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Non-Cartesian Balanced SSFP Pulse Sequences for Real-Time Cardiac MRI

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Abstract

Purpose—To develop a new spiral-in/out balanced steady-state free precession (bSSFP) pulse sequence for real-time cardiac MRI and compare it with radial and spiral-out techniques.

Methods—Non-Cartesian sampling strategies are efficient and robust to motion and thus have important advantages for real-time bSSFP cine imaging. This study describes a new symmetric spiral-in/out sequence with intrinsic gradient moment compensation and SSFP refocusing at $TE=TR/2$. In-vivo real-time cardiac imaging studies were performed to compare radial, spiral-out, and spiral-in/out bSSFP pulse sequences. Furthermore, phase-based fat-water separation taking advantage of the refocusing mechanism of the spiral-in/out bSSFP sequence was also studied.

Results—The image quality of the spiral-out and spiral-in/out bSSFP sequences was improved with off-resonance and k-space trajectory correction. The spiral-in/out bSSFP sequence had the highest SNR, CNR, and image quality ratings, with spiral-out bSSFP sequence second in each category and the radial bSSFP sequence third. The spiral-in/out bSSFP sequence provides separated fat and water images with no additional scan time.

Conclusions—In this work a new spiral-in/out bSSFP sequence was developed and tested. The superiority of spiral bSSFP sequences over the radial bSSFP sequence in terms of SNR and reduced artifacts was demonstrated in real-time MRI of cardiac function without image acceleration.

Keywords

spiral; spiral-in/out; bSSFP; real-time cardiac imaging; fat-water separation

INTRODUCTION

Balanced steady-state free precession (bSSFP, also known as True-FISP, FIESTA, or balanced FFE) pulse sequences are widely used in cardiac magnetic resonance imaging (MRI), because of their short acquisition time and high contrast between the blood pool and

myocardium (1–3). The standard method for assessing cardiac function using cardiac magnetic resonance (CMR) is to image during a breath-hold using an ECG-gated bSSFP pulse sequence with segmented Cartesian readouts. While this method yields high quality results in most patients, it can yield substandard image quality in patients who have arrhythmia or who are incapable of holding their breath. Also, more than ten breath-holds are typically required to cover the left ventricle, which lengthens the exam time and can tire the patient. Thus, a real-time bSSFP pulse sequence that could be performed without cardiac gating or breath holding would be of value in the assessment of cardiac function.

Non-Cartesian sampling patterns are often more time-efficient than a Cartesian sampling pattern, which fills k-space line by line. Therefore, non-Cartesian bSSFP sequences could play an important role in real-time cardiac MRI. Radial and spiral k-space trajectories are the most common non-Cartesian imaging trajectories. Radial methods reduce scan time by undersampling the outer portions of k-space (4). Spiral methods reduce scan time primarily by scanning more of k-space in a given TR (5), although undersampling is also possible using variable-density scanning (7–8). Both radial and spiral sequences are inherently more robust to flow and motion artifacts than Cartesian sequences, because they repeatedly sample the center of k-space and thus effectively average the high-energy center of k-space. This reduces artifacts resulting from motion between readouts. Another factor to consider is motion during the readout. In that case, spiral trajectories have an advantage in that the oscillating and rotating readout gradients do not accumulate gradient moments, and thus the main effect of motion during the trajectory is a blur along the direction of motion (5–6).

Both sequences have their disadvantages. Spiral bSSFP is more sensitive to off-resonance effects, whereas radial bSSFP suffers more from streaking artifacts due to undersampling. Although spiral bSSFP sequences have been used in research studies (9–10), clinical adoption has been slower than for Cartesian and radial bSSFP sequences. Therefore, one goal of this study was to compare radial and spiral bSSFP sequences for real-time CMR assessment of cardiac function.

