration. Five startup pulses are employed to drive the net magnetization vector into the steady-state for the subsequent MRA acquisition. The first startup pulse of the SSFP sequence is borrowed to measure the center line of k-space along the superior-inferior direction. Using the conventional diaphragm navigator processing pathway, the data from that echo is collected and transformed from the Fourier domain to the image domain. This image data represents the 1-dimensional projection line of the whole-heart MRA imaging volume in the superior-inferior orientation. The image data is processed and displayed using the same procedure as the diaphragm navigator including a cross-correlation analysis with the preceding line to measure displacement in the superior-inferior direction and prospectively gate the respiratory motion of the heart and track (i.e., adjust) the position of the imaging volume. An acceptance window at the end-expiration position is set during the first few seconds of the scan and the window width is pre-specified by the user. Fat and fold-over suppression pre-pulses precede the Heart-NAV to nullify the signal from the fat and the structures outside the field of view. Moreover, a nonselective IR pre-pulse timed to null the signal from the myocardium is applied to promote myocardium-blood contrast.

Variable-Density Poisson-Disc Undersampling

For the MRA data, a variable-density Poisson-disc undersampling pattern was implemented such that k-space sampling was dense centrally and decreased exponentially toward the periphery. With this approach, a new random point is chosen in the k_y - k_z plane of k-space, and the distances between this point and the previous points within a neighborhood region are calculated. If any of the distances is less than the threshold value, the point is rejected and a new random point is selected. The threshold value to accept or reject a point exponentially increases moving toward the periphery of k-space. This procedure is repeated until the minimum number of points in k-space is selected to achieve the desired degree of undersampling and acceleration. Then, 1–2% of the central part of k-space is fully sampled to use for calculation of coil sensitivities. The selected k-space points are sorted in a manner that minimizes the phase jump within each shot of the 3D-MRA acquisition. Assuming each shot consists of V k-space points (i.e., V views), k-space is first sorted based on magnitude in the k_y - k_z plane (i.e., $\sqrt{k_y^2 + k_z^2}$) from low to high and divided equally into V groups. The k-space points within each group are then sorted based on phase in the k_y - k_z plane (i.e., $\tan^{-1} \frac{k_z}{k_y}$). The first k-space points from each group are concatenated to generate a shot to be acquired at each cardiac cycle. Therefore, the points within each shot are sorted based on their Euclidean distance from the center of k-space, and the acquired line in each shot is filled from the center to outer k_y - k_z plane. Figure 2 shows the sorted k-space profile order for 6 different shots in a 3D-MRA acquisition with an undersampling factor of 6.

Compressed Sensing Parallel Image Reconstruction

To reconstruct the MRA images from the undersampled k-space data, we implemented the L_1 -ESPIRiTCS reconstruction algorithm (11) on both a CPU and a GPU. In this algorithm, the data from up to 28 receiver coils are projected onto 8 virtual orthogonal coils using

eigenvalue and eigenvector decomposition (12,13). The fully sampled 1–2% of the central part of k-space is then used to estimate the coil sensitivities of the 8 virtual receiver coils. Lastly, the images are reconstructed by iteratively solving the following non-linear cost function (9):

$$m^1, \dots, m^M = \underset{m^1, \dots, m^M}{\operatorname{argmin}} \sum_{i=1}^N \left\| \sum_{j=1}^M C_i^j m^j - y_i \right\|_{PF}^2 + \alpha |R(m^1, \dots, m^M)|$$

where m^j are the estimated images, M is the number of ESPIRiT sets of coil maps, N is the number of virtual coils, F is the Fourier transform, P is the operator imposing the undersampling pattern for each phase, y^j are the measured k-space data for each coil, C^j are the ESPIRiT coil maps estimated from the central part of k-space, R is a penalty function that incorporates *a priori* knowledge such as the compressibility of the wavelet domain using L_1 minimization of the wavelet coefficient, and α is the regularization parameter.

