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Free breathing three-dimensional late gadolinium enhancement cardiovascular magnetic resonance using outer volume suppressed projection navigators

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Abstract

Purpose—To develop a free-breathing, 3D late gadolinium enhancement (3D FB-LGE) cardiovascular magnetic resonance (CMR) technique and to compare it with clinically used 2D breath-hold LGE (2D BH-LGE).

Methods—The proposed 3D FB-LGE method consisted of inversion preparation, inversion delay, fat saturation, outer volume suppression, 1D-projection navigators, and a segmented stack of spirals acquisition. The 3D FB-LGE and 2D BH-LGE scans were performed on 29 cardiac patients. Qualitative analysis and quantitative analysis (in patients with scar) were performed.

Results—No significant differences were noted between the 3D FB-LGE and 2D BH-LGE datasets in terms of overall image quality score (2D: 4.69 ± 0.60 versus 3D: 4.55 ± 0.51 , $P = 0.46$) and image artifact score (2D: 1.10 ± 0.31 versus 3D: 1.17 ± 0.38 ; $P = 0.63$). The average difference in fractional scar volume between the 3D and 2D methods was 1.9 % ($n = 5$). Acquisition time was significantly shorter for the 3D FB-LGE over 2D BH-LGE by a factor of 2.83 ± 0.77 ($P < 0.0001$).

Conclusions—The 3D FB-LGE is a viable option for patients, particularly in acute settings or in patients who are unable to comply with breath-hold instructions.

Keywords

3D late gadolinium enhancement (LGE); stack-of-spirals imaging; motion correction; outer volume suppression; free-breathing; cardiovascular magnetic resonance

INTRODUCTION

Late gadolinium enhancement (LGE) cardiac MRI has become the method of choice for visualizing myocardial infarction (MI) as well as fibrosis caused by non-ischemic cardiomyopathy due to excellent scar contrast and high spatial resolution (1–4). The

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standard clinical LGE protocol acquires a single 2D slice during a breath-hold, which needs to be repeated more than 10 times to cover the entire left ventricular myocardium using short axis views and additional slices on long-axis orientations (5, 6). This protocol easily exceeds 10 min in total scan time, requiring on-the-fly adjustment of inversion delay to compensate for temporal variation in the concentration of contrast agent. Additionally, misregistration errors can occur between successive short axis slices due to variation in breath-hold positions, resulting in non-contiguous left ventricle (LV) coverage. Furthermore, repeated breath-holds increase patient fatigue resulting in poor breath-holding and subsequent motion artifacts. Single-shot imaging is available to speed up the scan and reduce the effects of arrhythmia and respiratory motion, but it has limitations of lower spatial resolution and/or longer acquisition window in each cardiac cycle than the standard protocol (7–9). Alternative 2D methods reported include free-breathing acquisitions that employ non-rigid registration of multiple image sets repeatedly acquired for each short-axis slice (10, 11). This technique has shown great promise particularly in vulnerable patients, but has drawbacks of reduced time efficiency due to the need for multiple acquisitions and difficulty of handling through-plane motion due to excitation of different slices.

Three-dimensional LGE is a promising alternative to 2D multi-slice LGE since its contiguous 3D spatial coverage potentially provides accurate quantification of the burden of infarct and allows for flexible post-analysis through arbitrary display views while minimizing the complexity of scan localization. While breath-hold 3D methods have been investigated with moderate slice thickness (comparable to those of 2D multi-slice protocols) (12–15), free-breathing scans with respiratory motion compensation would be necessary for high spatial resolution in all three dimensions (16, 17). Respiratory gating based on diaphragm navigator is the most established approach but typically involves long scan time due to suboptimal gating efficiency (18–20). Furthermore, the accuracy of the diaphragm navigator is limited due to subject-dependent motion correlation between the diaphragm and heart, and hysteresis during a respiratory cycle (21).