In addition to studying a conventional spiral-out bSSFP sequence, in this study we also developed a new spiral-in/out bSSFP sequence (11). The idea of this sequence is to move the echo time (TE) to the center of the repetition time (TR) as in a typical Cartesian or radial bSSFP sequence. Potential advantages of this approach include an increase in signal-to-noise ratio (SNR) and a reduction in off-resonance artifacts, because spins refocus at the center of the TR (3,12). A related advantage of this approach is that phase detection can be used for fat-water separation (13). Finally, a spiral-in/out bSSFP sequence has a simpler gradient rephaser design than a spiral-out bSSFP sequence, which requires a rephaser gradient that nulls both the zeroth and first order gradient moments to suppress artifacts arising from in-plane motion and blood flow (9). The spiral-in/out sequence is designed to have both zeroth and first order gradient moment nulling via symmetry. Therefore, another goal of this study was to design a spiral-in/out bSSFP sequence and compare it to radial and spiral-out bSSFP sequences.

In the following sections, the gradient design methods for the spiral-out and spiral-in/out bSSFP sequences are first introduced. Then, methods for correcting for off-resonance and k-

space trajectory infidelity are described, and fat/water separation is discussed. Finally, an experimental comparison of the performance of the radial and spiral bSSFP in normal volunteers is described.

METHODS

Gradient Design

In a typical bSSFP gradient design, the gradient-induced dephasing within one repetition time (TR) should be exactly zero. Furthermore, in real-time cardiac imaging, to suppress the in-plane motion and flow artifacts, the first-order gradient moment should also be compensated over each TR (9). Therefore, we aim to design the gradients to satisfy $m_0=0$ and $m_1=0$, where m_i is the i^{th} order gradient moment at the end of the TR.

The gradient design of the two spiral sequences is based on a Siemens Avanto 1.5T scanner with 40 mT/m maximum gradient amplitude and 170 T/m/s maximum slew rate. Here we will focus on the spiral readout and rephaser design, since the commonly used slice-selection gradients with sinc RF pulses are the same for all three sequences.

Spiral-Out bSSFP—Spiral-out gradients are comprised of the desired spiral readout gradients followed by rephaser gradients that null the 0th and 1st order gradient moments of the readout gradients. The algorithm introduced by Meyer et al. (5) was used for minimum-time spiral readout gradient design constrained by gradient slew rate and amplitude limits. The desired k-space trajectory is given as $k(\tau)=A\tau e^{i\omega\tau}$, where τ is a function of time and ω is a parameter determined from the Nyquist sampling criterion. In variable density spiral trajectory design, ω is linearly decreased with each gradient sample from ω_{max} at the center to ω_{min} at the edge of the k-space.

For the rephaser design, we used an algorithm described by Nayak et al. (9) to simultaneously compensate the 0th and 1st order gradient moments via triangular gradients. The algorithm proceeds as follows: (1) the spiral readout gradients are rotated so that the gradient on one axis ends with zero amplitude; (2) two to three sets of triangular gradients are designed with lengths derived from the gradient moment nulling equations; and (3) the gradients are rotated back. Figure 1 shows the resulting gradients and the corresponding trajectory of a single interleaf in a 32-interleaf variable density design.

In order to get N spiral interleaves to cover all of k-space, we rotate the gradients from a single interleaf by $2\pi/N$ each time. In addition, we reorder the interleaves in a bit-reversed manner to increase the distance between k-space samples on consecutive interleaves to reduce dynamic aliasing. For an arbitrary interleaf number N that is not a power of 2, we use a bit-reversed order table of the smallest power of 2 greater than N , discarding the order numbers equal to or larger than N .

Spiral-In/Out SSFP—The gradients for spiral-in/out bSSFP sequence are designed as follows: 1) generate the desired spiral-out arm of the spiral-in/out readout gradients; 2) generate time-optimal transition gradients following the spiral-out arm to move the k-space trajectory to the origin and simultaneously reduce the gradient magnitude to zero; 3) time

reverse the previous gradients and put the reversed gradients in front, so that the k-space trajectory is at the origin at the midpoint of the entire gradient waveform; 3) rotate and reorder to get all of the gradient waveforms.

