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Free-breathing Whole-heart 3D Cine Magnetic Resonance Imaging with Prospective Respiratory Motion Compensation

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Abstract

Purpose—To develop and validate a new prospective respiratory motion compensation algorithm for free-breathing whole-heart 3D cine steady-state free precession (SSFP) imaging.

Methods—In a 3D cine SSFP sequence, 4 excitations per cardiac cycle are re-purposed to prospectively track heart position. Specifically, their 1D image is reconstructed and routed into the scanner's standard diaphragmatic navigator processing system. If all 4 signals are in end-expiration, cine image data from the entire cardiac cycle is accepted for image reconstruction. Prospective validation was performed in patients (n=17) by comparing in each a conventional breath-hold 2D cine ventricular short-axis stack and a free-breathing whole-heart 3D cine dataset.

Results—All 3D cine SSFP acquisitions were successful and the mean scan time was 5.9 ± 2.7 minutes. Left and right ventricular end-diastolic, end-systolic, and stroke volumes by 3D cine SSFP were all larger than those from 2D cine SSFP. This bias was <6% except for right ventricular end-systolic volume which was 12%. The 3D cine images had a lower ventricular blood-to-myocardium contrast ratio, contrast-to-noise ratio, mass, and subjective quality score.

Conclusion—The novel prospective respiratory motion compensation method for 3D cine SSFP imaging was robust and efficient, and yielded slightly larger ventricular volumes, and lower mass compared to breath-hold 2D cine imaging.

Keywords

Free-breathing; whole-heart; 3D cine; ventricular function; respiratory motion correction; self-navigator

INTRODUCTION

A 2-dimension (2D) cine steady-state free precession (SSFP) sequence is the established method in cardiovascular magnetic resonance imaging (MRI) for the assessment of heart motion and quantitation of ventricular volumes and ejection fraction (1). Conventionally, 2–4 ventricular long-axis slices and a stack of 10–15 ventricular short-axis slices are precisely

planned to encompass both ventricles. One or two slices at a time are acquired while the patient holds their breath for 5–15 seconds to minimize respiratory motion artifact. This 2D breath-hold approach, however, has several widely-recognized drawbacks. The through-plane resolution is inferior to the in-plane resolution and, thus, stacks of contiguous slices are poorly suited for reformatting in other planes. Hence, there is a need to image the ventricles in multiple orientations, and with this comes a requirement for careful slice location planning by expert operators. Moreover, breath-holding may not be feasible in ill or younger patients (2). Even patients who can hold their breath may have difficulty achieving a consistent amplitude of breath-holding leading to slice-to-slice misregistration and errors in ventricular volume measurements (3).

An isotropic resolution non-angulated 3-dimensional (3D) cine SSFP sequence addresses the drawbacks of 2D slices as it is easy to plan and, once acquired, can be reformatted in multiple appropriate planes for interpretation and quantification. Acceleration techniques, such as parallel imaging (4,5) and compressed sensing (6), have been used to acquire 3D cine SSFP dataset in a single breath-hold (7–15). However, shortening the acquisition time to a breath-hold also requires sacrificing spatial and temporal resolution beyond that which is optimal for accurate measurements of left and right ventricular parameters (16) and, thus, this approach has not become the clinical routine. Free-breathing approaches for high spatiotemporal resolution 3D cine acquisition have been developed (16–22) but suffer from various drawbacks. These include incomplete coverage of the cardiac cycle, non-linear k-space profiles which preclude fast in-line image reconstruction and use of conventional acceleration techniques, or reliance on an external respiratory bellows belt which limits the effectiveness of respiratory motion compensation.

In this study, we report the development and implementation of a respiratory motion compensation technique that prospectively gates and tracks heart position for free-breathing, high spatial and temporal resolution 3D cine SSFP imaging across the entire cardiac cycle. Parallel imaging with sensitivity encoding (SENSE) (4) is used to shorten the acquisition time so that it is suitable for the clinical environment. The 3D cine and conventional 2D cine SSFP acquisitions are compared with regard to blood-to-myocardium contrast ratio, contrast-to-noise ratio, subjective image quality, and ventricular volumes and mass.