Respiratory motion estimation directly using the signal from the heart has potential for more accurate motion estimation (i.e. self-navigation). In principle, motion can be best estimated by calculating correlation between fully encoded images of the heart, but such an approach would be impractical due to long acquisition time for full k-space sampling with reasonably high spatial resolution. The acquisition of projection signal was proposed as a rapid respiratory navigator with great promise in cardiac cine imaging (22, 23). However, the accuracy of this approach appears to be limited since the projection signal also contains other structures (e.g. chest wall) which exhibit different motion from the heart. Alternative self-navigation techniques attempted to suppress the chest wall signal by using regional saturation pulses (24) and dual-projection acquisitions with dephasing gradient (25, 26) and were demonstrated for coronary angiography.

In this study, we aimed to develop a 3D free-breathing LGE method, which allows for near isotropic resolution, accurate respiratory motion correction and short scan time (< 4 min). The proposed method uses 3D stack-of-spirals k-space trajectories to speed up the scan, and outer volume suppressed 1D projection navigators for rapid and accurate estimation of

respiratory motion. We demonstrate the performance of the proposed technique in comparison with the clinically used 2D BH-LGE in cardiac patients.

METHODS

Pulse Sequence

The 3D FB-LGE imaging pulse sequence, shown in Figure 1(a), was cardiac-gated via electrocardiogram (ECG) and consisted of a 180° adiabatic inversion pulse, an inversion delay period (TI), a spectrally selective fat saturation pulse, a spatial outer volume suppression (OVS) pulse, a 1D projection acquisition and a segmented stack-of-spirals acquisition. The OVS preparation consisted of an adiabatic 90° tip-down pulse (27) followed by a spiral 2D spatially selective tip-up pulse with a circular passband of diameter 14 cm and a time-bandwidth-product (TBW) of 6 (28, 29) (Figure 1(b)). While the OVS preparation allows for reduced field of view (FOV) imaging for scan acceleration, another important benefit is improved accuracy of motion estimation by isolating the heart in the projection images that are acquired immediately afterwards. For motion estimation, 1D projection signals (duration = 5.7 ms) were obtained in three orthogonal directions by acquiring data along the k_x , k_y and k_z axes (22) (Figure 1(c)). The navigator acquisition used the same slab-selective RF excitation and readout TR as those for the subsequent spiral acquisition.

Data acquisition consisted of a 3D, dual density, segmented stack-of-spirals. Dual-density spirals were chosen for scan acceleration by fully sampling (Nyquist) the center of k-space and 1.7-fold undersampling the outer regions of k-space. An advantage of combining OVS preparation with dual-density spiral is that the OVS preparation suppresses spiral aliasing patterns that would occur as a result of the undersampling. To utilize the magnetization preparation most effectively, centric ordering was used along the k_z -direction to ensure the center of 3D k-space is sampled right after magnetization preparation in each cardiac cycle. During each segmented acquisition, the spiral index (kr) was kept unchanged while the k_z encoding index (kz) was either incremented or decremented from the k_z center. The acquisition of the same kr index ensures that potential signal variation during the transient data acquisition yields smooth 1-dimensional amplitude modulation along the k_z dimension (30). The kr index was incremented after all the data from the previous kr index is obtained (Figure 1(d)). The maximum gradient strength of the spiral waveform was 39 mT/m, and the maximum slew rate was 115 mT/m/s, which ensured stimulation levels were within scanner limits. To maintain phase coherence and to reduce flow artifacts, the zeroth and first moments of the spiral gradients were nulled.