The spiral-out arm is generated using the same algorithm as in the spiral-out bSSFP readout gradient design introduced in the previous subsection; variable density spiral design is preferred here as well. However, the effective interleaf number is doubled in calculation of ω , since each interleaf of spiral-in/out readout gradients can be viewed as two spiral arms symmetric about the origin (ignoring the traversal direction of the trajectory). To design the transition gradients, we also use the triangle-based method to simultaneously drive the gradient amplitude and the k-space position to zero.

By just time reversing the previous gradients, we null both the 0th and 1st gradient moments. The proof is provided in the Appendix. Figure 2 shows the resulting gradients and the corresponding trajectory of a single interleaf in a 32-interleaf variable density design. The higher-order moments are not zero at the end of the readout, but they are relatively small, because of symmetry and the oscillating nature of spiral gradients.

The rotation and reordering step is similar to the spiral-out gradient design, except the rotation angle is π/N for each interleaf, since one interleaf would overlap with itself after rotating at angle π .

The symmetric property of the gradients not only simplifies the rephaser design, but also shortens the minimum TR in comparison with the spiral-out bSSFP sequence, because the prephaser and the rephaser gradients can overlap with the slice-selection rephaser and prephaser gradients. In the spiral-out bSSFP sequence the overlap is only feasible on one side. This shortening can also be traded off for a longer spiral readout, as we did in our experiments.

Off-Resonance Effects

Banding artifacts can occur in bSSFP sequences due to main field inhomogeneity. In addition to careful shimming to increase homogeneity, reduction of the TR is an effective way to reduce banding artifacts. In this study, we used a Siemens gradient-echo cardiac shim Works-in-Progress package and optimized our gradient design to achieve a TR of about 3.5 ms for both spiral sequences. The TR for the radial bSSFP sequence is even lower (2.36 ms) due to the fact that only one radial spoke is acquired during each TR.

In addition, to reduce image blurring in the region of interest (ROI) for both spiral bSSFP sequences, a fat-suppressed low-resolution field map is acquired before the dynamic image acquisition using a spectral-spatial RF excitation pulse (14) and two single-shot spiral readouts with different echo times. A linear fit of the field map is performed and the constant and linear terms of the fit are used to correct for center frequency and k-space trajectory warping, respectively (15).

K-Space Trajectory Fidelity

In spiral sequences, linear eddy currents and associated anisotropic delays of the gradient system will affect the fidelity of the k-space trajectory and cause image blurring and/or distortion (16–17). It is possible to measure the actual k-space trajectory and use the measured trajectory in image reconstruction. However, it is not practical in real-time cardiac MRI to measure the trajectory for each acquisition, since the k-space measurement time is long and the actual trajectory depends on the image orientation. Here, we used a model-based method to estimate the actual trajectory. The trajectory estimation method has been studied for spiral-out GRE sequences (17) and can be directly applied to the spiral-out bSSFP sequence in this study. For the spiral-in/out bSSFP sequence, we used a similar method and studied the validity of the model for this trajectory. Furthermore, we simplified the model as introduced in (17) by using two fitting parameters instead of three.

The k-space trajectory estimation model is given as:

$$\begin{aligned} K(t) &= K_d(t) + K_e(t) \\ &= K_d(t) + \int_0^t s(\tau) \otimes H(\tau) d\tau \\ &\approx K_d(t) + A \int_0^t G_d(\tau) d\tau + B \int_0^t \left[\int_0^\tau G_d(\tau') d\tau' \right] d\tau \quad [1] \\ &= (1+A)K_d(t) + B \int_0^t K_d(\tau) d\tau \end{aligned}$$

where $K_d(t)$ is the k-space trajectory on one physical axis with a gradient delay T and $K_e(t)$ is the additional k-space term induced from the linear eddy currents and can be calculated by integrating the convolution of the slew rate $s(\tau)$ and an impulse response function characterizing the gradient system, $H(\tau)$. A Taylor series expansion is then performed on $H(\tau)$ and the 0th and 1st order terms are retained. To determine the values of the optimal delay time T , $A'=1+A$, and B on each physical axis, a set of trajectory measurements was performed on the scanner using a modified Duyn's method (17–18), followed by a least squares fit with the model given in Eq. [2]. The values are assumed to be constant and independent of the image orientation and spiral parameters. The estimated trajectory for a given scan is calculated as follows: (1) the gradients on each physical axis and the corresponding delayed k-space trajectory $K_d(t)$ are determined using the rotation matrix; (2) Eq. [2] is used to get the estimated k-space trajectory for each physical axis; and (3) the trajectories on the three physical axes are rotated back to the logical coordinate system and used for image reconstruction.

