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Comparison of Centric and Reverse-Centric Trajectories for Highly Accelerated 3D Saturation Recovery Cardiac Perfusion Imaging

Haonan Wang¹, Neal K. Bangerter^{1,2}, Daniel J Park¹, Ganesh Adluru², Eugene G. Kholmovski², Jian Xu³, and Edward DiBella²

¹Department of Electrical & Computer Engineering, Brigham Young University, Provo, UT, USA

²Department of Radiology, University of Utah, Salt Lake City, UT, USA

³Siemens Healthcare USA Inc., New York, NY, USA

Abstract

Purpose—Highly undersampled 3D saturation-recovery sequences are affected by k-space trajectory since the magnetization does not reach steady state during the acquisition and the slab excitation profile yields different flip angles in different slices. This study compares centric and reverse-centric 3D cardiac perfusion imaging.

Methods—An undersampled (98 phase encodes) 3D ECG-gated saturation-recovery sequence that alternates centric and reverse-centric acquisitions each time frame was used to image phantoms and in vivo subjects. Flip angle variation across the slices was measured, and contrast with each trajectory was analyzed by Bloch simulation.

Results—Significant variations in flip angle were observed across slices, leading to larger signal variation across slices for the centric acquisition. In simulation, severe transient artifacts were observed when using the centric trajectory with higher flip angles, placing practical limits on the maximum flip angle used. The reverse-centric trajectory provided less contrast, but was more robust to flip angle variations.

Conclusion—Both of the k-space trajectories can provide reasonable image quality. The centric trajectory can have higher CNR, but is more sensitive to flip angle variation. The reverse-centric trajectory is more robust to flip angle variation.

Keywords

3D cardiac perfusion; compressed sensing; k-space trajectory

INTRODUCTION

First-pass myocardial perfusion imaging provides a powerful and noninvasive method for characterizing ischemic heart disease (1,2). Much of the work in first-pass myocardial

perfusion imaging has been conducted with 2D perfusion techniques (2-8). However, a 3D myocardial perfusion acquisition may be desirable for clinical perfusion studies in order to get more complete coverage of the heart. The 3D perfusion acquisition can potentially provide a more accurate estimation of the size of ischemic zones (1) since it is intrinsically better registered between slices, and trades inter-frame out-of-plane motion problems that can impact 2D imaging for potentially easier out-of-slab motion problems. However, 3D perfusion imaging requires more phase encodes and hence a much longer readout than the 2D case. Because of this longer acquisition time, intra-frame cardiac and respiratory motion will affect image quality more than the 2D case. A highly accelerated acquisition scheme is thus essential for 3D cardiac perfusion to achieve sufficient spatial and temporal resolution. Recent advances in compressed sensing, parallel imaging, and related reconstruction techniques are beginning to enable such rapid acquisitions in 3D cardiac perfusion studies (1,9,10), although spatial and temporal resolutions are not as high as 2D.

A saturation-recovery sequence is often used in first-pass myocardial perfusion imaging to enhance T1 contrast (11). However, with a saturation recovery sequence, the magnetization is often not in steady state when the central portion of k-space is acquired (12-14). Variations in the transient signal level over the course of image acquisition can have a large impact on image contrast.

Kim analyzed this effect in the 2D case (15). He demonstrated that, in a fully-sampled 2D Cartesian acquisition, the T1-weighting due to the saturation recovery pulse is affected significantly by the choice of phase encode ordering in k-space. The dynamic T1-weighted signal can be modified by varying the k-space trajectory, without changing other parameters like saturation recovery time (SRT) and contrast dosage.

Furthermore, cardiac perfusion imaging often employs a very short TR, necessitating very short RF pulses (11,16). This can result in a less than ideal slab selective excitation profile, causing significant deviations from the nominal prescribed flip angle from slice to slice. These variations in flip angle have a significant impact on the signal evolution during the transient, potentially causing considerable cross-slice variations in contrast depending on the phase encode ordering scheme employed. We observed significant such variations in contrast using a centric trajectory. We hypothesized that a reverse-centric trajectory would be more robust to contrast variations across the slab volume (17).

In this work, we explore the effects of k-space trajectory and phase encode ordering using a highly-accelerated 3D acquisition scheme (acceleration factor $R = 11$). In this case, the magnetization may be far from steady state during many of the signal measurements (12-14). As a consequence the k-space trajectory and phase encode ordering employed can be critical factors in determining image contrast for 3D myocardial perfusion imaging, particularly when a rapid slab-selective excitation is employed. We analyze and compare the contrast resulting from a centric phase encode ordering vs. a reverse-centric phase encode ordering in highly accelerated 3D saturation recovery myocardial perfusion imaging. The effect of deviations from the nominal flip angle on signal levels and contrast for each trajectory (centric vs. reverse-centric) is also analyzed.