Image Reconstruction

All image reconstruction including motion estimation and correction for 3D FB-LGE data were done offline in MATLAB (The Mathworks, Natick, MA). The motion along each direction (n) was estimated for each cardiac cycle by minimizing the mean squared error (MSE) between a reference projection and the projection at the current cardiac cycle:

$$MSE(\Delta n) = \sum_{m=W_l}^{W_u} [(|I_{ref}(m)| - |I(m - \Delta n)|)^2]$$

where $I_{ref}(m)$ is the reference projection image, $I(m)$ is the projection for the current cardiac cycle. W_l and W_u are lower and upper bounds of the window, respectively, along the projection signal in each dimension which encloses the heart. To determine the windows to use in the 1D projection signals, rectangular regions of interest (ROIs) of fixed sizes are manually located on sagittal and axial images so as to best enclose the left ventricle (red boxes in Figure 2(g)). These ROIs are then mapped to the 1D projection signals to obtain corresponding windows along each direction (Figure 2(a)–(c)). To determine $I_{ref}(m)$, a pilot motion estimate is run along the S-I direction using the projection using an arbitrary choice as initial reference (we chose 30). The pilot estimate gives the average displacement needed for all choices of reference projection. The reference projection that requires the least amount of motion correction to be added in order to correct the entire dataset is used as the reference projection. The initial reference choice does not affect the final reference image used. Once motion estimation is completed (Figure 2(d) – (f)), the 3D LGE data were corrected by adding a corresponding linear phase based on the estimate for each direction.

The motion-corrected 3D non-Cartesian spiral data were interpolated onto a Cartesian grid using a gridding method with a Kaiser-Bessel kernel width of 3 (31), and Voronoi-based density compensation (32). Offline data processing in MATLAB on a desktop computer with a 3.6 GHz processor and 8GB random access memory (RAM) ranged from 60–75 s. After data processing, the $1.6 \times 1.6 \times 2 \text{ mm}^3$ 3D dataset was interpolated to $1 \times 1 \times 1 \text{ mm}^3$ resolution and exported to OsiriX software (Pixmeo, Geneva, Switzerland) to produce arbitrary views of the isotropic 3D data.

Patient Studies

All experiments were performed on a Siemens 1.5T clinical MR scanner (Siemens Medical Solutions, Erlangen, Germany) equipped with a maximum gradient amplitude of 45 mT/m and maximum slew rate of 200 mT/m/s. A 6-channel body coil and a spine matrix coil are simultaneously used for signal reception. This study was health insurance portability and accountability act (HIPAA) compliant with the study protocol approved by the Institutional Review Board (IRB). Twenty nine patients (10 men, 19 women, age = 48.10 ± 14.69) were recruited following written informed consent. Study participants were adult patients scheduled for clinically ordered cardiovascular MRI that includes contrast agent administration, with an exclusion criteria of low renal function as defined by a globular filtration rate (GFR) < 60 mL/min/1.73 m² and pregnant women.

Contrast agent was administered intravenously using 0.1 mmol/kg of MultiHance (Bracco Diagnostic, Princeton, NJ) injected at 2 mL/s followed by 20 mL saline flush at the same rate. The conventional 2D BH-LGE was performed at approximately 10 min post-contrast agent injection. The imaging parameters included TR= 8.69 ms, TE= 3.41 ms, TI= 205–350 ms (determined by prior TI-scout scans), flip angle=25°, FOV = 360 mm, matrix size =230 × 256, slice thickness = 8 mm, and acceleration factor (GRAPPA) of 2. The scan time for

each slice took ~15 cardiac cycles depending on the breath-hold capacity of the patient (range = 8–15 cardiac cycles), with the in-plane resolution = $1.6 \times 1.4 \text{ mm}^2$, (phase encode resolution range = 1.6–2 mm) The acquisition window was fixed at 86.9 ms, with 10 k-space lines acquired per cardiac cycle. The 2D BH-LGE was acquired with 1R-R gating scheme. Approximately 10 short axis slices were acquired for contiguous LV coverage, followed by 3–4 long axis slices along the 2-chamber, and 4-chamber directions.