To evaluate the effects of the k-space trajectory estimation model, we measured the actual k-space trajectory in three in-vivo experiments using the spiral-in/out bSSFP sequence. The estimated k-space trajectory was compared with the commonly used single-delay trajectory, which is calculated assuming an isotropic delay on each axis, ignoring linear eddy currents. Both the k-space trajectories themselves and the images reconstructed with these trajectories were compared to the measured trajectory and corresponding images.

Fat-Water Separation

In a general bSSFP sequence, the refocusing mechanism at $TE=TR/2$ will result in a phase cycling property for different local off-resonance frequencies. The phases of one pixel will alternate between 0 and 180 as a function of local off-resonance frequency with a period of

$1/TR$. In the spiral-in/out bSSFP sequence introduced above, TR is 3.69 ms and thus the corresponding period is 271 Hz. Since the off-resonance frequencies of fat and water at 1.5 T differ by about 220 Hz due to chemical shift, the fat and water signals will have opposite phases at a large range of local off-resonance frequencies. Furthermore, if the local off-resonance frequency due to other factors is small enough, the phase of the water signal will be 0 while the phase of the fat signal will be 180. Therefore, a phase detection method (13) can be used to separate the fat pixels from the water pixels without any further measurement.

In the current sequence parameter settings at 1.5T, this technique can only be applied with the spiral-in/out bSSFP sequence. The radial sequence has a very short TR, so that the water and fat pixels are very likely to be in the same phase band. The spiral-out sequence does not have the refocusing mechanism at the center of k-space.

Experimental Setup

Comparison studies were performed on a Siemens Avanto 1.5 T scanner equipped with a surface coil array. Six healthy volunteers (4 males and 2 females, ranging in age from 20 to 30) and one patient with informed consent participated in this study. For each volunteer, a mid ventricular short-axis view and a horizontal long-axis view were imaged under both breath-held and free-breathing conditions. For the patient, three short-axis views at the basal, mid, and apical sections of the left ventricle and a horizontal long-axis view were imaged under free-breathing conditions for patient comfort. For each set of experiments, the radial, spiral-out, and spiral-in/out bSSFP sequences were run consecutively at the same image plane with 8 mm slice thickness and $340 \times 340 \text{ mm}^2$ FOV and the same shim settings. Other sequence parameters and the resulting spatial and temporal resolutions are given in Table 1. The spatial resolution here is calculated from the full width at half maximum (FWHM) of the theoretical point spread functions (PSFs) of the non-Cartesian sampling patterns. The temporal resolution is expressed as the number of fully sampled images per second. The reconstructed matrix size was 128×128 for all three sequences. For a direct comparison, we matched the spatial and temporal resolution of the two spiral sequences with the radial sequence of a typical clinical scan. The minor differences in the flip angle are due to specific absorption rate (SAR) limits, since more RF pulses per unit time are applied with the radial bSSFP sequence.

Online reconstruction was implemented for all three sequences on the scanner. View sharing was used for all three sequences to double the reconstructed frame rate. The images from each coil were separately reconstructed and combined via a sum-of-squares method.

Sequence Comparison

Given the trajectories of the radial, spiral-out and spiral-in/out bSSFP sequences, the theoretical point spread functions can be calculated. The aliased energy contained in the side lobes of the PSFs was estimated and compared. Experimentally, we compared the SNR, contrast-to-noise ratio (CNR) and the overall image quality of the reconstructed images with the three sequences.