Human Study

To evaluate the whole-heart MRA sequence with Heart-NAV respiratory motion compensation and CS reconstruction described above, patients referred for a magnetic resonance examination and with a clinical indication for contrast-enhanced whole-heart MRA were prospectively recruited. The Boston Children's Hospital Committee on Clinical Investigation approved this study, and written informed consent was obtained from all subjects. Examinations were performed on a 1.5T Achieva dStream scanner (Philips Healthcare, Best, the Netherlands). In each subject, a high temporal resolution cine SSFP 4-chamber slice was acquired to identify the trigger delay and shot duration for the whole-heart MRA. The subject was then administered a bolus infusion of 0.03 mmol/kg of gadofosveset trisodium contrast. Prior to each MRA, a Look-Locker sequence was performed to determine the inversion time to null the myocardial signal.

Two comparisons were initially performed, each in a different patient cohort. In the first, the whole-heart MRAs were performed with 1) a conventional approach using an undersampling factor of 2 (in the anterior-posterior direction) and SENSE reconstruction, and 2) variable-density Poisson-disc undersampling at a factor of 6 with CS reconstruction. Acquisition parameters for both were as follows: field of view 386 (SI) $\times$ 230 (AP) $\times$ 146 (RL) mm, voxel size 1.6 mm³ reconstructed to 0.8 mm³, flip angle 90°, echo time 2.4 ms, repetition time 4.7 ms, bandwidth 543 Hz, shot duration 43–120 ms, Heart-NAV acceptance window 3 mm, and tracking factor 1. Assuming no undersampling, 100% respiratory gating efficiency, and a heart rate of 84 beats per minutes, the scan time was 9.6 minutes. For the second comparison, whole-heart MRAs were performed with similar acceleration rates: 1) an undersampling factor of 6 (2 anterior-posterior direction $\times$ 3 right-to-left direction) and SENSE reconstruction, and 2) variable-density Poisson-disc undersampling at a factor of 6 with CS reconstruction. Acquisition parameters for both were as follows: field of view 306 (SI) $\times$ 209 (AP) $\times$ 127 (RL) mm, voxel size 1.2 mm³ reconstructed to 0.6 mm³, flip angle 90°, echo time 2.5 ms, repetition time 5.0 ms, bandwidth 543 Hz, shot duration 47–140 ms, Heart-NAV acceptance window 3 mm, and tracking factor 1. The sequences were acquired

in random order. Assuming no undersampling, 100% respiratory gating efficiency, and a heart rate of 88 beats per minutes, the scan time was 14.1 minutes. The actual scan time of the MRA sequences was measured using a stopwatch. The higher spatial resolution for the second comparison was chosen to leverage the acceleration for better spatial resolution and keep the scan time approximately the same as the standard scan with SENSE factor 2.

Following the initial two comparisons, it became of interest to explore an intermediate CS acceleration factor of 4. At this point, gadofosveset trisodium contrast had been removed from the market and was thus not available for use. We therefore did a comparison of CS factor 4 and SENSE factor 4 using a conventional whole-heart 3D-MRA SSFP sequence with a T2-preparation pulse and no intravenous contrast. Acquisition parameters were the same as those in the second comparison above.

Image Analysis

The MRA datasets acquired with SENSE were reconstructed in-line using the standard scanner hardware immediately following scan completion. MRA datasets acquired with variable-density Poisson-disc undersampling were reconstructed offline using the *L1-ESPIRiTCS* algorithm on both a CPU and a GPU. The CPU node had dual Intel E5 2650 CPUs @ 2.00 GHz and 128 GByte of random-access memory and the GPU node had an NVIDIA Tesla K40m GPU with 2880 computing cores and 12 GB of global memory. The CS reconstruction times for both the CPU and the GPU were measured by embedding time stamps within the reconstruction software code.

