Motion Correction in High-Resolution Coronary MRI Using Measurements from in-situ Tracking and Imaging Catheters

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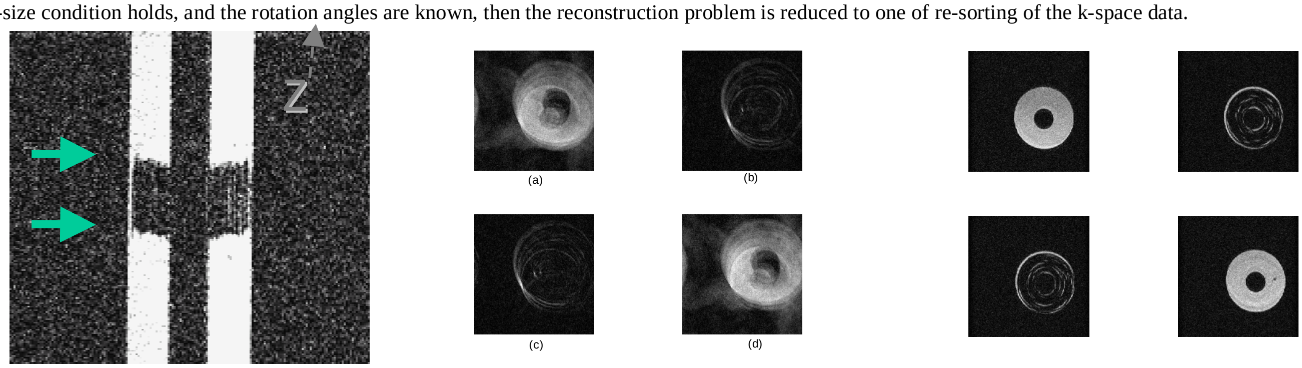
Abstract:

Cardiac and respiratory induced motions (~2 cm/cycle) are significant impediments to high resolution MR coronary artery imaging. Motion reduction techniques based on breath-holding and three-dimensional navigators may be inadequate for high resolution (<500 microns) vessel-wall or luminal imaging. Recently, an MR tracking method that can be used to measure the motion of coronary arteries in the frequency, phase and slice encoding directions has been proposed and demonstrated [1,2]. This 1-2mm diameter catheter consists of 3 MR-Tracking coils placed at 25 mm distances from the tip of the catheter. When placed in the coronary arteries, this arrangement is capable of measuring both translational and rotational motion at different segments of the coronaries. We propose a post-acquisition technique to remove the effects of both translational and rotational motion from raw k-space data based on measurements derived from *in-situ* catheter-based MR tracking coils in the coronary.

Methods

A 3D gradient recalled echo (GRE) sequence with parameters TR/TE/θ=14.4ms/4.3ms/30^0, 6x6cm FOV, 256x256, 8 Views-Per-Segment (VPS) or 8 phase encoding lines per cardiac RR cycle, 16 partitions/slab, 2 mm thick slice, 1 NEX data was collected on a double-lumen "plaque" phantom with plaque sections created with thin water-filled voids (between sections of Mylar tape). MRI data was acquired with the phantom's luminal axis parallel to the GRE z-axis as shown in Figure 1A. From a pig's right coronary artery (RCA) translational data obtained over an RR cycle (HR 85 BPM) during a single breath-hold with the intra-vascular (IV) tracking coils/catheter arrangement, motion in the cardiac cycle was identified and fitted with a polynomial function. In a standard linear k-space coverage, the general translational motion function can be represented as $x(t)=\Sigma_n 1/n!x^n(0)t^n$ (n=0, 1,..,∞), where the $x^n(0)$ corresponds to the initial value of the nth order moment of a moving spin system. In a conventional 3D Fourier technique, in the frequency-encoding direction, phase accrual is at the digitizer rate (Δt), whereas for both the in-plane phase-encoding and slice phase-encoding directions, it is at intervals of TR, and extends to TR x VPS. The motion dynamics of the coronary are such that within the quiescent region of the cardiac cycle, the second (acceleration) and higher order moments are negligible (<5%). Thus, the phase accrual (phase and slice-select directions) and frequency shifts (frequency direction) calculated from the velocity profile of the measured coronary displacement data were used to corrupt the vessel "plaque" phantom k-space data prior to reconstruction. A complementary phase function derived from the fit of the displacement data was then used to compensate the raw k-space data in the frequency, phase and slice selective directions. In the case of rotational motion, the catheter measurements may be represented as the vector ($\mathbf{r}$,φ,θ). The assumption of rigid body motion (coronary shearing and stretching are assumed negligible) requires that the radial vector $\mathbf{r}$ remains constant. With this assumption, for each set of VPS (taken in the quiescent region), the k-space trajectory is rotated by an angle φ, in-plane, and θ, out-of-plane (the slice encoding direction), and invoking the Affine transformation property of Fourier transforms, each rotation of a projection of an object about its center in one domain results in the same rotation of the respective inverse transform domain [3]. Hence, for each k-space trajectory, we simply rotate it both in the in-slice and out-of-slice directions using the angular displacement measurements. This compensation scheme requires that the spatial extent of the acquired volumetric slab bounds the dynamic range of the coronary motion, since partial volume effects would dominate and make compensation impossible. Relaxation effects must also be addressed in that case. If this slab-size condition holds, and the rotation angles are known, then the reconstruction problem is reduced to one of re-sorting of the k-space data.



B C

Figure 1: A) Sagittal slice through a plaque phantom showing that the " Plaque " region extends between the green arrows. Grey dashed arrow defines acquisition Z-axis. 3D GRE Axial slices were collected in the region between the green arrows. B) With 10 times exaggerated in-plane rotational motion in the 100-220 msec post-RR window (a ~0.5 radians total rotation). Artifacts similar to projection reconstruction appear. Periodic motion results in radial streaking including smearing, blurring and geometric distortions. Note that the thin (~250 micron) white partitions (" plaque") can be resolved in the motion corrected images in C.

Results:

The results of corrupting the phantom with translational and rotational motion as measured from a pig's RCA are shown in Fig. 1B. Since translational motion results in phase accrual in k-space, this effect manifests itself as ghosting in the phase encoding direction. The most problematic motion is that due to rotation. The effect of rotational motion on a standard 2D spin warp k-space coverage is to introduce non-uniformity in the k-space data. The result is a rotation of each acquisition projection in k-space resulting in a misregistration or smearing of spatial frequencies. This smearing in k-space is reversible if the rotation vector is known per VPS acquisition by inverse rotating, gridding and Fourier transforming each data set resulting in Figure 1C.

Conclusions:

Coronary translational and rotational motion compensation based on utilizing an MR tracking catheter has been demonstrated on phantom data utilizing in-vivo motion dynamics from a pig's RCA. Future work will involve correction of motion-corrupted coronary artery images. It will utilize detailed motional data, acquired using *in-situ* MR-tracking catheters.

References:

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