METHODS

Pulse Sequence

We modified a 3D saturation-recovery TurboFLASH pulse sequence to acquire data at an acceleration factor of $R = 11$ using a polynomial variable density sampling pattern that varied with each time frame (18,19). The sequence was implemented on a 3T scanner (Siemens Medical Systems, Erlangen, Germany). The raw acquisition matrix was 144 (readout) $\times$ 108 (phase encode) $\times$ 10 (phase encode slice direction, including 25% oversampling). An asymmetric echo was employed in the readout direction (83%), and no partial Fourier acquisition was used in the phase encode and slice directions. A non-selective saturation recovery pulse was used each heartbeat. Slab-selective RF pulses with $600\mu\text{s}$ duration were used for excitation of each readout. One oversampled slice at each edge was discarded resulting in an image matrix of $144 \times 108 \times 8$ across a slab with readout FOV 300-350mm, phase FOV 225-263mm, and slab thickness of 40-60mm. This yields a voxel size of 2-2.4mm in the readout and phase directions, and 4-6mm in the slice direction. No start-up pulses were used to drive the magnetization closer to steady state prior to signal acquisition. The acceleration factor $R = 11$ resulted in 98 phase encodes per temporal frame. The fast RF pulse allowed a TR/TE of 2.6/1.1ms. Other parameters employed were: saturation recovery time $\text{SRT} = 150\text{ms}$, bandwidth = 1389 Hz/pixel, and nominal flip angle = 12 degrees. A total of 110 temporal frames (more than necessary to track the contrast uptake) were acquired in all of the studies.

A basic diagram illustrating the pulse sequence is shown in Figure 1. The sequence was programmed to interleave centric and reverse-centric phase encode orderings every other temporal frame. This approach was adopted in order to most directly compare contrast and signal between the two trajectories (15). The total of 110 temporal frames acquired in the studies yielded 55 temporal frames for each trajectory.

Image Reconstruction

The same image reconstruction algorithm was used for both simulated and actual data. The interleaved data set (with 110 time frames) was separated into centric and reverse-centric datasets, each consisting of 55 time frames. Each under-sampled dataset was then reconstructed using spatiotemporal constraints (20) with TV in 3 spatial dimensions and in the temporal dimension. The temporal weight factor was chosen to be 10 times higher than the spatial TV weight factor. All images were reconstructed offline using MATLAB (Mathworks, Natick, MA).

Numerical Phantom Simulations

We performed a numerical phantom simulation of the sequence to estimate the signal levels across a range of T_1 values, with SRT and T_2^* fixed. The effect of flip angle variations across the slab profile on signal level and contrast was also estimated. Image artifact and blurring for each trajectory from transient variations in the signal were assessed. The simulation calculated the expected signal levels for the region of interest at each excitation/readout during the approach to steady state based on equation [1] below (14,21). These

signal levels were then applied as weightings to the k-space data that would be sampled during the corresponding readouts (20).

$$M_{xy}(n) = M \left(1 - e^{TD/T1}\right) \alpha^{n-1} + M \left(1 - e^{-TR/T1}\right) * \frac{(1 - \alpha^{n-1})}{(1 - \alpha)} \quad [1]$$

where

$$M = M_0 \sin(FA) e^{-TE/T_2^*}, \quad \alpha = \cos(FA) e^{-TR/T1},$$

and n is the n^{th} excitation and phase encode.

The simulation process is illustrated in Figure 2. The k-space data was derived from a simple 3D numerical phantom with fixed T1 times. The numerical phantom simulation mirrored the R = 11 acceleration of our actual sequence, and uses the same undersampling pattern, phase encode order, and flip angle variation across the slab profile. To validate the accuracy of our numerical phantom simulation, we performed experiments on phantoms (as described in “Phantom Experiments” below) and compared them to the results predicted from our numerical simulation.

Phantom Experiments

To validate the accuracy of our numerical phantom simulation described above, we acquired phantom data and compared the results to those predicted by the numerical simulation. We imaged two uniform phantoms with measured T1 values of ~300ms and ~1000ms. The phantom with T1 of ~300ms mimics the expected peak signal intensity in myocardial tissue during a perfusion study (15,22,23), and that with T1 of ~1000ms mimics the expected signal intensity in unenhanced myocardium. System noise was measured separately with the same phantom loading and the measured noise level was then applied in our simulations.