The subjective image quality of the lower left pulmonary vein (LLPV), main pulmonary artery (MPA), ascending aorta (AAo), and ventricular septum (VS) was graded independently by 2 clinicians based on a 5-point scale (1-poor/non-diagnostic; 2-fair/moderate blurring; 3-acceptable/mild blurring; 4-good/no blurring; and 5-excellent/sharp borders). Scores ≥ 3 were pre-defined as diagnostic. For all patients, the whole-heart 3D-MRA was performed as part of a comprehensive examination protocol which included assessment of ventricular function and blood flow. The primary clinical goal of the whole-heart 3D-MRA sequence was also recorded. Images were visually assessed for aliasing artifact. In addition, objective border sharpness was measured using an in-house software tool which yields values from zero to infinity with higher values being sharper (14). To quantitatively compare GPU and CPU CS reconstructions, the mean voxel intensity difference between the 2 reconstructions was calculated. The difference should be 0 if the images reconstructed on GPU and CPU are identical.

Statistical Analysis

Descriptive statistics are reported as median (range) and mean $\pm$ standard deviation. A two-tailed paired Student's t-test was used to compare the border sharpness scores, and a non-parametric signed-rank test was used to compare the subjective image quality scores. A *p*-value ≤ 0.05 was considered statistically significant.

RESULTS

Whole-heart MRA SENSE Factor 2 Versus CS Factor 6

Fifteen patients ((6 females, median age 19 years (range, 2–41), median weight 60 kg (range, 10–96), median heart rate 85 bpm (range, 60–115)) underwent contrast-enhanced whole-heart 3D-MRA IR acquisitions with a SENSE factor of 2 and with variable-density Poisson-disc undersampling at a factor of 6 with CS reconstruction. The primary clinical goal of the whole-heart 3D-MRA sequence was assessment of the thoracic aorta (n=5), pulmonary arteries (n=4), coronary artery course (n=3), intracardiac anatomy (n=2), and pulmonary veins (n=1). All acquisitions and image reconstructions were successfully completed. Representative images from 2 subjects are shown in Figure 3. The mean subjective image quality scores, objective border sharpness, and scan time for all subjects are shown in Table 1. There was no significant difference in the objective border sharpness between the SENSE factor 2 and CS factor 6 for all 4 locations. The subjective image quality scores for CS factor 6 were significantly lower than those for SENSE factor 2 by approximately 1 grade for all locations. None of the SENSE acquisitions had aliasing artifact and 4 of 15 CS acquisitions had aliasing artifact. The mean scan time for the CS acquisition was approximately 62% shorter than that for the SENSE acquisition. The mean scan efficiency was $52 \pm 16\%$ for SENSE and $40 \pm 11\%$ for CS.

Whole-heart MRA SENSE Factor 6 Versus CS Factor 6

Thirteen patients ((6 females, median age 11 years (range, 2–57), median weight 38 kg (range, 10–106), median heart rate 93 bpm (range, 65–120)) underwent contrast-enhanced whole-heart 3D-MRA IR acquisitions with a SENSE factor of 6 and with variable-density Poisson-disc undersampling at a factor of 6 with CS reconstruction. The primary clinical goal of the whole-heart 3D-MRA sequence was assessment of the thoracic aorta (n=5), pulmonary veins (n=3), intracardiac anatomy (n=2), pulmonary arteries (n=1), coronary artery course (n=1), and systemic veins (n=1). All acquisitions and image reconstructions were successfully completed. Representative images from 2 subjects are shown in Figure 4. The mean subjective image quality scores, objective border sharpness, and scan time for all subjects are shown in Table 2. There was no significant difference in the objective border sharpness between SENSE factor 6 and CS factor 6 for all 4 locations. The subjective image quality scores for CS factor 6 were significantly better than for SENSE factor 6 for the MPA, AAO, and VS. Eight of 13 SENSE acquisitions and 3 of 13 CS acquisitions had aliasing artifact. Mean scan time was not significantly different between the 2 acquisitions. The mean scan efficiency was $39 \pm 14\%$ for SENSE and $40 \pm 12\%$ for CS.