METHODS

Whole-heart 3D Cine SSFP Sequence with Heart-NAV

A diagram of the whole-heart 3D cine SSFP sequence is shown in Figure 1. A 3D cine SSFP sequence is prescribed in a sagittal orientation to encompass the ventricles with frequency encoding in the superior-inferior direction. A centric profile ordering is used to fill k-space. In this scheme, the same portion of k-space is sampled throughout a given cardiac cycle. K-space filling begins in the center on the first cycle and then progressively moves out from the center with subsequent cycles as shown in Figure 2. The Heart-NAV signal described below is acquired periodically at the center of k-space along the readout (k_x) direction.

Standard retrospective cardiac gating is employed using a vectorcardiogram signal. The prospective respiratory motion compensation technique is based on our previously reported

Heart-NAV approach (23). For this application, at the beginning of every 5th phase (i.e., segment) of the cardiac cycle, one excitation is modified so that the phase-encoding gradient is turned off and the centerline of k-space along the superior-inferior direction is read yielding the “HEART-NAV” signal. This signal is collected and transformed from the Fourier domain to the image domain, and represents the 1-dimensional projection line of the 3D SSFP imaging volume in the superior-inferior orientation. The Heart-NAV image data is processed and displayed using the scanner’s diaphragm navigator analysis pathway. This includes a cross-correlation analysis with the preceding Heart-NAV line to measure displacement in the superior-inferior direction (Figure 3). Specifically, the first acquired projection line is set as a reference ($P_0(x)$), where P_0 represents the pixel intensity at position x along the superior-inferior direction. On the projection line, a region of interest or kernel is defined to include the projection of the heart ($K(x)$). The kernel location is based on the position of the conventional diaphragmatic navigator during the prescription of the 3D cine SSFP sequence and has a fixed length (w) of 80 mm. The next acquired projection line $P_i(x)$ is overlaid on the kernel ($K(x)$) and shifted Δx pixels to maximize the cross-correlation function (CC) as follows:

$$CC(\Delta x) = \frac{\sum_w (P_i(x - \Delta x) - \text{mean}(P_i(x - \Delta x))) \times (K(x) - \text{mean}(K(x)))}{\sqrt{\sum_w (P_i(x - \Delta x) - \text{mean}(P_i(x - \Delta x)))^2 \times \sum_w (K(x) - \text{mean}(K(x)))^2}}$$

The value of Δx which maximized the cross-correlation ($CC(\Delta x)$) is then chosen as the respiratory-induced heart displacement at the i^{th} cardiac phase and used for prospective gating and tracking.

As with the conventional diaphragm navigator, the first few beats at the beginning of the scan (≈ 10 seconds) are automatically used to monitor the navigator signal, identify its most superior position (i.e., end-expiration), and set this position as the reference for the imaging phase. An acceptance window at end-expiration is then established using this position and a width that is pre-specified by the user. Subsequently, during the imaging phase, if all Heart-NAV locations in a cardiac cycle are within the acceptance window, the acquired 3D cine data for that entire cycle is accepted for reconstruction; otherwise, it is re-acquired in the next cycle. Also, during the imaging phase, prospective tracking is performed by moving the imaging volume by Δx , where Δx is the distance from the reference projection line as determined by Heart-NAV. This Δx is used to adjust the imaging volume position for the 5 subsequent cardiac phases until the next Heart-NAV is acquired.

Note that because the Heart-NAV radiofrequency pulse excites the same volume as the pulses used for cine SSFP data, the equilibrium state of the net magnetization vector is preserved and flash artifact is avoided. Moreover, for a 3D cine SSFP sequence with 20 acquired heart phases per cardiac cycle, there will be 4 Heart-NAVs per cycle. With each Heart-NAV lasting 1 repetition time or about 3 ms, this leads to a total of 12 ms per cycle (spread out over the cycle) devoted to respiratory motion compensation. At a heart rate of 80 bpm or cycle duration of 750 ms, this amounts to 1.6% of the cycle.