SNR & CNR—With a non-Cartesian sampling pattern, the aliasing due to undersampling often has the appearance of additional noise. Therefore, to directly represent the visual clarity of the images, the apparent SNR of the blood is measured by dividing the mean image intensity at the specified blood region and the mean image intensity outside the chest and multiplying the result by 1.13 to account for the fact that the mean magnitude of the apparent noise is measured instead of its standard deviation (19). Similarly, the apparent CNR of the blood and myocardium is also measured by subtracting the apparent SNR of the blood with that of the myocardium.

For each dynamic image series, 5 images covering the heart cycle were selected from 40 images for SNR and CNR measurement. The values were then averaged as the SNR and CNR for this image series. Differences in SNR and CNR for the three pulse sequences were analyzed using repeated measures analysis of variance, in order to isolate the SNR and CNR effects of different pulse sequences from the effects of subject-to-subject variations. Free-breathing and breath-held experiments were treated as repetitions of the experiment, because the measured SNR and CNR were similar for the two types of scans. Pairwise comparisons were performed using the Tukey-Kramer method.

Image Quality Rating—To estimate the overall image quality based on its clinical value, two experienced cardiologists performed a blind rating of these images/videos on a 0–5 scale. The criteria used were as follows:

- 5: Excellent. There are minimal artifacts and the images were equivalent to those from a breath-held gated cine sequence.
- 4: Very good. There are some minor artifacts.
- 3: Good. The images are clinically acceptable with diagnostic quality.
- 2: Poor. There are significant artifacts that would limit diagnostic utility.
- 1: Very poor. The images do not have diagnostic quality.

Differences in image quality ratings for the three pulse sequences were assessed using the Kruskal-Wallis test. Pairwise comparisons were performed using the Wilcoxon signed rank test with a Bonferroni correction for multiple comparisons.

Ejection Fraction Calculation—To compare the three real-time bSSFP sequences for LV quantification, the ejection fraction (EF) was calculated by acquiring a stack of short axis views covering the entire LV on three healthy volunteers. A standard Cartesian breath-held cardiac-gated bSSFP cine sequence was used at the same slice locations to calculate the EF and used as the gold standard. The EF was calculated using manual segmentation in OsiriX.

RESULTS

k-Space Trajectory Fidelity

Figure 3 shows a comparison of the single-delay trajectory, the measured trajectory and the estimated trajectory for a spiral-in/out bSSFP sequence in one study. The estimated

trajectory closely approximates the measured trajectory, whereas the single-delay trajectory deviates more from the measured trajectory as time increases, as predicted by Eq. [2] where the integral term increases with time. The root mean square error (RMSE) between the estimated trajectory and the measured trajectory was reduced by $59.6\% \pm 4.9\%$ as compared with the RMSE between the single-delay trajectory and the measured trajectory.

Figure 4 shows the images reconstructed with the single-delay trajectory, the measured trajectory and the estimated trajectory, as well as the difference images between the first two calculated trajectories and the measured trajectory from one volunteer study. The blurring in the top right corner of the images is significantly reduced using the measured trajectory and the estimated trajectory. In the difference image between the single-delay trajectory and the measured trajectory, the brightest part is the chest and the back, reflecting a minor image distortion. By using the estimated trajectory, the distortion is mostly corrected, as shown in the difference image between the estimated trajectory and the measured trajectory. The RMSE of the difference images was reduced by $39.4\% \pm 7.9\%$ with the estimated trajectory.

Fat-Water Separation

Figure 5 shows the separated water and fat images in one short-axis breath-held experiment using the spiral-in/out bSSFP sequence. The pixels of the chest wall and the fat around the myocardium are separated from the blood and myocardial pixels. The separation was very robust in each volunteer, except for phase wrapping in some areas. This demonstrates the effectiveness of this method and a potential advantage of the spiral-in/out bSSFP sequence with the refocusing mechanism.

Comparison of the Three Sequences

Figure 6 shows the center line of the 2D PSFs for all three sequences. The main lobes have similar FWHMs, indicating the same spatial resolution. However, the side lobes of the two spiral sequences have much lower values than the radial sequence, indicating that the aliasing is greatly reduced. The calculated aliased energy is reduced by 85% and 82% with the spiral-out and spiral-in/out sequences compared with the radial sequence.

