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3D Breath-Held Cardiac Function With Projection Reconstruction in Steady State Free Precession Validated Using 2D Cine MRI

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Abstract

Purpose: To develop and validate a three-dimensional (3D) single breath-hold, projection reconstruction (PR), balanced steady state free precession (SSFP) method for cardiac function evaluation against a two-dimensional (2D) multislice Fourier (Cartesian) transform (FT) SSFP method.

Materials and Methods: The 3D PR SSFP sequence used projections in the x-y plane and partitions in z, providing 70–80 msec temporal resolution and $1.7 \times 1.7 \times 8$ –10 mm in a 24-heartbeat breath hold. A total of 10 volunteers were imaged with both methods, and the measurements of global cardiac function were compared.

Results: Mean signal-to-noise ratios (SNRs) for the blood and myocardium were 114 and 42 (2D) and 59 and 21 (3D). Bland-Altman analysis comparing the 2D and 3D ejection fraction (EF), left ventricular end diastolic volume (LVEDV) and end systolic volume (LVESV), and end diastolic myocardial mass (LVEDM) provided values of bias ± 2 SD of $0.6\% \pm 7.7\%$ for LVEF, $5.9 \text{ mL} \pm 20 \text{ mL}$ for LVEDV, $-2.8 \text{ mL} \pm 12 \text{ mL}$ for LVESV, and $-0.61 \text{ g} \pm 13 \text{ g}$ for LVEDM. 3D interobserver variability was greater than 2D for LVEDM and LVESV.

Conclusion: In a single breath hold, the 3D PR method provides comparable information to the standard 2D FT method, which employs 10–12 breath holds.

Keywords

radial imaging; projection reconstruction; cardiac function; validation; 3D SSFP; left ventricle; global function; magnetic resonance imaging

Two-dimensional fourier transform (FT) (Cartesian) breath-held electrocardiogram (ECG)-gated cardiac function imaging with MRI is an accurate and reliable method for cardiac function evaluation (1,2). However, many breath holds are required to obtain multiple two-dimensional (2D) slices covering the entire apical–basal extent of the heart. Three-dimensional (3D) methods could improve patient examination by reducing the 10–12 breath holds required for complete assessment of cardiac function to a single breath hold, thereby decreasing exam time, improving patient tolerance, and simplifying the cardiac exam. Furthermore, slice-

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misregistration caused by variability in breath-holding may lead to errors in measurement of left ventricular (LV) volumes; this is not a concern for 3D breath-hold imaging.

3D multiphase imaging of the heart has recently been achieved using contrast agents (3,4). More recently, preliminary studies have investigated the use of 3D balanced steady state free precession (SSFP) imaging of LV function using short TRs (5), the variable sampling in time (VAST) technique (6), sensitivity encoding (SENSE) (7), broad-use linear acquisition speed-up technique (*k-t* BLAST) (8), and using our approach of undersampled projection reconstruction (PR) (radial imaging) (9). Due to the requirements of short scan times for breath-hold cardiac imaging, these techniques employed unique k-space acquisition strategies, provided low spatial and/or temporal resolution, and required long breath-holds, sometimes with limited slab coverage. For these reasons, and due to lack of validation studies, the methods are not yet routinely used for clinical cardiac MR exams. Undersampled PR is promising since it provides images with spatial resolution that is roughly independent of the number of projections (and therefore scan time), although streak artifacts appear when the number of projections is reduced (10). Approximately 32 to 64 projections are required for relatively artifact-free cardiac images (11-14). Using 3D PR with projections in the x-y plane and partitions in the z direction, breath-held, segmented, ECG-gated SSFP acquisitions can provide whole heart coverage in a feasible breath hold, good through-plane spatial resolution, a temporal resolution of 70–80 msec, and higher in-plane spatial resolution than has been achieved with other 3D single breath-hold approaches.

The goal of this study was to validate the 3D PR single breath hold cardiac function technique as a suitable alternative to standard multi-breath hold 2D Fourier transform (FT) techniques for characterizing LV ejection fraction (LVEF), end diastolic volume (LVEDV), end systolic volume (LVESV), and myocardial mass (LVEDM).

MATERIALS AND METHODS

A total of 10 healthy subjects were imaged, five male and five female, with an average age of 30 ± 8 years, average weight of 67 ± 13 kg, and an average heart rate of 60 ± 9 beats per minute. All subjects provided informed consent consistent with the Internal Review Board of our institution. The volunteers were imaged with the 2D FT SSFP technique and the 3D PR SSFP technique.