We measured the flip angle variations across the slab profile by first measuring the flip angle of the center slice using the dual-angle B1 mapping method (24). A birdcage coil with excellent field uniformity was employed. The flip angle in the center slice was measured to be 12 degrees by the dual angle method. The flip angles at the other (non-center) slices were then determined using the signal equation and the signal intensities of each slice. We expect this technique to yield a more accurate estimate of flip angle than applying the dual angle method to slices where the flip angle is small (the edge slices).

Both phantoms were oriented such that the slice direction was along the long axis of each phantom, which coincided with the main field direction. In cardiac perfusion studies, the relatively homogeneous body RF coil is used for excitation, and thus we expect the flip angle variation to mainly be affected by the frequency characteristics of the RF pulse. For a homogeneous transmit coil, we also expect the slab profile to rotate with variations in the acquisition axis, and expect the variations within a slice to be small, since the slab excitation profile induces variations in flip angle predominantly in the slice direction. We then applied the flip angles measured across the slab profile and the measured T1 values of each vial to the numerical phantom simulation, allowing us to compare the measured contrast vs. the

contrast predicted by the numerical simulation in both the centric and reverse centric cases. Specifically, the signal levels from this phantom experiment were compared to the signal levels predicted by the numerical simulation at the flip angle measured for each slice, as a verification of the accuracy of our simulation.

In Vivo Experiments

The sequence was tested in 4 male volunteers (age = 52, 56, 64 and 66) with approval from the University of Utah institutional review board (IRB) and informed consent from each subject. Imaging was also performed in two canines under an approved IACUC protocol. The highly-accelerated 3D pulse sequence previously described was employed, and 110 temporal frames (55 centric, 55 reverse centric, interleaved as previously described) were acquired after gadolinium injection. Orientation was chosen to yield a short-axis stack of slices. Subjects were instructed to hold their breath as long as possible during the acquisition. Image reconstruction was performed using the spatiotemporal constrained reconstruction. Enhancement curves were generated for manually drawn ROIs in the myocardium in each slice.

RESULTS

Validation of Numerical Simulation using Phantom Experiments

The normalized SNR vs. slice number for both vial phantoms ($T_1 = \sim 300\text{ms}$ and $T_1 = \sim 1000\text{ms}$) is shown in Figure 3. Also shown in the same figure are the normalized SNR values predicted in each slice by the numerical phantom simulation.

The simulation results agree with the measured results. A flatter profile across the slab is predicted by simulation and observed in the phantom experiments when using the reverse-centric trajectory. The centric trajectory, on the other hand, yields better contrast towards the center of the slab where the nominal flip angle is actually achieved, but suffers from large variations in normalized SNR and contrast towards the edges of the slab where the flip angle decreases.

Recall that the simulation SNR were calculated based on the flip angles actually measured at each slice across the slab profile. These measured flip angle values for each slice were (2.5, 4.6, 7.7, 10.4, 11.8, 12, 10.8, 8.4, 5.2, and 2.7 degrees) (Figure 3). The 2.5 and 2.7 degree edge slices correspond to the two oversampled slices. Simulations with ideal RF pulses (no flip angle variation across the slices) are detailed in the supplementary online material, and show the effects when flip angle is increased.

In Vivo Experiments

Figure 4 shows representative 3D perfusion images from one of the human subjects. The average signal intensities for each time frame are shown from manually segmented ROIs in the myocardial wall region for each slice. As expected from the simulation and phantom results, the reverse-centric trajectory yields higher signal intensity than the centric acquisition at the edge slices where the flip angle is attenuated, while the centric trajectory shows higher signal intensity in the center slices. It is also evident that the reverse-centric

trajectory exhibits less variation across slices than the centric trajectory. These results agree with the simulation and phantom experiments. Images from the other subjects are shown in Sup Fig S3 in the online supplementary material.

DISCUSSION

The initial finding that motivated this work was that the signal profiles across slices for centric and reverse-centric 3D cardiac perfusion were quite different, even though the same RF excitation was employed in each case (presumably leading to the same slab excitation profile). This finding is consistent with the slab profile from the centric case being proportional to the sine of the flip angle, and the reverse-centric case being closer to the steady-state spoiled gradient-recalled echo (SPGR) signal that is less affected by flip angle, which follows from equation [1] in the manuscript.