Figure 7 shows systolic and diastolic frames from short-axis and long-axis free-breathing experiments with the radial (top row), spiral-out (medium row) and spiral-in/out (bottom row) bSSFP sequences. In the radial bSSFP images, the signal in some parts of the chest and/or back is missing, which is not observed in the spiral-out and spiral-in/out bSSFP images. The artifact level in the two spiral bSSFP images is much lower than that in the radial bSSFP images. The corresponding videos are included as online supporting information.

SNR & CNR—Figure 8 shows the apparent SNR of blood and the CNR between blood and myocardium for the radial, spiral-out and spiral-in/out bSSFP sequences with short-axis and long-axis views. The SNR and CNR were significantly higher for the spiral-out sequence than for the radial sequence, and for the spiral-in/out sequence than for the spiral-out sequence. The spiral-in/out sequence had the highest SNR and CNR for each volunteer for both short-axis and long-axis views.

Image Quality Rating—Figure 9 shows the blinded image rating results. Each bar shows the average of the ratings for a particular subject imaged with the given sequence in the given orientation, including both reviewers. For the six normal volunteers, the data included results from one breath-held and one free-breathing scan at each orientation. For the patient, the data included three short-axis slices and one long-axis slice, each acquired during free breathing. The spiral-in/out bSSFP sequence had significantly higher ratings than the spiral-out bSSFP sequence, which in turn had significantly higher ratings than the radial bSSFP sequence.

Ejection Fraction—The ejection fractions calculated from images acquired using a gated cine sequence and each of the real-time bSSFP sequences are given in Supplementary Table S1. The EF calculated from the real-time sequences differed from gated cine by 1.5% to 5.4%. Each of the real-time sequences slightly underestimated the EF for each volunteer relative to gated cine.

DISCUSSION

In this study, we implemented a spiral-out bSSFP sequence and developed a new spiral-in/out bSSFP sequence to realize the bSSFP refocusing mechanism. The performance of the two sequences for real-time imaging of cardiac function was compared with a clinically available radial bSSFP sequence using protocols with similar spatial and temporal resolution. Image reconstruction for each of the sequences was performed in real-time using the scanner's image reconstruction computer.

As predicted from the theoretical PSFs, the two spiral bSSFP sequences have much lower aliasing levels than the radial bSSFP sequence. This is because the radial trajectory reduces the scan time by undersampling the outer region of k-space, while the spiral trajectory relies on more efficient k-space coverage per excitation. The outer k-space undersampling percentage is about 50% for the radial trajectory while it is only 20% for the variable density spiral trajectory. The apparent SNR and CNR of both spiral bSSFP sequences were also increased compared with the radial bSSFP sequence. This increase is mainly due to the reduction of the aliasing level, because aliasing increases the apparent noise. The total data sampling time with both spiral sequences was 12% longer than the radial sequence, which also accounts for some of the increase in SNR and CNR. The overall image quality of both spiral bSSFP sequences reflected by the blind rating results also shows that the spiral bSSFP sequences perform significantly better than the radial bSSFP sequence.

For the reconstruction of the spiral bSSFP sequences, we also studied the k-space trajectory fidelity and used a model-based method to estimate the k-space trajectory. The model corrects for the effect of anisotropic gradient delay and linear eddy currents and improves the reconstructed image quality. This demonstrates that the model-based trajectory estimation method is applicable for spiral-in/out trajectories.

The refocusing achieved by the spiral-in/out bSSFP sequence can be used for fat-water separation without additional measurements using the phase cycling property. The result shows a distinct separation between water and fat pixels and thus demonstrates a potential

advantage over the spiral-out bSSFP sequence. However, we also noticed in regions with flow and/or severe B_0 inhomogeneity that the phases of fat and water are sometimes swapped during the entire image series. Further study is required for a more robust separation.