The 3D FB-LGE acquisition was performed either immediately preceding (n=19) or immediately following (n=10) the 2D BH-LGE scan. For the scans that preceded 2D LGE, the start time of 3D FB-LGE varied from 7–13 min, whereas for the scans that followed the 2D LGE, the 3D FB-LGE scan start time varied from 17–25 min post-contrast injection time. The dual-density spiral scan was designed in a way that yielded an in-plane FOV of 30 cm^2 near the k-space center and a FOV of 18 cm^2 in the outer k-space region (i.e., 1.7-fold undersampling) with a transition at 20% of the k-space radius. Other imaging parameters were scan orientation = sagittal, spatial resolution = $1.6 \times 1.6 \times 2 \text{ mm}^3$, number of spiral interleaves = 10, number of partition encodes = 80, spiral readout duration = 6.36 ms, TR=10.76ms, flip angle= 25° , view per segment = 4, acquisition window = 43.04 ms, total scan time = 200 heartbeats (3 min, 20s at 60 bpm). The inversion time to be used was determined by a breath-hold TI-scout scan just before the 3D FB-LGE scan. The acquisition start time for each patient was adjusted to start at mid-diastole based on 4-chamber cine images taken as a part of the routine cardiac MR protocol. For patients with arrhythmias typically exhibiting shortened diastolic periods, an acquisition window at end-systole was chosen (11 patients) (33). Furthermore, a delay of 300ms was added following end-systolic acquisition to avoid early triggering following irregular heartbeats, which may result in image artifacts with 1R-R imaging (6).

To enable comparison of 2D and 3D LGE datasets, the views obtained during the clinical 2D LGE scans (short-axis, 2-chamber and 4-chamber views) were reproduced using the 3D LGE data. The 3D images were reformatted to have the same slice thickness as the 2D LGE images (8 mm). The 2D and 3D LGE image sets were evaluated by a cardiologist (S.R., >15 years experience in cardiac MRI) and a cardiac radiologist (J.J. with 10 years experience in cardiac MRI). The images from the 2D and 3D techniques were anonymized, randomized and blinded to the technique used before being presented to the observers. The images were scored individually by the two observers, and a consensus score agreed upon for the conflicted scores. The 2D and 3D images were rated based on Likert scales for overall image quality and image artifacts. The scale for image quality was: 1-non-diagnostic; 2-poor quality and limited interpretation; 3-moderate quality, adequate for interpretation; 4-good quality, permits identification of fine structures; 5-excellent quality, allows definition of fine anatomic structures. The scale for the image artifact score was: 1-no artifacts; 2-some artifacts, but does not interfere with diagnostics; 3- moderate, may interfere with diagnosis; 4-severe, affects diagnosis.

Quantitative analysis was done in 5 patients' LGE images which exhibited hyperenhancement. The epicardial and endocardial borders of the myocardium were manually delineated. For the slices containing an infarct, ROIs were drawn in the myocardium to enclose the scar areas and in the normally nulled myocardium. The

fractional scar volume was calculated as the volume of the scar to volume of entire myocardium. The signal enhancement ratio was calculated as $(S_{\text{scar}} - S_{\text{normal}})/S_{\text{scar}}$ where S_{scar} is the mean signal intensity, S_{normal} is the mean signal intensity in normal myocardium. The scar contrast-to-noise ratio was not calculated due to spatially variant noise distribution in parallel imaging for 2D BH-LGE and due to the residual spiral aliasing artifacts from variable density spiral encoding in 3D FB-LGE.

Wilcoxon signed-rank tests were used to compare the 2D and 3D data as the values did not show a normal distribution. For all tests, a P-value < .05 was considered significant. Agreement between the 2D and 3D techniques in measuring the fractional scar volume was assessed using Pearson correlation coefficient.

RESULTS

Figure 2 shows the representative results of motion estimation and correction using the 1-D projection navigators. The estimated motion in the 29 patients ranged from 6–15 mm along the SI direction, 2–5 mm along the AP direction, and 1–4 mm along the RL direction. The steady shift seen in all 3 dimensions may be attributed to diaphragmatic drift (Figure 2 (d–f)). Images obtained before motion correction (Figure 2(g)) and after motion correction (Figure 2(h)), along the sagittal, axial and coronal directions (left-right), demonstrate significant improvement in image quality, and suppression of motion-induced blurring (yellow arrows).