Imaging was performed with a GE 1.5-T Excite scanner. A commercially provided 3D SSFP sequence was modified to collect projections in the x-y plane and partitions in the slice dimension, as shown in Fig. 1. Modifications provided for a prospective ECG-gated, segmented, multiphase acquisition using projections. Scan parameters for the 3D PR were: 192 readout points (N_r) (full echoes) $\times$ 36–48 projections (N_p) $\times$ 12–14 N_z , flip = 45°, TR/TE = 3.4/1.5 msec, ± 125 kHz receiver bandwidth, 8–10 mm slices zero-filled to 4–5 mm, FOV = 32 cm, 21–24 views per segment, 71–82 msec temporal resolution, 24 heartbeat breath-hold, 12 cardiac phases, and no view-sharing. Zero-filling was used for reconstructing the 3D acquisitions because it allows better visualization of the acquired slice resolution (15).

Interleaved projection sets were acquired in alternate cardiac phases as in the UNFOLD or reduced field of view (rFOV) methods (14,16,17). In the rFOV method, two interleaved sets of projections are acquired, e.g., 72 projections are obtained, split into two interleaved sets of 36 N_p each. The first or second interleaved set (36 N_p each) are obtained in alternate cardiac phases. After reconstruction of the time series, additional artifact suppression is obtained through low-pass temporal filtering, which removes any artifacts that “flickers” in time at the Nyquist (18).

The undersampling before rFOV processing was roughly 5 (192 Nr χ 36 Np). The trajectory was centered so that each projection crossed the center of k-space at the echo time, using a recently described method (19). Reconstruction of the 3D PR data was performed offline using gridding (20) followed by fast Fourier transform (FFT; www.fftw.org), and rFOV processing.

Scan parameters for the 2D FT technique were: 192×160 encoding matrix, FOV = 32×24 cm, flip = 45° , TR/TE = 3.4/1.5 msec, receiver bandwidth = ± 125 kHz, 8-mm slices, 3-mm slice gaps, ECG-gating, 12 views-per-segment, 40-msec temporal resolution, full R-R coverage, with arrhythmia rejection and retrospective ECG-gating, and a cine view-sharing technique to reconstruct 30 phases. The slices were acquired using 12 heartbeat breath holds, acquired in 10–12 separate breath holds.

On all datasets (2D FT slices and the corresponding 3D PR slices for each volunteer), the signal-to-noise ratio (SNR) of blood and myocardium, and the contrast to noise ratio (CNR) between blood and myocardium, and between myocardium and lung, were measured. Noise measures on undersampled PR images do not represent thermal noise only; the artifacts present in the image will contribute to the root-mean-square-deviation (RMSD) of the background, providing an effective noise measure that includes artifact. Care was taken to measure noise in the PR images in a background region without discrete streak artifacts.

For quantitative analysis, the end systolic frame was defined as the frame with the smallest LV cavity area. The end diastolic frame was defined as the first frame directly after the QRS complex. The most basal slice for both systole and diastole was chosen based on the presence of myocardium in that frame. An experienced cardiologist drew regions of interest (ROI) around the epicardial and endocardial surfaces (excluding trabeculae and papillary muscles) on each slice, using commercially provided software (Cine Tool; General Electric, Waukesha, WI). These tracings allowed automatic calculation of LVEF, LVEDV, LVESV, and LVEDM.

For all individual measurements ($N = 10$) using both the 2D FT and the 3D PR technique, the means and standard deviations (SDs) were calculated. Statistical differences between the 2D and 3D methods were assessed by a Student's paired t-test. The cutoff for statistical significance was $P < 0.05$. The limits of agreement between the two techniques were estimated as the mean difference between the techniques ± 2 SD of the differences, using the Bland-Altman analysis (21). A second experienced reader also traced the LV contours in the 10 volunteers to allow measurement of interobserver variability. For each technique (2D and 3D), the interobserver variability was assessed by calculating the correlation (R^2) between the two sets of measurements. The variability, defined as the SD of the differences between corresponding observations, divided by the average value (in percent), and the mean and SD of differences between corresponding observations, was also calculated.

RESULTS

Figure 2 compares slices from the 3D-PR and 2D FT acquisitions, and the volume-rendered stack of slices for each acquisition method. The 2D data required 12 breath holds, while the 3D data set required a single breath hold. Note the slice misregistration in the 2D images, both in the septum and the anterior papillary muscle. The completeness of coverage is demonstrated by the tapering of the septal wall at the base.

Table 1 summarizes the average SNR and CNR for the 2D SSFP and the 3D PR SSFP techniques. The 2D method provided roughly two times greater SNR of the blood and the myocardium compared with the 3D method. The myocardium-lung CNR for the 3D PR SSFP method was especially low.