Comparing the two spiral bSSFP sequences, the symmetry property of the spiral-in/out trajectory facilitates the gradient rephaser design and also shortens the minimum TR. The apparent SNR and CNR was higher with the spiral-in/out sequence than the spiral-out sequence. The total data acquisition time and the aliasing level are similar for both sequences; the refocusing mechanism may account for an increase in the SNR and CNR, since the signal intensity at TE becomes stronger with refocused spins. More experiments are required for the validation of this result. The overall image quality of the spiral-in/out sequence is also significantly improved compared with the spiral-out sequence.

The ejection fraction calculated from images acquired using the non-Cartesian real-time sequences was slightly lower than that calculated from images acquired using breath-held gated cine in three volunteers. While the sample size was not large enough to draw definitive conclusions, the lower temporal resolution of the real-time sequences was apparent when calculating end diastolic volume. This effect is likely to be shared by other real-time sequences with similar temporal resolution.

There are several limitations of this study. Spatial and temporal parallel imaging techniques for both spiral and radial sequences were not applied, and thus the comparison was focused on the image acquisition techniques. There have been a number of research studies demonstrating excellent dynamic image quality using radial trajectories with either real-time or retrospective acceleration, but the vendor implementation of radial imaging did not include parallel imaging as an option at the time of this study. The vendor radial sequence in this study used a linear view ordering, which may not be optimal. There were minor streak artifacts in fully-sampled radial images (not shown), which might be due to gradient delays. The amplitude of these artifacts were much lower than typical undersampling artifacts, but a radial sequence with better gradient delay compensation might have performed better. For a fair comparison, parallel imaging techniques were not used to reconstruct the spiral images, and the results show the noise-like aliasing pattern of an undersampled variable-density spiral trajectory. This appears as noise enhancement instead of the streaking artifacts of a radial trajectory.

While the cardiologists reviewed the images blinded to the acquisition pulse sequence, some bias may be possible, because the streaking artifacts are quite conspicuous on the radial images and were easily recognized by one of the image raters. Only one patient was studied, so a larger patient study would be needed to verify the performance of the real-time sequences clinically. The quantitative comparison of ejection fraction demonstrated that each of the real-time sequences achieved reasonable accuracy in estimating ejection fraction, but the study was not large enough to determine the relative accuracy and precision of the different real-time sequences.

While the spiral bSSFP pulse sequences outperformed the radial bSSFP pulse sequence in this study, it may be that radial sequences have advantages in other situations. The shorter TR of the radial sequence in this study would make it more robust in the presence of off resonance, which is especially important at 3T. More generally, the relative advantages of spiral and radial bSSFP pulse sequences will depend upon the imaging application, field strength, and other factors. The tradeoffs will be more complex when incorporating accelerated image reconstruction techniques. However, it seems likely that covering more of k-space using a spiral scan will yield improved image reconstruction results, since the image reconstruction would thus need to estimate fewer missing data samples.

In this study, we applied the spiral-in/out bSSFP pulse sequence to real-time imaging of cardiac function. This sequence could be applied more broadly. For example, it could be adapted to cardiac-gated cine imaging to yield high-resolution movies in a short breath-hold. More generally, Cartesian bSSFP sequences are broadly used for a variety of applications, and the proposed spiral-in/out bSSFP sequence could provide a faster alternative, particularly when paired with parallel imaging. The short readouts of this sequence lead to minimal blur from off-resonance effects and thus only require rapid linear off-resonance correction, thus mitigating one concern with spiral scanning.

In conclusion, we developed a new spiral-in/out bSSFP sequence and demonstrated that it performed better than a spiral-out bSSFP sequence and a radial bSSFP sequence for real-time MRI of cardiac function without image acceleration.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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APPENDIX