Figure 3 contains representative cross correlation (CC) between the reference projection signal and the remaining projection signals across the respiratory cycle along the SI (Figure 3(a)), AP (Figure 3(b)) and RL (Figure 3(c)) directions. The CC values of the original projection images (dashed black) fluctuate, yielding local minimums when the estimated displacements (dashed blue) are large. After all the projection signals are shifted based on the motion estimates, CC becomes closer to 1 with reduced fluctuation, due to increased similarity among the shifted projections. Across all subjects, the CC values were 0.86 ± 0.02 (S-I), 0.95 ± 0.04 (A-P), and 0.97 ± 0.11 (R-L) for uncorrected projections, and were increased to 0.92 ± 0.05 (S-I), 0.96 ± 0.04 (A-P) and 0.98 ± 0.08 (R-L) for corrected projections.

The overall image quality score was 4.69 ± 0.60 and 4.55 ± 0.51 in 2D and 3D LGE data, respectively, showing no significant differences ($P = 0.46$). The difference in image artifact score was also not statistically significant (1.10 ± 0.31 vs 1.17 ± 0.38 , $P = 0.63$). Table 1 gives a break-down of the consensus scores by the observers. 2D BH-LGE received the highest image quality score in a larger number of data sets than 3D FB-LGE (22 vs 16) but received the score of 3 in two patients who could not maintain good breath-hold. 3D FB-LGE received scores of higher than 3 consistently, as indicated by the smaller standard deviation. Table 2 shows the quantitative analysis done on patients with scar ($n=5$). The mean scar enhancement ratio was 0.73 ± 0.09 and 0.61 ± 0.06 in 2D and 3D LGE data sets, respectively, and the average difference in fractional scar volumes was 1.9%. The scan time was significantly shorter in 3D FB-LGE (182 ± 32.81 sec) than in 2D BH-LGE (507.41 ± 133.64 sec), which yielded scan acceleration by a factor of 2.8 ($P < 0.0001$).

Figure 4 shows representative images obtained using 2D BH-LGE (Figure 4a) and 3D FB-LGE (Figure 4b) in a 23 year old female patient with hypertrophic cardiomyopathy, manifesting primarily as increased septal wall thickness compared to the lateral wall (yellow arrow). The contiguous 3D coverage allows more accurate delineation of structures (red arrow). The 3D FB-LGE images are free from motion artifacts with excellent image quality comparable to the 2D BH-LGE images. Figures 5(a) and (b) show LGE images of a 55 year old male patient with a recent history of ST elevated myocardial infarct (STEMI). Both the techniques exhibit similar enhancement patterns indicating MI along the anterior wall, with a transmural extent less than 50%. The fractional scar volume in this patient is measured as 7.1 % and 8.4 % for 2D BH-LGE and 3D FB-LGE, respectively.

Figure 6 contains a set of 2D (a) and 3D (b) LGE images in a cardiomyopathy patient (67 year old female) who had difficulty holding her breath. The 2D BH-LGE suffers from motion related image artifacts that vary over slices reflecting the patients breath-hold quality for the corresponding slice acquisition. The 3D FB-LGE provides significantly improved images as a result of successful respiratory motion correction. Figure 7 contains a set of 2D (a) and 3D (b) LGE images in a patient with severe arrhythmia (38 year old female). 2D BH-LGE with the typical diastolic acquisition yielded significant image artifact due to the data inconsistency caused by variations in R-R intervals. 3D FB-LGE with the end-systolic acquisition and post-acquisition delay significantly improved the image quality by reducing the data inconsistency.