Subjects

To evaluate the whole-heart 3D cine SSFP sequence with Heart-NAV respiratory motion compensation in clinical practice and compare it to the conventional 2D cine SSFP acquisition, we performed a prospective study. Subjects were eligible if they were referred for a cardiovascular MRI examination with administration of a contrast agent and did not require sedation. The Boston Children's Hospital Committee on Clinical Investigation approved this study, and written informed consent was obtained from all subjects.

Cardiovascular MRI protocol

MRI examinations were performed with a 1.5T Achieva dStream scanner (Philips Healthcare, Best, the Netherlands) and vectorcardiogram gating. In each subject, a conventional 2D cine SSFP sequence was used to acquire a stack of 12 slices prescribed in a ventricular short-axis plane to completely encompass the left and right ventricles. The imaging was performed with breath-holding at end-expiration for 10–12 seconds to acquire 2 slices. The acquisition parameters were as follows: field-of-view 260×260 mm, in-plane resolution 1.8×1.8 mm reconstructed to 1.25×1.25 mm, slice thickness 8 mm, slice gap 0–2 mm, flip angle 60°, echo time 1.4 ms, repetition time 2.8 ms, bandwidth 1.1 kHz, heart phases 20 interpolated to 30, and SENSE factor 2. Three to 10 minutes after the administration of 0.15 mmol/kg gadobutrol contrast, a free-breathing 3D cine SSFP sequence with Heart-NAV for respiratory motion compensation was acquired in a sagittal plane with the following parameters: field-of-view 512×250×180 mm, isotropic resolution 2.0 mm³, flip angle 60°, echo time 1.5 ms, repetition time 3.0 ms, bandwidth 1.7 kHz, heart phases 20 interpolated to 30, respiratory acceptance window 7 mm, tracking factor 1, and SENSE factor 3 (2 anterior-posterior direction, 1.5 right-to-left direction). We used an in-line coil compression technique (24) so that the amount of data was small enough for the scanner to complete the acquisition. The scan time of the 3D cine SSFP sequence was measured prospectively with a stopwatch.

Image Analysis

The 2D and 3D cine images were reconstructed in-line on the scanner and then transferred to a workstation where they were analyzed using commercially available software (CVI⁴², Circle Cardiovascular Imaging, Calgary, Canada). The 3D cine images were reformatted into a short-axis, 2-chamber, 3-chamber and 4-chamber views. The orientation, number, thickness, and interslice gap of reformatted short-axis slices were matched to the 2D cine short-axis parameters for each subject. The right and left ventricular myocardial boundaries on the short-axis 2D cine and reformatted 3D cine images were each delineated by 2 independent observers who were blinded to the numeric results. Left and right ventricular end-diastolic volume (EDV), end-systolic volume (ESV), stroke volume (SV), and ejection fraction (EF), as well as left ventricular end-diastolic mass were calculated using a standard summation of disks approach. Papillary muscles of the left ventricle were included in the mass.

The ventricular blood-to-myocardium contrast ratio (CR) (11) and contrast-to-noise ratio (CNR) (25) were calculated as follows:

$$CR = \frac{\bar{S}_{blood}}{\bar{S}_{myocardium}}, \quad CNR = \frac{2 \times (\bar{S}_{blood} - \bar{S}_{myocardium})}{(Noise_{blood} + Noise_{myocardium})},$$

where $\bar{S}$ is the mean of signal intensity and $Noise$ is the standard deviation of the signal within the specified region of interest (ROI). The ROI for blood was positioned in the center of the left ventricular cavity at end-diastole, and the ROI for myocardium was positioned in the ventricular septum at the mid-ventricular level at end-diastole. Because SENSE was used for image reconstruction, the noise in the CNR formula was calculated as the standard deviation of the signal intensity in the respective ROIs (8). Finally, the sharpness of the ventricular borders on the short-axis images was subjectively scored by a physician experienced in cardiac MRI according to a 4-point scale: 1) poor image quality, borders cannot be traced; 2) fair image quality, borders are significantly blurred; 3) good image quality, ventricular borders are slightly blurred; and 4) excellent image quality, ventricular borders are sharp (26).