Proof of gradient moment nulling in spiral-in/out trajectory design

It is easy to see that $m_0=0$ at the end of TR, since the rephaser gradient is designed such that the spiral-out arm has zero area, and the time-reversed spiral-in arm will thus also have zero area. The 1st order gradient moment m_1 at the end of TR (denoted by T) can be calculated as:

$$\begin{aligned}
 m_1 &= \int_0^T G(u)u \, du = \int_0^{T/2} G(u)u \, du + \int_{T/2}^T G(u)u \, du \\
 &= \int_0^{T/2} G(u)u \, du + \int_0^{T/2} G(T-u)(T-u) \, du \\
 &= \int_0^{T/2} [G(u) - G(T-u)]u \, du + T \int_0^{T/2} G(T-u) \, du
 \end{aligned}$$

where $G(t)$ is the gradient waveform at time t . Based on the gradient design procedure, we can see that $G(u)=G(T-u)$ due to symmetry and $\int_0^{T/2} G(T-u)du=0$; therefore, we have $m_1=0$.

FIG. 1.

Spiral-out bSSFP gradients (left) and the corresponding k-space trajectory (right). The spiral readout gradients are 1.44 ms in length with 720 readout samples. The rephaser gradients that simultaneously null the 0th and 1st order gradient moments are 1.07 ms in length. The rephaser can overlap with the next slice-select prephaser to reduce TR. On the right, the solid and dotted lines correspond to the k-space trajectories of the spiral readout gradients and the rephaser, respectively.

FIG. 2.

Spiral-in/out bSSFP gradients (left) and the corresponding k-space trajectory (right). The spiral readout gradients are 1.60 ms in length with 800 readout samples. The prephaser and rephaser gradients that null the 0th order gradient moments are each 0.68 ms in length. Both the prephaser and the rephaser can overlap with the slice-selection gradient rephaser and prephaser to reduce TR. On the right, the solid and dotted lines correspond to the k-space trajectories of the spiral readout gradients and the prephaser/rephaser, respectively.

FIG. 3.

Comparison of the single-delay trajectory, the estimated trajectory and the measured trajectory for a spiral-in/out bSSFP sequence. Only one interleaved is shown for clarity. The arrows indicate the k-space traversal direction. The estimated trajectory is much closer to the measured trajectory than the single-delay trajectory, especially towards the end of the readout, as predicted by Eq. [2].

FIG. 4. Images reconstructed with (a) the single-delay trajectory, (b) the estimated trajectory and (c) the measured trajectory using the spiral-in/out bSSFP sequence and the difference images between the measured trajectory and the single-delay trajectory (d) and the estimated trajectory (e).

FIG. 5. Separated water and fat images in a short-axis breath-held experiment using the spiral-in/out bSSFP sequence.

FIG. 6.

Theoretical PSFs at the center line for radial, spiral-out and spiral-in/out bSSFP sequences with matched spatial and temporal resolutions. The increased side lobes of the radial sequence lead to prominent streak artifacts in the radial images.

FIG. 7. Short-axis and long-axis free-breathing cardiac images with radial (top row), spiral-out (medium row) and spiral-in/out (bottom row) bSSFP sequences. Along each row from left to right are the short axis systolic image, short axis diastolic image, long axis systolic image, and long axis diastolic image.

FIG. 8. SNR_{blood} (top row) and $CNR_{blood/myocardium}$ (bottom row) of short-axis (left column) and long-axis (right column) cardiac images with radial, spiral-out and spiral-in/out bSSFP sequences. The different bars for each method represent the values computed for six different volunteers. The asterisks indicate statistically significant differences between sequences ($p < 0.05$). The spiral-in/out sequence had the highest SNR and CNR for each volunteer for both short-axis and long-axis views.

FIG. 9.

Average image ratings of short-axis (left) and long-axis (right) views with radial, spiral-out and spiral-in/out bSSFP sequences. The different bars for each method represent the ratings for six normal volunteers and one patient (yellow). The asterisks indicate statistically significant differences between sequences ($p < 0.05$).

Table 1

Sequence Parameters

	TR/TE (ms)	Flip Angle	# of Interleaf	Spatial Res. (mm ²)	Temporal Res. (Hz)
Radial	2.36/1.18	46°	48	3.24 * 3.24	8.83
Spiral-Out	3.76/0.99	50°	32	3.20 * 3.20	8.31
Spiral-In/Out	3.69/1.84	50°	32	3.15 * 3.15	8.47