Retrospective estimation of 3D respiratory motion vectors in coronary MRI

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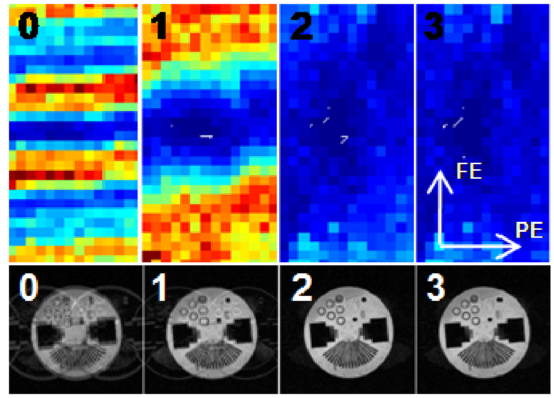
Introduction: Diaphragm navigator gating and tracking is an accepted technique for respiratory motion-compensation in free-breathing cardiac MR. Navigator tracking was developed in an effort to increase the gating window and hence navigator efficiency. It is typically performed using a fixed tracking factor of 0.6 for all patients [1, 2], and not accounting for differences between patients [2]. Although, navigator gating is successful in majority of data acquisitions, frequently it results in very long data acquisition or unsuccessful completion due to patient motion or drifting. Furthermore, the severity of motion artifacts manifest differently among coil elements depending on proximity of each element to the motion area, e.g. posterior coil suffers less motion artifact originating from the chest wall. This suggests it might be beneficial to use a coil specific correction. In this study, we propose to use the correlation between the navigator signals to k-space phase to estimate a patient-, direction- and coil-specific motion vector that can used to correct multiple-averaged data. Subsequently, the phase corrected k-space data will be used in combination with a larger navigator window to prospectively reconstruct the images.



0.000 0.000	0.696 -0.004	0.981 -0.009	1.009 -0.025

Fig 1: The top row shows variance of phase data across three averages of a 2D scan of a phantom undergoing inter-shot motion after 0,1,2, and 3 stages of iterative motion-estimation and phase correction. Bottom row shows the corresponding images. The table shows the estimated factors in the vertical and horizontal directions; the true value is (1,0).

Method: In *k*-space, translations of the anatomy between different acquisitions of the same data points cause a proportional phase difference, though this is wrapped around 2π. For each individual coil, we estimated the respiratory motion vector α that best fit the observed k-space phase differences: , where is the navigator value when the *i*th acquisition of a given k-space point was obtained. We would like to perform least-squares estimation of the three components of α, which corresponds to the three directions of the motion. However, the phase errors are greater than 2*π* at high spatial frequencies, so the linear relation between navigator and phase differences does not hold except near the center of *k*-space. We address this problem by using an iterative scheme. At each stage, we perform a least-squares estimate to fit the phase differences in that part of *k*-space with no 2*π* phase wrapping. We then apply phase-correction to all the 3D volumes using our estimated motion vector, which partially corrects the phase errors. The procedure can then be repeated using data from a larger range of k-space lines. After phase correction based on the respiratory motion, we detect and correct for bulk motion between acquisitions. Finally, the phase-corrected *k*-space data acquired within the reconstruction gating window of 10mm were averaged.

All scans were obtained on a 1.5T clinical MR scanner (Philips Healthcare, Best, Netherlands). Phantom data were acquired with the body coil, using multi-shot with 30 phase-encode lines per shot, centric phase-ordering, and manual motion of the phantom after every two or four shots. Each acquisition required four shots to complete, and 3 acquisitions were obtained. A projection navigator was placed on the top edge of the phantom to measure the motion in the frequency encoding direction. For the *in vivo* study, targeted coronary scans of four healthy adult subjects were obtained during free-breathing with a 5-element cardiac coil and a pencil-beam navigator placed on the right hemidiaphragm. An ECG triggered, free breathing SSFP with TE/TR/θ=4.3/2.1/90° was used. Three acquisitions (averages) were obtained to image the right coronary arteries. The respiratory navigator was enabled, but all data were accepted.

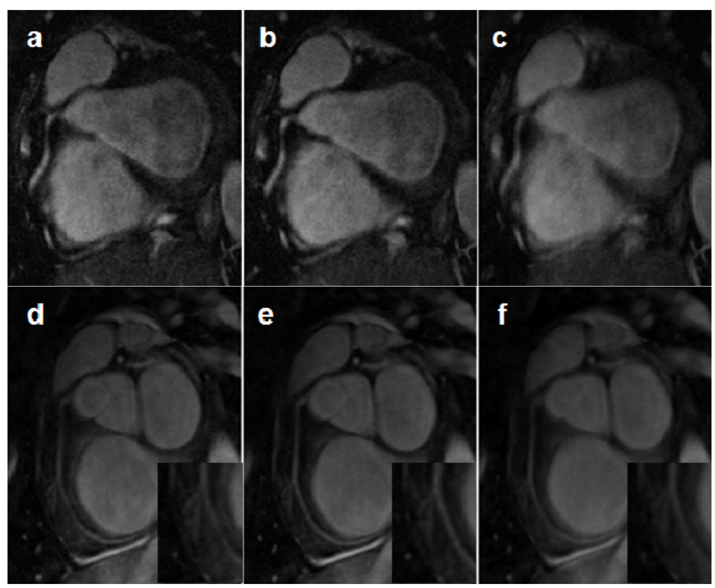


Fig 2: Single slices of 3D coronary-targeted scans of two healthy subjects reconstructed using: (a,d) retrospective gating with a 5mm gate, (b,e) tracking and correction in a 10mm gate using the coil-specific motion vectors estimated from 3 acquisitions, and (c,f) unweighted average of 3 acquisitions. Shown is the root sum of squares of the coil images.

Results: As shown in Figure 1, without phase correction, the phantom image reconstructed by averaging all data within the 10mm gate exhibits ghosting artifacts (panel 0). The successive panels show how the 2D motion vector is iteratively refined towards the true value of 1 in the vertical direction and 0 in the horizontal direction, by applying the corresponding phase correction to reduce the phase variations between the acquisitions. Because the coronary-targeted scans were preformed with a double-oblique slice orientation and the principal direction of respiratory motion is SI, all three components of the estimated motion vector were calculated to be nonzero. Figure 2 shows phase correction using the estimated motion tracking vectors yields images (b,e) with less blurring than simple averaging (a,d), but comparable to a 5 mm retrospective gated image (c,f).

Conclusions: We presented a novel patient-, direction-, and coil-specific respiratory motion vector estimation method that can be used to correct the motion-induced phase variation.

References: [1] Wang, MRM,1995. [2] Danias, AJR 1999; [3] Scott, Radiology, 2009.

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