Table 2 shows the results of the quantitative measures of LV volumes obtained with the 3D PR SSFP technique and the 2D FT SSFP standard. The mean values were similar for LVEF, LVEDV, LVESV, and LVEDM. The 3D PR SSFP technique overestimated the LVEDM by 1%, the LVEDV by 3.5%, and the LVESV by 4.5%; it underestimated LVEF by 1.6%. However, these differences were not significant, as evidenced by the *P*-values. The Bland-Altman analysis of the agreement between the two techniques, reported as the mean difference between paired measurements ± 2 SD of the differences, is also shown in Table 2.

The correlation between 2D FT and 3D PR LVEDV, LVESV, and LVEDM measures are exhibited in Fig. 3a. Linear regression with all data produces a slope of 1.05 with an intercept of -2.0 mL. The R^2 value for all measures of LV volume was 0.98. Figure 3b shows the Bland-Altman plot for LV volumes, plotting mean volumes as measured by both techniques against the difference in volumes. Figure 3c shows the Bland-Altman plot for LV end diastolic mass, plotting mean mass as measured by both techniques against the difference in mass. The results for LV end systolic myocardial mass were very similar (not shown).

Table 3 shows the results of interobserver variability (see Materials and Methods). The 3D measurements have greater interobserver variability than the 2D measurements for LVESV and LVEDM. The 2D interobserver variability is similar to that obtained in other studies (2, 22).

DISCUSSION

Our experience during these studies is that complete volumetric coverage in a single breath hold has the advantage of simplicity and efficiency compared to obtaining data in approximately 12 breath holds as required for 2D FT SSFP. The images were of sufficient quality to allow for both qualitative and quantitative assessment of ventricular function (Fig. 2). The LVEF, LVEDV, LVESV, and LVEDM values obtained by the 3D PR SSFP method were accurate, as evidenced by their good agreement with the previously validated 2D FT technique. However, it might be valuable to further validate the technique in a patient cohort with a range of heart rates and morphologies.

For PR, the in-plane resolution is more strongly limited by SNR than by scan time constraints, since high and isotropic resolution can be obtained by extending the readout resolution. Because of this property, the resolution achieved in the present study (1.7×1.7 mm) is higher than reported resolutions of other 3D single breath-hold methods, e.g., 2×2 mm (8), 2.2×4.4 mm (7), 1.5×3.8 (5), and 1.6×2.25 (23). This high in-plane resolution provides increased accuracy of tracing the LV on short-axis slices, although temporal resolution may be more important. The 3D PR method provided speed increases by a factor of roughly five, through a combination of undersampling and rFOV. rFOV alone contributed to accelerate the imaging by a factor of roughly 1.4–2.0, by suppressing artifacts and allowing greater undersampling. The other reported 3D methods had comparable temporal resolutions and breath-hold times, except the *k-t* BLAST approach, which provided a better combination of temporal resolution and breath-hold time ($\approx 20\%$ less scan time) and thinner slices (8). However, for the VAST (23) and *k-t* BLAST (24) methods, the cine images have images have spatial frequency dependent temporal resolution.

Our study sought to validate the 3D single breath hold method using the 2D method as the existing standard. Therefore, the scan parameters for both methods were optimized individually, and no attempt was made to make them identical. For example, UNFOLD methods would provide speed advantages for the 2D method (25), just as for 3D PR, but this was not employed for the 2D, since the parameters were taken from the 2D protocol that is used clinically for cardiac function at our institution.

The factor of two difference in SNR (Table 1) that was measured between the 2D and 3D techniques for both blood and myocardium was higher than expected. Sources of SNR difference are the differences between PR and Fourier sampling efficiency, differences in filtering, as well as the differences between voxel size (the nominal voxel size was roughly equal in this study), the square root of scan time (i.e., $TR \times Ny = 410$ msec for the 2D images, $TR \times Np \times Nz \approx 1710$ msec for the 3D images), and the influence of undersampling artifacts on the apparent noise in PR images. Besides undersampling artifacts, streak artifacts due to the influence of off-resonance and flow on the SSFP steady state were observed. 3D balanced SSFP images of the heart may have reduced ventricular signal due to reduced in-flow related signal enhancement of the 3D method, compared to a thin-slice 2D technique (26), although this does not explain the reduced myocardial signal. Overall, the 3D PR SSFP images, despite having decreased SNR and CNR relative to the 2D SSFP images, were of sufficient SNR (>20) and CNR for accurate quantitative analysis.