Statistical Analysis

Descriptive statistics are reported as median and range, or mean $\pm$ standard deviation, as appropriate. The Wilcoxon signed-rank test was used to compare the image quality scores, and a paired two-tailed Student's t-test was used to compare the CR, and CNR; a p-value 0.05 was considered statistically significant. Bland-Altman analysis was used to assess agreement (27). The mean of the differences (3D-2D) and mean of the differences expressed as a percentage ($\frac{3D - 2D}{0.5 \times (3D + 2D)} \times 100$) were calculated.

RESULTS

Subjects

Seventeen patients (9 male) were enrolled in the study and all completed the protocol. The median age was 26 years (range 9–70), the median weight was 69 kg (range 20–107), and the mean heart rate during the MRI scan was of 76 ± 12 bpm. Their principal diagnoses were as follows: aortic valve stenosis and/or regurgitation (n=4), repaired tetralogy of Fallot (n=4), atrial septal defect (n=2), cardiomyopathy (n=2), repaired transposition of the great arteries (n=2), connective tissue disorder (n=1), repaired double-outlet right ventricle (n=1), and Ebstein anomaly of the tricuspid valve (n=1).

Image Quality and Scan Time

Representative free-breathing 3D cine SSFP images are shown in Figures 4, 5, and Supporting Video S1. Mid-ventricular short-axis slices in diastole using breath-hold 2D cine SSFP and free-breathing 3D cine SSFP in 3 patients are compared in Figure 6. Minimal ghosting artifact and no flashing artifact were observed. The subjective image quality score, CR, and CNR for 2D and 3D cine SSFP are compared in Table 1. The 3D cine SSFP images had lower mean image quality scores. The myocardium signal intensity for the 2D acquisition was lower leading to a significantly better ventricular blood-to-myocardium CR and CNR. Scan time of free-breathing 3D cine sequence was 5.9 ± 2.7 minutes.

Ventricular Measurements

Left and right ventricular measurements for breath-hold 2D cine and free-breathing 3D cine acquisitions are compared in Table 2 and Figure 7. Of note, 3D cine measurements of EDV, ESV, and SV for both ventricles were all larger (systematic bias); however, the differences were less than 6% except for the right ventricular ESV. The 3D cine measurements of left ventricular end-diastolic mass were smaller with a mean difference of 6.9%. Interobserver agreement for left and right ventricular measurements is reported in Table 3.

DISCUSSION

We developed and evaluated in patients a new prospective respiratory motion compensation method (Heart-NAV) for a free-breathing whole-heart 3D cine SSFP sequence. Heart-NAV leverages the scanner's existing diaphragmatic navigator technology, preserves the equilibrium state of the net magnetization vector preventing flash artifact, is compatible with retrospective cardiac gating, and occupies only 1.6% of cardiac cycle. In a prospective evaluation in patients, all 3D cine SSFP acquisitions were successful and the mean scan time was 5.9 minutes. Ventricular volumes from 3D cine SSFP were slightly larger and left ventricular mass was slightly smaller than those from a conventional breath-hold 2D cine acquisition. The image quality score, and the blood-to-myocardium CR and CNR were also lower compared to 2D cine images.

Left and right ventricular volumes were larger and left ventricular mass was smaller by 3D cine SSFP than those by 2D cine SSFP. The biases were all <7% except for right ventricular ESV which was 12%, and all biases were in the range observed for inter-scan variability in patients with congenital heart disease (28). The temporal resolution of the 3D and 2D cine data was the same, and reformatting of the 3D data was performed to match the orientation and slice number of the 2D data. However, the 3D data had a lower in-plane spatial resolution and was acquired after a contrast agent was administered. These differences have been associated with higher ventricular volume and lower left ventricular mass measurements as we found (29,30). In addition, although the 2D and 3D image datasets were both acquired at end-expiration, it is likely that the different breathing patterns (i.e., breath-holding versus free-breathing) contributed to the bias in volume calculations. Compared with free-breathing, breath-holding increases intrathoracic pressure which, in turn, decreases venous return from the inferior vena cava (31). These changes cause a reduction in right and left ventricular volumes as observed in our study and others (32,33).