Whole-heart MRA SENSE Factor 4 Versus CS Factor 4

Five patients ((3 females, median age 31 years (range, 18–38), median weight 57 kg (range, 55–85), median heart rate 84 bpm (range, 70–90)) underwent two whole-heart 3D-MRA SSFP acquisitions; one with a SENSE factor of 4 and the other with a variable-density Poisson-disc undersampling at a factor of 4 with CS reconstruction. The primary clinical goal of the whole-heart 3D-MRA sequence was assessment of intracardiac anatomy (n=2), thoracic aorta (n=1), pulmonary arteries (n=1), and coronary artery course (n=1). All acquisitions and image reconstructions were successfully completed. Mean scan time was

7.7 $\pm$ 1.5 minutes for SENSE factor 4 and 7.8 $\pm$ 1.2 minutes for CS factor 4. Representative images from all 5 subjects are shown in Figure 5. Given the small sample size, statistical comparison of image quality scores and border sharpness was not performed; however, both clinicians reported that CS factor 4 yielded better overall subjective image quality in each of the 5 patients.

CPU Versus GPU

The mean voxel intensity difference between images reconstructed by CS algorithm on the CPU and the GPU was quite small (mean difference 4.0×10^{-8} , maximum difference 2.0×10^{-6} , and mean voxel value 0.1), indicating that they were comparable. The reconstruction time with the GPU was significantly shorter than with the CPU (1.3 $\pm$ 0.4 min versus 17 $\pm$ 8 min, $p < 0.001$).

DISCUSSION

We developed and evaluated in patients a prospective variable-density Poisson-disc undersampling pattern to acquire a high-resolution contrast-enhanced whole-heart 3D-MRA during free-breathing in approximately 5 minutes. In this technique, $\approx 1\%$ of central part of k-space was fully acquired and the sampling rate was exponentially decreased toward the peripheral part of k-space to achieve an approximately 6-fold acceleration in scan time. Our data acquisition was also prospectively adjusted to correct for respiratory motion by tracking the heart using our Heart-NAV technique. Images were then reconstructed off-line using a CS algorithm (L_1 -ESPIRiT).

Compared to a SENSE factor of 2, our technique had a lower subjective image quality score by approximately 1 grade and no difference in objective border sharpness of the ascending aorta, main pulmonary artery, lower pulmonary vein, and ventricular septum. The worse image quality score is most likely related to a lower signal-to-noise ratio that comes from the greater degree of k-space undersampling and acceleration. We could not quantitatively compare the signal-to-noise ratios because a reliable measurement technique is not available for CS reconstructed data (15). Compared to a SENSE factor of 6, our technique had significantly better subjective image quality scores in 3 of 4 locations, no difference in objective border sharpness, and no difference in scan time. Similarly, in the non-contrast MRAs performed with an acceleration factor of 4, image quality of CS reconstructed images tended to be better than SENSE. The superior image quality is probably due to the L_1 minimization norm in CS reconstruction algorithm which mitigates the noise better than SENSE. In addition, our variable-density Poisson-disc undersampling algorithm samples the central part of k-space more densely than SENSE.

The processing time for CS reconstruction using a GPU was dramatically shorter than with a CPU. This is not surprising as the GPU has much greater parallel processing capability. The average CS reconstruction time of 1.3 minutes on the GPU suggests that this approach may be suitable for in-line reconstruction on the scanner in the clinical environment. The slight difference in image intensity that we found between the 2 computer hardware configurations may be because the CS algorithm was implemented using sequential coding on the CPU and

parallel coding on the GPU leading to a different order of operations, and therefore, unequal results (16).

