Adaptation of Coronary Imaging Pulse Sequences for Self-Calibrated Non-Cartesian Parallel Imaging

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Introduction:

Speed and timing are critical determinants of the image quality of coronary magnetic resonance angiography (MRA). In navigator gated/corrected, free-breathing coronary MRA, a subject-specific trigger delay and acquisition window can be used to reduce motion-related artifacts. Many existing coronary MRA pulse sequences were designed to work under these timing constraints. The advent of parallel imaging warrants a re-examination of these pulse sequence designs. While parallel imaging has generally been used to accelerate overall data acquisition speed, it provides an added design parameter in coronary imaging where not only speed but also timing and data acquisition duration are crucial. Parallel imaging now has the capability to perform self-calibrated reconstruction on MR data acquired with non-Cartesian trajectories [1], making many existing coronary MRA sequences easily adaptable to parallel imaging. Because we no longer need to acquire a separate sensitivity calibration scan or to modify the trajectories, accelerated images can be directly compared to the non-parallel counterparts. In this work, a 3-D spiral and a 3-D radial MRA sequence were tested with various lengths of data acquisition window, and the overall scan time was kept relatively constant by applying the appropriate acceleration factors with parallel imaging.

Methods and Materials:

Accelerated images were reconstructed using Self-calibrating Parallel imaging with Augmented k-Space radius (SPARS) [1]. SPARS extracts low-resolution *in-vivo* coil sensitivities from the densely sampled central *k*-space region. SPARS then employs a *k*-space locality constraint to perform parallel image reconstruction on the non-Cartesian MR signal data. For consistency, a *k*-space radius of 2 was used for this work.

All scans were performed on a 1.5T Gyroscan ACS-NT whole body MR system (Philips Medical Systems, Best, NL). MR signal data of coronary arteries of healthy volunteers were acquired with a 5-element Synergy array. Two ECG-triggered, navigator-gated/corrected non-Cartesian MRA sequences were investigated: 1) 3-D spiral sequence: 42-interleaf spiral/ 2 interleaves per RR interval/ flip angles 45°-60°/ sampling window 70ms; 2) 3-D radial sequence: 368 projections/ balanced TFE/ TR 5.6ms/ flip angle 110° / sampling window 200ms. Both sequences used conventional phase encoding in the slice direction (10 slices). The two sequences were modified with various undersampling factors and acquisition window durations, resulting in different overall acceleration factors (Tables 1 and 2). Undersampled data sets were reconstructed using SPARS, and reference data sets were reconstructed with a conventional regridding algorithm.

Results:

Image reconstructions corresponding to the entries of Tables 1 and 2 are shown in Figures 1 and 2.

Fig 1: Spiral Reconstructions

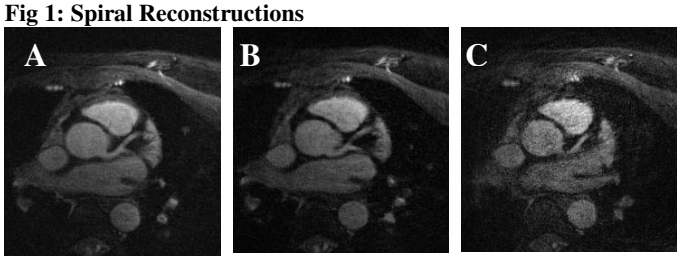

Fig 2: Radial Reconstructions

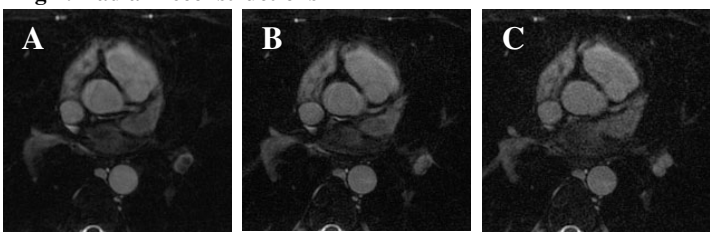

Table 1 # of Spiral Interleaf	# of Interleaf/ RR Interval	Acquisition Window	Under-sampling Factor	Overall Acceleration
a) 42 (ref)	2 (45°/60°)	70 ms	1.0	1
b) 21	1 (90°)	35 ms	2.0	1
c) 14	1 (90°)	35 ms	3.0	1.5

Table 2 # of Radial Projections	# of Project/ RR Intervals	Acquisition Window	Under-sampling Factor	Overall Acceleration
a) 368 (ref)	25	200ms	1.0	1
b) 250	17	140ms	1.4	1
c) 123	12	100ms	3.0	1.5

Figure 1**a** and **b** display comparable baseline SNR despite the fact that **b** was reconstructed from a two-fold undersampled dataset. This is consistent with our expectation because unlike **a** which splits the magnetization energy between the two spiral interleaves in the same RR interval, **b** can take advantage of a full 90° RF excitation. Besides the g-factor penalty, there is no theoretical SNR loss between the two. **b** also carries an advantage that every acquisition receives the same magnetization energy, whereas **a** needs to address the effect of T_1 recovery between the two RF excitations (e.g. the 45°/60° pulses). **c** shows an noticeable SNR loss compared to **b** in exchange for an accelerated acquisition.

Figure 2, **a** and **b** display apparently similar baseline SNR. However, the subtle differences in the images reflect the tradeoffs: **a** is more sensitive to motion-related artifacts; **b** has regions of poor RF coil coverage which lead to suboptimal parallel imaging performance. A third factor that may have played a role is the diminishing effectiveness of a fat-saturation pulse in projections acquired late in the RR interval. **c** shows a modest SNR loss compared to **a** and **b** but has the shortest acquisition window, advantageous in overcoming beat-to-beat heart rate variations.

Conclusion:

This work has demonstrated the feasibility and potential benefits of adapting coronary MRA sequences to parallel imaging. As parallel imaging moves forwards to higher acceleration factors (e.g. 16-fold) with new coil array designs [2], parallel imaging will be an increasingly important parameter in future coronary imaging sequence optimization.

References: 1. Yeh, EN. et al. ISMRM 2002:2390. 2 Zhu, Y. et al. ISMRM 2003:22.