Several factors influenced the signal intensity of blood and myocardium in our study. In the 2D cine acquisition, blood flow through the imaging plane contributes to the bright signal of the ventricular blood pool (34). In the 3D cine acquisition, the larger volume of excitation leads to saturation of inflowing blood, a darker blood pool, and inferior image quality (34). Thus, similar to other investigators (17,18,21), we chose to mitigate the saturation effect by performing 3D cine SSFP imaging after administering a T1-shortening intravenous contrast agent. Although this increases the blood pool signal and the quality of 3D cine SSFP imaging, the contrast agent also perfuses into the myocardium and increases its signal as well. This caused a lower blood-to-myocardium CR and CNR for the 3D cine images

compared to the 2D cine images, which were performed before contrast agent administration.

Compared to the short-axis 2D cine SSFP acquisition, the reformatted 3D cine SSFP images had a worse subjective image quality score. The lower blood-to-myocardium CR and CNR in the 3D cine images likely contributed to this finding. In addition, although acquired at a spatial resolution of 2.0 mm^3 , the 3D images were reformatted into a double oblique short-axis view which can increase the in-plane spatial resolution up to approximately 3.5 mm ($2.0 \text{ mm} \times \sqrt{3}$) compared with 1.8 mm for the 2D images. Lastly, incomplete respiratory motion compensation may have also contributed to the lower quality score. Despite the lower 3D image quality, interobserver agreement for ventricular measurements was similar to the range seen with the 2D images.

Multiple approaches have been described to compensate for the respiratory-induced heart motion during free-breathing whole-heart 3D cine acquisitions. Uribe *et al.* added a startup pulse at the beginning of each cardiac phase (segment) to read the centerline of k-space in real-time rather than using an excitation pulse as we did (16). They used the centerline reconstruction in conjunction with a reference projection acquired during a 4-second breath-hold preparatory scan to gate to end-expiration. One drawback to this approach is that the sum of all the startup pulses occupied nearly 6% of cardiac cycle, which may impact the accuracy of motion assessment and measurements. To address this shortcoming, Henningsson *et al.* used a respiratory bellows belt around the chest or abdomen to perform respiratory gating without occupying any portion of cardiac cycle or interrupting the acquisition (17). However, chest and abdominal wall position do not closely match heart position and, thus, its effectiveness is limited (35). To improve motion compensation, other investigators utilized a 3D radial k-space trajectory so as to traverse the central part of k-space during each readout (18–20). The central part of k-space was used to retrospectively sort the image data into respiratory and cardiac phases. A 3D dataset was then generated for each respiratory and cardiac phase using a complex non-Cartesian SENSE or compressed sensing reconstruction algorithm. To reduce the complexity of the non-Cartesian image reconstruction, Han *et al.* took a similar approach but used a spiral-like k-space trajectory on Cartesian grids that traverses the central part of k-space during each readout (21). Usman *et al.* also used the same spiral-like trajectory and Cartesian grid strategy (22); however, rather than sorting the k-space data into different respiratory phases, they used a soft-weighting technique to weight and combine the k-space data acquired at different phases of the respiratory cycle (36).

Compared to these techniques, our Heart-NAV approach for free-breathing 3D cine SSFP acquisition has several strengths. Although it is compatible with non-linear k-space profile orderings, the current version uses a conventional Cartesian k-space trajectory that makes fast in-line image reconstruction on the scanner possible and allows for the application of existing parallel imaging techniques. These features are important because they facilitate implementation in the clinical environment and dissemination to other sites to explore the utility and robustness of 3D cine imaging. In contrast to the external respiratory bellows belt approach, Heart-NAV tracks the heart position and prospectively adjusts the imaging plane according to position within the acceptance window. Lastly, by using only 12 ms per

heartbeat for respiratory motion compensation, the Heart-NAV approach temporally samples nearly the full cardiac cycle.