DISCUSSION

In this study, we developed a 3D free-breathing LGE CMR technique using a stack-of-spirals acquisition, and respiratory motion correction based on outer volume suppressed projection navigators. The time-efficiency inherent in spiral encoding along with 100 % navigator efficiency allowed for 3D acquisition with near-isotropic resolution in scan times as short as 2.5 min. The free-breathing nature of the 3D LGE sequence reduces patient fatigue, and is particularly beneficial for patients who have trouble with breath-holding, thus reducing the likelihood of image artifacts. The 3D encoding ensures contiguous LV coverage and prevents slice misregistration errors, potentially allowing for more accurate quantification of myocardial volume.

In respiratory self-gating techniques that use projection signals, it is important to limit the source of the gating signal to the heart to improve the accuracy of the motion estimation. To our best knowledge, the dual-projection navigator method proposed by Lai et al. (25) is the only self-gating technique that involves suppression of the volume outside the heart. Here two S-I projections were averaged with appropriate A-P dephasing gradients that yielded an S-I projection of spins that was weighted by a sinusoidal function in the A-P direction, and the resulting artificial bright and dark bands were positioned over the heart and chest wall, respectively, to achieve chest wall suppression. This approach is very fast without need for magnetization preparation but has limitations including smooth transitions from the bright and dark bands and only one-dimensional selectivity set to the A-P direction. Our approach, at the cost of additional time for OVS magnetization preparation, enables two-dimensional

selectivity with much sharper transitions between pass- and stop-bands, thus minimizing the contribution of non-cardiac structures to the projection signal.

The 3D FB-LGE acquisition occurred at either mid-diastole or end-systole. The mid-diastolic acquisition is the most commonly used in LGE imaging and the most comparable to the 2D breath-hold acquisition in this study. While the quiescent phase in mid-diastole is longer making it suitable to image patients with normal heart-rhythms, the diastolic phase shortens to a greater extent as compared to end-systolic phase with increasing heart-rates (34), or in patients with arrhythmias (35). In our study, therefore, end-systole acquisition was used for patients exhibiting heart rate patterns of atrial fibrillation or pre-ventricular contractions. The acquisition window of approximately 43 ms was expected to be short enough to prevent motion effects for the end-systolic acquisition. Another issue with variability in R-R intervals is premature triggering which may compromise image quality (6). We tried to mitigate this effect by adding a delay following the acquisition to avoid undesired early triggering. While the mid-diastole acquisition in 2D LGE acquisitions caused significant ghosting artifacts in some arrhythmia patients (Figure 6), the end-systole acquisition coupled with a delay gave acceptable results in our study group. Further studies are required to study the robustness of end-systole versus mid-diastole acquisitions in arrhythmia patients.

In our study the 3D LGE scan was done as early as 7 minutes post-contrast agent injection. The mean difference in the optimal TI time obtained from TI-scout images taken right before each 3D scan and 2D scan from the 7–9 minute range was 25.83 ms. For 3D scans done in the range of 9–13 minutes range the mean difference in optimal TI times between 3D and 2D scans was 21.67 ms. With a $P > 0.1$ ($P = 0.57$) a significant difference was not observed in our patient cohort. Hence, we assume that the corresponding image quality for the 3D LGE studies done in the 7–9 minute range is not significantly affected by earlier start times of the 3D LGE scans. Although it would be expected that the heart rate would affect image quality by reducing the time for T1 relaxation when using 1R-R acquisition (6), we did not see significant image quality degradation with increasing heart rate in the 3D LGE. Two possible reasons to explain this could be that the maximum heart rate seen in our study was 90 (which is not as high as reported in other studies (11)). A second reason could be that patients who had a high heart rate >85 had intermittent arrhythmias too, which resulted in a ~ 300 ms delay being added following end-systole acquisition, subsequently resulting in 2RR acquisition only when arrhythmias occurred. This may result in more benign motion effects. The mean image quality in the 80–90 bpm range was higher in the 3D FB-LGE technique (4.60 ± 0.55) compared to the 2D BH-LGE (4.40 ± 0.89).