Other investigators have employed CS reconstruction for cardiac magnetic resonance using a variety of undersampling patterns, and some have also compared their CS approach to parallel imaging. Akcakaya *et al.* performed high-resolution whole-heart 3D-MRA using SENSE rate 6 as well as CS rate 6 in healthy volunteers (15). Similar to our results, they found that the CS images had 1 grade better subjective image quality and an equivalent objectively measured sharpness for the left coronary artery. In their CS implementation, they employed uniform random undersampling which may produce large gaps among the collected samples and lead to reduced sharpness in the reconstructed images. To address this concern, Vasanawala *et al.* employed a Poisson-disc random sampling pattern which enforces both a high degree of incoherence and a uniform distance between sampled k-space data (17). They also found that their CS images had superior image quality compared to those obtained with parallel imaging (autocalibrating reconstruction for Cartesian sampling) using a similar acceleration rate of approximately 4. For a further reduction of scan time, Liu *et al.* pursued a potential enhancement to the Poisson-disc approach by adding variable-density sampling in which the center of k-space is more frequently acquired than the periphery (18). They were aiming to accelerate a 4-dimensional flow acquisition and retrospectively applied the variable-density Poisson-disc undersampling to a fully acquired k-space dataset. This retrospective undersampling of data, however, may not match the results from prospective undersampling because the non-uniform jumps in k-space and the associated large gradient changes and eddy currents do not occur (19). Subsequently, Cheng *et al.* developed a prospective variable-density Poisson-disc undersampling pattern for 4-dimensional flow acquisitions (20). They used a radial-like k-space profile ordering on a Cartesian grid and acquired each shot from different k-space regions using a golden-ratio permutation to reduce respiratory motion artifact. In contrast, our technique does not require any specific k-space profile ordering because it uses other means for respiratory motion correction (i.e., Heart-NAV).

Rather than undersampling k-space data, an alternative approach to accelerate the whole-heart 3D-MRA sequence is to acquire data without respiratory gating (i.e., 100% scan efficiency) and retrospectively correct for the respiratory-induced heart motion using 2D rigid (21), 3D rigid (22,23), affine (24,25), or non-rigid transformations (26). These techniques, however, require computationally intense and retrospective motion correction algorithms which, at present, prolong image reconstruction time and limit tracking the imaging volume. Ultimately, a combination of k-space undersampling and motion correction may yield optimal results in the clinical setting.

Though the initial results of our 3D-MRA technique are encouraging, there are several limitations that should be addressed, and ways in which our strategy might be expanded. The superposition of static regions, such as chest wall and fat, may have reduced the fidelity of Heart-NAV tracking and limited image quality. The fidelity of Heart-NAV could be improved by using a combination of phased-array surface coil signals to reduce the superimposed signal from the static regions and fat surrounding the heart (27). In our study, CS reconstruction was performed off-line thereby limiting its clinical applicability.

Encouraged by the fast reconstruction time we found using a single GPU, we plan to reduce the reconstruction time to less than a minute by using multiple GPUs, and implement it in-line on the scanner. More broadly, we intend to extend our technique to acquire multiple whole-heart 3D-MRA datasets across the cardiac cycle (*i.e.*, 3D cine acquisitions) (28–31). Heart-NAV could be used for respiratory motion compensation in a free-breathing scan without interrupting the steady-state of the net magnetization vector, and variable-density Poisson-disc undersampling could reduce the imaging time to a clinically acceptable level.

CONCLUSION

We developed and evaluated in patients a high-resolution contrast-enhanced whole-heart 3D-MRA technique using a variable-density Poisson-disc undersampling pattern and CS reconstruction with a 6-fold acceleration factor. Compared to SENSE with similar acceleration factor, this technique had a better image quality score and comparable vessel sharpness. Based on these results, further development of this approach is warranted.

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Figure 1.

Diagram for contrast-enhanced whole-heart 3D-MRA IR with Heart-NAV. Fat sup, fat suppression pulse; FOS, fold-over suppression pulse; IR, inversion recovery; SP, startup pulses; SSFP, steady-state free precession pulse; TR, repetition time.

