Centric and reverse-centric trajectories for undersampled 3D saturation recovery cardiac perfusion imaging

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INTRODUCTION: First-pass myocardial perfusion imaging provides a powerful noninvasive method for characterizing ischemic heart disease. Compared with traditional 2D perfusion techniques, 3D myocardial perfusion acquisition is desirable for clinical perfusion studies in order to cover the whole heart. A 3D acquisition can also provide more accurate estimation of the size of ischemia [1]. Moreover, 3D perfusion imaging is intrinsically better registered between slices and avoids inter-frame out-of-plane motion problems that can impact 2D imaging. However, 3D imaging requires a much longer readout than the 2D case, so a highly accelerated imaging acquisition scheme is essential to achieve sufficient spatial and temporal resolution. Currently, compressed sensing and parallel imaging can enable highly accelerated imaging [2]. However, with a saturation recovery sequence, the magnetization often does not reach steady state. As a consequence the k-space trajectory is a critical factor in determining image contrast. In this study, we analyze and compare the contrast resulting from a centric phase encode ordering vs. a reverse-centric phase encode ordering in highly accelerated 3D myocardial perfusion imaging.

METHODS: We modified a 3D saturation-recovery TurboFLASH pulse sequence to acquire data at an acceleration factor of R = 11 using a variable density phase encode mask [4]. The sequence was implemented on a 3T Siemens Trio whole-body scanner (Siemens Medical Systems, Erlangen, Germany). The acquisition matrix was 144 x 108 x 8 with 110 temporal frames. With an acceleration factor R = 11, there were 98 phase encodes per temporal frame. A fast RF pulse was used allowing TR/TE = 2.6/1.1ms. Other parameters were TD = 150 ms, bandwidth = 1152 Hz/pixel, and flip angle = 12 deg. The sequence was programmed to interleave centric and reverse-centric phase encode orderings every other temporal frame to allow a fair comparison, yielding 55 temporal frames for each trajectory [2].

Simulations: We performed a numerical simulation of the sequence to estimate the signal levels across a range of T1 values. Image contrast was assumed to be defined by the simulated signal level during the first phase encode in the centric case and the final phase encode in the reverse centric case. Since a slab-selective excitation was used, a non-uniform slab excitation profile is expected, yielding variations in flip angle from slice to slice. These variations in flip angle across slices will affect the images differently, depending on the order of the k-space trajectory. Simulation and actual acquired phantom data (described below) were used to characterize the slab profile. Finally, a more sophisticated simulation was performed that applied the appropriate signal weightings in k-space to a simple simulated heart mimicking the left and right ventricular blood pools and myocardium. Image reconstruction was performed using an STCR algorithm with total variation (TV) applied as the constraint term in both the temporal and spatial directions [5].

Phantom Experiments: A uniform phantom with T1~1sec was imaged in a uniform sensitivity bird-cage coil using our sequence to determine the resulting slab excitation profiles. The measured slab profiles were matched in the two simulations by varying flip angles. We then imaged a phantom containing an array of vials with different concentrations of Gd-DTPA. The vials were oriented such that the slice direction was along the axis of each vial and the slab profile for vials with T1~2500 and 250msec were measured.

In Vivo Experiments: The sequence was tested in vivo on a healthy volunteer using a 32-channel coil. The flip angle was reduced to 10 degrees, and the number of temporal frames was reduced to 30 for both the centric and reverse centric ordering. The orderings (centric and reverse centric) were again interleaved for comparison of contrast after a single injection. Image reconstruction was performed offline, again using an STCR algorithm with TV applied as the constraint term in both the temporal and spatial directions.

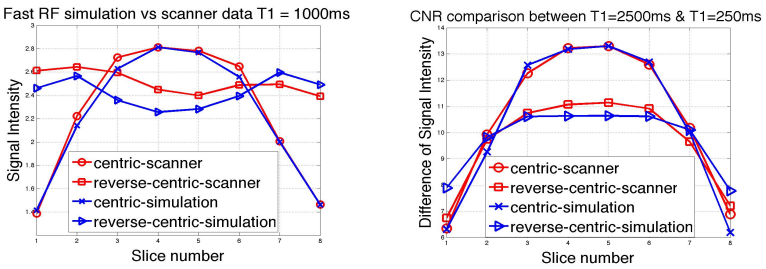

Figure 1: Comparison of simulation results and real data for slice profiles (left) and CNR (right). To make the simulation match the phantom results, the flip angle varied more than 50% from the center to the edge for the T1=1000 ms phantom and for the contrast between the T1=2500 and 250ms phantoms.

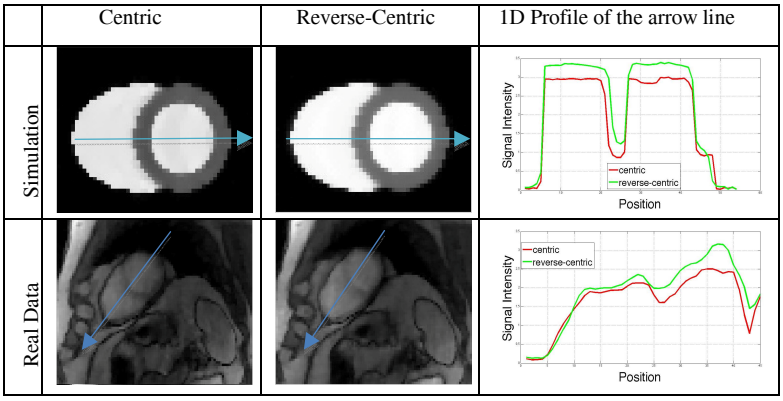

Figure 2: Simulation result from piece-wise constant heart phantom and in vivo data. The first column is the centric result, the second column reverse-centric, and the third column is the 1D profile of the signal intensity along the direction from left ventricle to myocardial wall, as indicated by the blue arrow on each graph.

RESULTS: As expected, the signal level for a given vial varies from slice to slice due to the inhomogeneous slab excitation profile (Fig. 1). The effect of the inhomogeneous slab excitation appears more pronounced in the centric phase encode ordering scheme, since it is proportional to sin(alpha). The impact of flip angle variations is less pronounced in the reverse-centric ordering, where the center of k-space is acquired when the signal is closer to steady state. Our simulation model matches the actual measured data closely (with flip angles of 6, 9.1, 11.2, 11.8, 11.7, 10.9, 8.5, and 6.2 degrees corresponding to these 8 slices). The centric ordering provides higher peak but less uniform signal for the parameter set used. In Figure 2, from the simulated heart result and the real scan data, the reverse centric ordering has higher signal intensity than the centric, presumably due to the different flip angles in vivo.

DISCUSSION: In highly accelerated 3D imaging, much of the data is sampled when the signal is not in steady state. These transient effects are particularly pronounced in centric phase-encode ordering schemes, and have a significant impact on image contrast. Furthermore, they heighten contrast variations from slice to slice due to inhomogeneous slab excitation profiles, because the transient signal intensity is sensitive to the flip angle. This has particular implications for quantitative perfusion measurements, and is in addition to trajectory effects seen in 2D perfusion imaging [3]. When we apply a 3D acquisition for cardiac perfusion, it is important to consider the flip angle, saturation recovery time, and trajectory and find the best combination.

REFERENCES: [1] T Shin et al., JCMR 10:57, 2008. [2] Lustig M et al. MRM 58:1182-95, 2007. [3] D Kim, MRM 59:202–208, 2008. [4] L Feng et al, ISMRM, 2011. [5] G Adluru et al., JMRI 29:466–473, 2009