These findings illustrate the different nature of the 3D perfusion acquisition compared to 2D multi-slice methods. Kim compared centric and other trajectories for 2D (15), and the trajectory mostly alters contrast due to the variation in effective SRT and is essentially a 1D weighting or point spread function in the phase encode direction. These in-plane effects are similar for 2D and 3D acquisitions. In contrast, 3D acquisitions also have slab profile effects that interact with the k-space trajectory to determine the contrast of the images. The results here indicate that for relatively standard flip angles and fast RF slab select pulses, the edge slices exhibit much reduced signal relative to the center slices when a centric trajectory is employed.

Reverse-centric yields more consistent contrast when significant variations in flip angle are present across the slab profile, reflecting more of an SPGR-type of contrast, which also depends less on the SRT. The centric trajectory shows higher CNR than the reverse-centric as the flip angle increases, which is also noticeable in the in vivo experiment. However, the increased artifact at higher flip angles limits the flip angle that can be used with a centric trajectory in practice. This is discussed in the supplementary material.

In theory, the centric and reverse-centric cases are identical if the magnetization is immediately at steady-state. Recent works have shown that for a given T1 and SRT, a flip angle can be calculated such that the magnetization is immediately at steady-state and the flip angle is only weakly dependent on T1, as shown and discussed in (25-27). However, since the flip angle varies across the slab volume, it may not be possible for all slices to reach steady-state immediately.

The results are also a function of the oversampling factor and the RF pulse design. A 25% oversampling was used here with 8 slices; 10 slices were acquired but only the central 8 used. This and the RF pulse design will directly affect the flip angle attenuation in the outer slices.

Although the normalized SNR was reported using a separate noise scan, the reconstruction method gives spatially variant SNR that can also depend on the object being reconstructed. Standard SNR measures are often still useful as an approximation.

There are several limitations of the current study. For quantitative results, the scale will directly affect the perfusion estimates in different slices, particularly since a single arterial input function (AIF) from a basal slice is likely to be used. Even if the scaling can be modeled and compensated, the reduced signal and contrast in the edge slices would be expected to decrease the precision of perfusion measurements in the off-center slices. This effect will be greater with a centric acquisition. Further work is needed to determine whether the quantification result can in fact be improved with a reverse-centric trajectory. For non-quantitative application, the scaling may not alter the detection of disease when images are read by physicians, where variations in noise levels and contrast across slices may have a smaller impact on qualitative assessment. However, additional studies in patients with ischemia will be required to determine the impact of flip angle and trajectory on visualization of perfusion defects.

Table 1 summarizes 3D perfusion studies reported to date. Only a few of the studies report the type of trajectory used, which was centric in one case and linear in two others. We briefly explored the effect of a linear trajectory, and as expected the results were between those obtained for the centric and reverse-centric trajectories. Transient artifacts were reduced compared to centric ordering, but the signal level was also reduced, negating some of the positive contrast effects achieved with the centric ordering. Compared with reverse-centric, a linear ordering also preserved more of the flip angle variation. Linear, centric, and reverse-centric all match if a certain flip angle is chosen, though since the flip angle changes with slice, they would not match in all slices. We note too that other types of 3D acquisitions have recently been reported in which the order does not matter since no saturation preparation was employed and the imaging was done at steady state (26,33). Further work is needed to determine the best set of tradeoffs for myocardial perfusion imaging.

We used all of the 110 time frames for the reconstruction. Since the study subjects could only hold their breath for part of the acquisition, respiratory artifacts were expected. For the time curve plot, we only used the first portion of each acquisition where a good breath hold was obtained. The images suffer from significant artifacts if respiratory motion is present. Even with good breath-holds, dark rim artifacts can be present; more rapid, higher resolution 3D readouts would be useful to reduce such artifacts. An improved 3D RF excitation would also be desirable. In this study, the default fast 3D RF slab selective pulse was used, which has a pronounced slab profile. However, achieving a flatter profile will likely require a longer RF excitation, which will decrease temporal resolution and potentially make it difficult to capture the heart when nearly static. Flatter profiles would still show differences for centric and reverse-centric trajectories, but the differences would be expected to be less pronounced.

CONCLUSION

Both centric and reverse centric k-space trajectories can be used to successfully perform highly undersampled 3D cardiac perfusion imaging. However, the different trajectory orderings offer different trade-offs. The centric k-space trajectory can provide higher CNR for the acquisition as compared to the reverse-centric trajectory, but is more sensitive to flip angle variation between slices. On the other hand, the reverse-centric trajectory is more

robust to flip angle variations, which is important for 3D quantitative analysis. Further work is needed to evaluate the impact of these differences for specific tasks.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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FIGURE 1.