The Bland-Altman analysis (Table 2) and Fig. 3 show that there is some disagreement between techniques, although favorable or comparable to that found in other validation studies (22, 27,28). The radial streaking artifacts that are characteristic of undersampled PR imaging did not interfere with the qualitative analysis. Experience has shown that the myocardium is darker on 3D SSFP PR images compared to FT SSFP images, resulting in reduced epicardial contrast (at the myocar-dial-lung interface) (see Table 1 for myocardial-lung CNR and see Fig. 2). This may reduce the accuracy of epicardial tracings, thereby reducing the accuracy and reproducibility of the 3D LVEDM measurements. A potential cause of differences in the comparison between 2D FT and 3D PR is the gradient dewarping procedure, which was not applied in the offline reconstruction of the 3D PR images, but the very small bias between 2D and 3D measurements (Table 2) means this effect is small. The misregistration between the many breath holds of the 2D FT method could also cause an error in the measurements of volume and mass, leading to disagreement (see Fig. 2) (29). The zero-filled reconstruction of the 3D method might have affected the volume measurements by influencing choice of the most basal slice.

Temporal undersampling by the 3D method may be the most important source of disagreement for LVESV. A recent study of the influence of temporal and spatial resolution (30) on measurement of global function demonstrated that LVEF decreases by 3% with reduced temporal resolution (from 60 to 20 msec). This general trend, though muted, can be observed in Table 2, which shows that the mean 3D values for LVEF are 0.6% less and end systolic volumes are 2.8 mL greater than the 2D values. A similar study (31) has demonstrated that a temporal resolution better than 70 msec is needed for measuring global function accurately. Our temporal resolution (≈ 71 –80 msec) met this less stringent criteria. However, further improvement in temporal resolution is an important goal.

In conclusion, the 3D PR SSFP technique investigated here provides complete qualitative evaluation of cardiac function in one 24-heartbeat breath hold. The SNR is acceptable and the quantitative values for global function agree within limits which are similar to other validation studies comparing LV function methods. This 3D single breath hold method has the potential to replace nine to 12 separate breath-hold scans, thereby shortening total scan time or providing time for other imaging sequences that comprise a comprehensive cardiac exam.

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Figure 1.
The 3D PR SSFP acquires projections in the x-y plane, and partitions in z.

Figure 2.

Comparison of cardiac function imaging with the 2D FT SSFP and single breath-hold 3D PR SSFP techniques. A single slice and the volume rendered stack are shown. Observe the slice misregistration of the 2D stack evident in the septal wall and papillary muscles, and the complete apex to base coverage of both methods.

Figure 3.

a: Plot of LV end systolic and LV end diastolic volumes and masses of the 10 volunteers as measured by the 2D FT and the 3D PR method. **b:** Bland-Altman plot showing the limits of agreement between the 2D and 3D method for LV volumes. **c:** Bland-Altman plot showing the limits of agreement between the 2D and 3D method for LV mass.

Table 1

SNR and CNR Comparisons Between the 2D FT and 3D PR Methods

Method	Blood SNR	Myo SNR	Myo-blood CNR	Myo-lung CNR
2D FT SSFP	114 ± 31	42 ± 8	75 ± 26	33 ± 12
3D PR SSFP	59 ± 18	21 ± 9	43 ± 15	12 ± 8

Myo = myocardium.

Table 2Average Measurements ($N = 10$) and Measurement Differences of Cardiac Function for 2D FT vs. 3D PR

	LVEF (%)	LVEDV (mL)	LVESV (mL)	LVEDM (g)
3D PR SSFP	62 ± 3	172 ± 40	64 ± 13	93 ± 33
2D FT SSFP	63 ± 4	166 ± 36	61 ± 14	92 ± 28
<i>P</i> -value	0.623	0.102	0.172	0.775
Mean bias $\pm$ 2 SD	$0.6\% \pm 7.7\%$	$5.9 \text{ mL} \pm 20 \text{ mL}$	$-2.8 \text{ mL} \pm 12 \text{ mL}$	$-0.61 \text{ g} \pm 13 \text{ g}$

LVEF = left ventricular ejection fraction, LVEDV = left ventricular end diastolic volume, LVESV = left ventricular end systolic volume, LVEDM = left ventricular end diastolic mass, SD = standard deviation.

Table 3Interobserver Variability ($N = 10$) of Cardiac Function Measurements for Both 2D FT and 3D PR

	LVEDV (mL)	LVESV (mL)	LVEDM (g)
2D FT SSFPR ²	0.93	0.95	0.97
Variability (%)	5.4	4.9	6.1
Mean bias $\pm$ SD	-6.43 ± 9.3 mL	-0.42 ± 3.12 mL	9.89 ± 6.75 g
3D PR SSFPR ²	0.96	0.71	0.89
Variability (%)	5.3	10	11
Mean bias $\pm$ SD	-11.3 ± 9.00 mL	0.26 ± 6.95 mL	15.7 ± 10.35 g

LVEF = left ventricular ejection fraction, LVEDV = left ventricular end diastolic volume, LVESV = left ventricular end systolic volume, LVEDM = left ventricular end diastolic mass, SD.