The proposed motion correction scheme approximated respiratory motion as translations along three orthogonal axes. The omission of rotation and non-rigid-body deformation is a limitation of the proposed technique, and is likely the primary reason for the smaller number of FB-LGE data that received the highest score (16 out of 29 = 5) than BH-LGE (22 out of 29 = 5). A simple approach to this issue would be to discard outliers with too much motion through prospective gating using the established diaphragm navigator (18) or retrospective gating after multiple acquisitions (10). The selected acceptance of data, however, involves increase in total scan time. Another possible approach of our particular interest was recently

proposed, which compensates for non-rigid motion by combining several scaled 3D translations (e.g. 0.5×, 1×, 1.5× with respect to reference estimates) in a spatially dependent way that yields the most focused images (36, 37). Our initial experience of this approach using 3D FB-LGE data with the projection-based estimates as the reference has shown slight improvement of image quality (Supporting Figure S1). Further investigation would be needed to assess the practical value of additional non-rigid correction with consideration of the substantial increase in reconstruction time caused by the iterative nature of the algorithm on a larger set of LGE data with scar.

In the qualitative scoring for image quality, 78% of 2D BH-LGE and 55% of 3D FB-LGE were rated as excellent (rating score=5), while 17% of 2D BH-LGE and 45% of 3D FB-LGE were rated at a score of 4. Apart from reasons related to translation only correction, the projection signal may not have included sufficiently discernable features of the heart's anatomy presumably due to incomplete suppression of the OVS. An important point to note is that while two patients who had difficulty with breath-holding, were assigned the lowest image quality score of 3 in the 2D BH-LGE data, none of the patients received such a low score in the 3D FB-LGE dataset. Indeed, patients with arrhythmia and inability to hold breaths represent the most vulnerable patients with the highest risk of adverse outcomes, shown to result in deteriorating BH-LGE image quality for these patients (11). This demonstrates a critical advantage of the free-breathing nature of the 3D LGE and illustrates the need for imaging techniques that work sufficiently well for the sickest patients.

The main limitation in this study is the small number of patients with scar (5 out of 29 patients), which has prevented drawing any statistical comparisons of 2D and 3D LGE in depiction of scar. The diagnostic performance of the proposed method needs to be further investigated in a larger cohort of patients with various patterns of scar. Diffuse or subtle hyperenhancement was noted in the 3D and 2D techniques in 5 additional patients who were diagnosed with non-ischemic cardiomyopathy, but were not quantified. The epicardial and endocardial borders were manually delineated based on the contrast between the blood pool and myocardium. This may be a potential source of quantification errors in the myocardial infarct volume fraction calculation in the 2D and 3D LGE images. The acquisition scheme used here was 1 R-R, which may produce artifacts and reduce scar contrast due to incomplete relaxation of magnetization in patients with severe R-R variability (6). Although 2 R-R triggering would have significantly resolved this issue, this option was eschewed for a design with shorter scan time. A minor limitation of the 3D FB-LGE is that reconstruction is done offline, and images are not immediately available to the physician following the scan.

CONCLUSION

We have developed a free-breathing 3D LGE method using outer volume suppressed projection acquisition for respiratory motion correction and stack-of-spirals acquisition for scan time reduction. The proposed 3D LGE method allows for near-isotropic spatial resolution and complete myocardial coverage in a scan time of approximately 3 min while maintaining similar image quality to the conventional 2D breath-hold LGE. The proposed 3D LGE technique offers a viable alternative to the conventional 2D LGE approach particularly in patients having trouble with breath-holding.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

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FIG. 1.