Illustration of the acquisition scheme for the comparison between the centric and reverse-centric trajectories. The trajectories were hard coded to alternate at each heartbeat. The lower part of the figure illustrates a small section near the center of k-space to clarify the ordering of the k-space trajectories. The number in each sample (circle) shows the order of the phase encodes, and is based on the distance from the k-space center. Note the figure is approximately to scale ($\Delta ky = 0.003\text{mm}^{-1} - 0.0035\text{mm}^{-1}$, $\Delta kz = 0.017\text{mm}^{-1} - 0.025\text{mm}^{-1}$, with small variations due to the size of subjects). The slice direction k-space sampling is much larger than the phase direction.

FIGURE 2.

Illustration of the simulation scheme for a single time frame. We first generate a vial-shaped numerical phantom and assume that each slice receives a different flip angle based on its position. We then weight the signal intensity of each slice to account for both flip angle and calculated signal level and calculate the k-space values (we permute readout and phase encode directions after the 3DFT for illustration purposes). Then, following the undersampled k-space trajectory, we acquire one 3D readout line for each RF pulse. We repeat this process 98 times to simulate the acquisition for one time frame. The figure shown here portrays the centric ordering; we also simulated the reverse-centric ordering in the same way, but with reversed acquisition order.

FIGURE 3.

Comparison of simulation results to actual phantom data across slices for normalized SNR (left and middle) and CNR (right). Results for the centric k-space trajectory are shown in the top row; the reverse-centric trajectory results are shown in the bottom row. Actual measured flip angles across the slab were applied to the simulation for comparison. The simulation agrees with the measured results. The table at the bottom of the figure shows the flip angle measured at each slice.

FIGURE 4.

Comparison of in vivo results for centric and reverse-centric trajectories. The images shown are the peak contrast enhanced frame of myocardial wall. The time curves shown beneath each image show the signal level change over time from an ROI in the myocardial wall region.

Table 1
Summary of 3D cardiac perfusion papers

	Year, first author and reference number of paper	Field Strength	Acceleration method	TR/TE/SRT/ Acquisition window (msec)	Acquisition pattern/Trajectory
1	2008 Shin, T (1)	3T	6× SENSE	2.3/0.9/130/304	Cartesian/ not given
2	2011 Vitanis, V (10)	3T	10× compartment based k-t PCA	1.8/0.7/150/225	Cartesian/ not given
3	2011 Manka, R (28)	3T	6.3× k-t SENSE	1.8/0.7/150/200	Cartesian/ not given
4	2012 Jogiya, R (29)	3T	10× k-t PCA	1.8/0.7/150 /not reported	Cartesian/ not given
5	2012 Manka, R (30)	1.5T	10× k-t PCA	1.9/0.8/150 /not reported	Cartesian/ not given
6	2012 Chen, L (25)	3T	stack of radial TV regularization	2.1~2.9/1.1~1.4 /140~160/300	stack of radial/ centric in kz
7	2012 DiBella, E (26)	3T	stack of radial TV regularization	2.4/1.3/without sat/310	stack of radial/ not relevant
8	2013 Shin, T (31)	1.5T	stack of spiral k-t SENSE	9.8/7.1/140/230	stack of spiral/ not given
9	2014 Akcakaya, M (32)	1.5T	10× two-step patch-based procedure	2.1/1.2/100/250	Cartesian/ Centric
10	2014 Giri, S (33)	1.5T	3× GRAPPA	2.7/1.04/without sat/300~380	Cartesian/ not relevant
11	2014 Schmidt, J (34)	3T	10× k-t PCA with motion correction	2~2.2/0.9~1/140 /205-225	Cartesian/ Linear
12	2014 Motwani, M (35)	3T	10× k-t PCA	1.8/0.7/150/192	Cartesian/ Linear
Notes	<ul style="list-style-type: none"> ■ SPGR with saturation recovery was applied in study 1-6 and 8, 9, 11, 12. SPGR without saturation recovery was applied in study 7. b-SSFP without saturation recovery was applied in study 10. ■ No start up pulses mentioned in study 1-9 and 11,12. Dummy pulses were applied in study 10 to maintain steady state. ■ Study 9 reports the use of the centric order and 11, 12 use linear order. 				