Figure 2.

Sorted k-space profile ordering for the whole-heart 3D-MRA sequence. The central $\approx 1\%$ of k-space is fully acquired and 16% of the peripheral part of k-space is randomly acquired using a variable-density Poisson-disc undersampling pattern.

Figure 3.

Whole-heart 3D-MRA acquisitions with SENSE factor 2 and with CS factor 6 in 2 patients, ages 8 and 2 years, with congenital heart disease. AO, aorta; LA, left atrium; LV, left ventricle; PA, pulmonary artery; PV, pulmonary vein; RA, right atrium; RV, right ventricle; VS, ventricular septum.

Figure 4: Whole-heart 3D-MRA acquisitions with SENSE factor 6 and with CS factor 6 in 2 patients, ages 9 and 4 years, with congenital heart disease. Arrows show aliasing artifact in the coronal and sagittal views in patient 1. AO, aorta; LA, left atrium; LV, left ventricle; PA, pulmonary artery; PV, pulmonary vein; RA, right atrium; RV, right ventricle.

Figure 5: Whole-heart 3D-MRA acquisitions with a T2-preparation pulse with SENSE factor 4 and with CS factor 4 in 5 patients. AO, aorta; LA, left atrium; LV, left ventricle; PA, pulmonary artery; PV, pulmonary vein; RA, right atrium; RV, right ventricle.

Table 1. Border sharpness, image quality, and scan time for whole-heart MRA for SENSE x2 versus CS x6 (n=15).

	SENSE x2		CS x6		p-value	
	Border sharpness	Image quality	Border sharpness	Image quality	Border sharpness	Image quality
LLPV	0.69 ± 0.13	4.17 ± 0.75	0.74 ± 0.17	3.27 ± 0.64	0.23	<0.001
MPA	0.76 ± 0.09	4.57 ± 0.57	0.77 ± 0.15	3.60 ± 0.62	0.79	<0.001
AAo	0.68 ± 0.14	4.66 ± 0.55	0.73 ± 0.18	3.77 ± 0.57	0.31	<0.001
VS	0.44 ± 0.13	4.21 ± 0.92	0.43 ± 0.13	3.42 ± 0.79	0.92	<0.001
Scan time (min)	9.57 ± 5.47		3.62 ± 1.86		<0.001	

Values are mean ± standard deviation. Sharpness measure: 0 (blurred) to infinity (sharp). Image quality score: 1 (poor) to 5 (excellent). AAo, ascending aorta; LLPV, lower pulmonary vein; MPA, main pulmonary artery; VS, ventricular septum.

Table 2.

Border sharpness, image quality, and scan time for whole-heart MRA for SENSE x6 versus CS x6 (n=13).

	SENSE x6		CS x6		<i>p-value</i>	
	Border sharpness	Image quality	Border sharpness	Image quality	Border sharpness	Image quality
LLPV	0.75 ± 0.24	2.88 ± 1.08	0.78 ± 0.14	3.08 ± 0.70	0.50	0.38
MPA	0.78 ± 0.18	3.12 ± 0.97	0.76 ± 0.19	3.73 ± 0.88	0.46	<0.001
AAo	0.80 ± 0.13	3.19 ± 0.91	0.79 ± 0.13	3.73 ± 0.78	0.74	0.002
VS	0.40 ± 0.17	3.0 ± 1.02	0.40 ± 0.07	3.46 ± 0.77	0.95	<0.001
Scan time (min)	5.51 ± 1.96		5.23 ± 1.91		0.24	

Values are mean ± standard deviation. Sharpness measure: 0 (blurred) to infinity (sharp). Image quality score: 1 (poor) to 5 (excellent). AAo, ascending aorta; LLPV, lower pulmonary vein; MPA, main pulmonary artery; VS, ventricular septum.