Based on this preliminary experience, we are encouraged by the image quality, robustness, and clinical practicality of free-breathing whole-heart 3D cine acquisitions using Heart-NAV. It is easy to plan, has an acceptable scan time, eliminates the drawbacks of breath-holding, and provides the reformatting advantages of a volume dataset. Looking forward, the Heart-NAV technique is amenable to further refinements. In the current implementation, the Heart-NAV signal is received by all of the phased array coil elements and thus includes static signal from the chest and spine adjacent to the heart. The fidelity of the Heart-NAV signal could be improved by using an anterior phased array coil element which is closer to the heart to receive the signal and thereby suppress the surrounding static signal (37). Furthermore, the scan time of 3D cine acquisition can be shortened by using compressed sensing image reconstruction (38) and respiratory biofeedback (39). Lastly, note that for respiratory gating, cine SSFP data is accepted on a per cardiac cycle basis—if all of the Heart-NAV signals in a cycle are in the acceptance window, then the cine data from the whole cycle is accepted. It would be more time efficient to allow data to be accepted from a portion of the cycle if the adjacent Heart-NAV was in the acceptance window. The reduction in scan time could be traded for further improvements in spatial or temporal resolution.

Our study has several limitations. The number of subjects was relatively small, and it is thus possible that additional differences between the free-breathing 3D and breath-hold 2D acquisitions might emerge with a larger sample size. Patients who were anesthetized were excluded from this study because this would have prolonged their examination time. However, we would expect that Heart-NAV would perform effectively under these conditions since anesthetized patient usually have a regular breathing pattern and heart rate. We have not yet systemically optimized the 3D cine SSFP parameters. A 2 mm³ spatial resolution was chosen because it provided an acceptable balance between diagnostic image quality and scan time in preliminary tests. Application of higher order acceleration should allow us to comfortably increase the resolution. Similarly, we chose to sample heart position with Heart-NAV 4 times per cardiac cycle because less frequently would reduce the effectiveness of respiratory motion compensation and more frequently would decrease the temporal sampling of the cardiac cycle and risk inducing eddy current artifact.

CONCLUSION

We developed a new prospective respiratory motion compensation algorithm, Heart-NAV, for free-breathing whole-heart 3D cine SSFP imaging. Ventricular volume measurements using this technique were slightly larger than those obtained with a conventional breath-hold 2D cine SSFP acquisition, left ventricular mass was marginally smaller, and scan time was within a clinically acceptable range. Such 3D cine acquisitions eliminate the need for breath-holding, simplify scanning, and enable volume-based reformatting and analysis.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

Diagram of the Heart-NAV technique to prospectively gate and track the respiratory motion of the heart during a whole-heart 3D cine SSFP sequence.

Figure 2.
K-space profile ordering for the whole-heart 3D cine SSFP sequence.

Figure 3. Trace of respiratory induced-heart motion measured by Heart-NAV during a 3D cine SSFP acquisition. Solid blue lines: acceptance window (7 mm). Solid vertical red line: transition from Heart-NAV training phase to acquisition phase. Horizontal red lines: estimated heart location. Lower green lines: accepted 3D cine SSFP data.

Figure 4. Whole-heart 3D cine SSFP images at end-diastole and end-systole in a 22-year-old male patient with connective tissue disorder.

Figure 5.
Whole-heart 3D cine SSFP images at end-diastole and end-systole in a 24-year-old female patient with tetralogy of Fallot who had undergone pulmonary valve replacement.

Figure 6. Mid-ventricular short-axis images in diastole acquired with breath-hold 2D cine SSFP and a free-breathing 3D cine SSFP with Heart-NAV from 3 patients.

Figure 7.

Bland-Altman plots of agreement comparing left and right ventricular (LV, RV) measurements for free-breathing 3D cine SSFP and conventional 2D cine SSFP. The solid line indicates the mean difference (bias) and the dashed lines show ± 1.96 standard deviations of difference.

Table 1

Subjective image quality score, blood-to-myocardium contrast ratio (CR), and blood-to-myocardium contrast-to-noise ratio (CNR) for breath-hold 2D cine SSFP and free-breathing 3D cine SSFP acquisitions (n=17).

	Image quality score		CR	CNR
	Left ventricle	Right ventricle		
2D cine SSFP	3.7 ± 0.5	3.6 ± 0.5	5.3 ± 1.3	29.1 ± 7.8
3D cine SSFP	3.1 ± 0.5	2.6 ± 0.6	3.1 ± 0.7	19.9 ± 8.4
P-value	<0.001	<0.001	<0.001	0.010

Values are mean ± standard deviation.