Schematic illustration of the 3D FB-LGE sequence. The pulse sequence timing diagram is shown in (a), the OVS implementation (b) which uses a adiabatic 90° tip-down pulse followed by a -90° 2D spiral tip-up pulse. The motion correction scheme using 1D projection signals (c) that sample the central lines of 3D k-space along three orthogonal directions. The interleaved center-out (along k_z) sampling pattern used by the segmented stack-of-spirals during data acquisition is shown in (d).

FIG. 2.

Motion correction using outer volume suppressed projection navigators. 1D projection signals along the S-I (a), A-P (b) and R-L (c) directions during the entire scan. The ROIs shown in (g) (red boxes in sagittal and axial orientations) are used to determine the windows to be used in the 1D projections (red lines in (a), (b) and (c)). Estimated motion along the S-I (d), A-P (e), and R-L (f) directions. Selected sagittal, axial and coronal images before (g) and after (h) the projection-based motion estimation and correction. Fine anatomic

structures including the myocardial boundary and papillary muscles are visualized well following motion correction (yellow arrows).

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FIG. 3.

Representative cross correlation (CC) between the reference and other projection signals, and corresponding motion estimates along S-I (a), A-P (b), and R-L (c) directions. The CC of the original projection signals fluctuates over the cardiac cycle depending on displacements from the reference position (dotted black line). The CC of motion-corrected projection signals shows less fluctuation and is closer to 1, indicating increased similarity between projection signals (solid red line).

FIG. 4.

Representative 2D BH-LGE and 3D FB-LGE images from a cardiomyopathy patient. A 23-year-old female patient with hypertrophic cardiomyopathy which manifests as increased septal wall thickness compared to the lateral wall (yellow arrows). Contiguous 3D coverage allows better delineation of structures (red arrows). Images from the clinically used 2D BH-LGE (**a**) and the 3D FB-LGE (**b**) show comparable image quality (qualitative image quality rating score for both image sets were excellent, with no artifacts present).

FIG. 5.

Exemplary 2D BH-LGE (**a**) and 3D FB-LGE (**b**) images in a 55-year-old male patient with a recent history of STEMI. Both techniques show subendocardial enhancement along the anteroseptal, anterior and anterolateral walls with a transmural extent of less than 50%. The 3D images better depicts the infarcts in apical slices due to higher through-plane resolution and fat suppression.

FIG. 6. Exemplary 2D BH-LGE and 3D FB-LGE images of patient having trouble with breath-holds. A 67-year-old female patient with suspected cardiomyopathy who had trouble with breath-holding exhibits considerable motion artifacts in the 2D BH-LGE (a) images. The 3D FB-LGE (b) provides significantly improved images of this patient devoid of motion artifacts (arrows).

FIG. 7.

Exemplary 2D BH-LGE and 3D FB-LGE images of a 38-year-old female patient with arrhythmia. The 2D BH-LGE (**a**) using the typical diastolic acquisition yielded noticeable image artifacts due to inconsistent diastolic phases in the presence of arrhythmia, while 3D FB-LGE (**b**) images using an end-systole acquisition provides acceptable image quality without significant image artifacts.

Table 1

Break-down of qualitative image scores

	Image Quality					Image Artifact			
	1	2	3	4	5	1	2	3	4
Rating									
2D BH-LGE	0	0	2	5	22	26	3	0	0
3D FB-LGE	0	0	0	13	16	24	5	0	0

Image quality was scored on a 5-point scale (1=non-diagnostic to 5=excellent), and image artifact was scored on a 4-point scale (1=no artifact to 4= severe artifact).

Table 2

Quantitative Analysis on patients with scar (n=5)

	<u>Scar Enhancement Ratio</u>		<u>Fractional Scar Volume (%)</u>	
	2D BH-LGE	3D FB-LGE	2D BH-LGE	3D FB-LGE
Subject 1	0.76	0.71	1.4	2.0
Subject 2	0.60	0.57	12.3	16.6
Subject 3	0.75	0.63	7.0	4.8
Subject 4	0.84	0.56	8.0	12.8
Subject 5	0.70	0.57	8.2	10.3