Table 2

Ventricular measurements for the breath-hold 2D cine SSFP and free-breathing 3D cine SSFP sequences (n=17).

	Left ventricle					Right ventricle				
	EDV (ml)	ESV (ml)	SV (ml)	EF (%)	Mass (g)	EDV (ml)	ESV (ml)	SV (ml)	EF (%)	
2D cine SSFP	168.6±56.9	76.2±31.7	92.3±28.8	55.4±6.6	117.1±64.7	202.0±64.2	96.1±38.5	105.9±32.1	53.2±8.2	
3D cine SSFP	178.1±62.8	80.2±34.8	97.9±31.1	55.8±5.8	109.8±61.9	212.9±66.1	106.2±37.1	109.7±32.0	51.1±7.9	
Mean difference (3D-2D)	9.5±10.6	4.0±8.5	5.5±9.2	0.5±4.4	-7.3±8.1	10.9±17.9	10.1±10.2	3.8±9.8	-2.1±3.3	
Mean % difference (3D-2D)	5.1±4.9%	4.3±7.4%	6.0±9.4%	1.0±7.8%	-6.9±7.6	5.3±11.2%	11.7±11.9%	3.9±9.9%	-2.0±3.0%	
P-value	0.002	0.071	0.024	0.670	0.002	0.023	0.001	0.135	0.016	

Values are mean ± standard deviation.

EDV, end-diastolic volume; EF, ejection fraction; ESV, end-systolic volume; and SV, stroke volume.

Table 3

Interobserver agreement for the ventricular measurements from the breath-hold 2D cine SSFP and free-breathing 3D cine SSFP sequences (n=17).

	2D					3D				
	EDV (ml)	ESV (ml)	SV (ml)	EF (%)	Mass (g)	EDV (ml)	ESV (ml)	SV (ml)	EF (%)	Mass (g)
<i>Left ventricle</i>										
Observer 1 (O1)	168.6±56.9	76.2±31.7	92.3±28.8	55.4±6.6	117.1±64.7	178.1±62.8	80.2±34.8	97.9±31.1	55.8±5.8	109.8±61.9
Observer 2 (O2)	162.9±56.9	72.0±31.3	90.9±29.6	56.5±6.5	127.5±69.1	178.5±60.8	78.1±32.9	100.4±30.8	57.1±5.8	125.1±70.9
Mean difference (O1–O2)	5.7±4.4	4.3±4.4	1.4±6.5	-1.1±2.7	-10.4±9.6	-0.4±5.3	2.1±3.0	-2.5±4.6	-1.2±1.4	-15.3±13.5
Mean % difference (O1–O2)	3.6±2.7%	6.2±5.5%	1.6±6.6%	-2.1±4.7%	-8.7±9.3	-0.6±3.0%	2.4±3.0%	-2.9±4.8%	-2.2±2.5%	-12.3±10.3
P-value	<0.001	0.001	0.381	0.102	<0.001	0.755	0.008	0.036	0.002	<0.001
<i>Right ventricle</i>										
Observer 1 (O1)	202.0±64.2	96.1±38.5	105.9±32.1	53.2±8.2		213.0±66.1	106.2±37.1	106.7±37.0	50.1±8.9	
Observer 2 (O2)	202.2±64.5	98.3±36.8	103.9±32.6	52.0±7.0		219.3±66.7	111.0±35.6	108.3±36.7	49.3±6.6	
Mean difference (O1–O2)	-0.2±3.3	-2.2±6.6	2.0±6.5	1.2±2.9		-6.3±9.4	-4.8±5.8	-1.6±9.7	0.7±3.9	
Mean % difference (O1–O2)	-0.1±1.8%	-3.0±6.4%	1.9±6.1%	2.0±6.1%		-3.4±5.5%	-5.5±6.1%	-2.6±13.2%	0.8±8.8%	
P-value	0.828	0.197	0.227	0.101		0.013	0.004	0.515	0.456	

Values are mean ± standard deviation.

EDV, end-diastolic volume; EF, ejection fraction; ESV, end-systolic volume; and SV, stroke volume.