A Real-Time Motion Compensation Package and Active Marker Headband for Brain MRI

M. B. Ooi[1], S. Krueger[2], W. J. Thomas[3], S. V. Swaminathan[4], and T. R. Brown[1,3]

[1]Biomedical Engineering, Columbia University, New York, New York, United States, [2]Philips Research Europe, Hamburg, Germany, [3]Radiology, Columbia University, New York, New York, United States, [4]Philips Medical Systems, Cleveland, Ohio, United States

Introduction: Even a few millimeters of patient movement during an MRI exam can result in motion artifacts, manifested as blurring or ghosting, that render the images diagnostically unusable. Often, in subject populations with a high potential diagnostic benefit, e.g. young children, elderly subjects, and patients with dementia, the utility of MRI is curtailed by motion artifacts, underlining the importance of an effective motion correction strategy. Several prospective compensation techniques for brain MRI have been proposed – including image-based methods (1), 3D-navigators (2-4), and optical markers (5). Each technique uses different reference data to solve for the six degrees of freedom (6-DOF) needed to define rigid-body motion. Here, we propose the use of active markers for real-time motion tracking.

The tracking of micro RF-coil "active markers" was first introduced in 1986 (6), and later used for motion correction in an "offline" inter-image update scheme (7). Recently, this has been extended into a real-time, intra-image compensation strategy (8), with preliminary results demonstrating quantitative improvement in image quality in moving phantoms (9). The current work presents a complete prospective correction package for brain MRI that requires minimal additional hardware and can be flexibly incorporated into a large family of imaging sequences, promoting transferability to the clinical setting.

Methods: Experiments were performed on a 1.5T Philips Achieva (Philips Medical Systems, Best, The Netherlands). The setup for a motion corrected brain exam is shown in Fig. 1. A headband integrated with three active markers is securely placed on the forehead of a volunteer, maintaining a fixed location relative to the head so that any head movement is reflected in the active markers (three points in 3D-space are sufficient to uniquely solve for the 6-DOF). The multi-channel MR system allows for parallel imaging using a standard bird-cage coil (Conn. A), and position tracking with our active marker headband (Conn. B). The tracking-coils were interfaced to the scanner via a vendor-provided Synergy Multi-Connect (SMC) Box. The active marker design and tracking pulse-sequence used for localization has been detailed in a separate submission (current conference), as well as in previous work (10); within the space defined by the bird-cage coil, validation tests have shown marker precision of <10μm.

At scan start, the initial locations of the active markers are determined. The tracking sequence is then interleaved into the imaging sequence at a user-adjustable rate. Comparing current active-marker positions with the initial reference locations, the 6-DOF rigid-body motion is determined (11) and fed back to update the gradient currents – thereby maintaining the image-plane at a fixed orientation relative to the head – before the next segment of k-space is acquired. In cases of extreme/abrupt motion, corrupted lines of k-space are rejected and reacquired with the updated geometry (8). This iteration of tracking and imaging continues until all of the desired k-space is acquired. Images are conventionally reconstructed on the MR system, and immediately available for viewing.

Prospective correction was applied to a 3D-MPRAGE scan (TE/TR/TI/shot interval = 4/8.3/1000/1500 ms, $\theta = 8°$, FOV = 240×240×100 mm, voxel size = 1.25×1.25×2 mm, slices = 50, turbo factor = 48) to establish the scheme's effectiveness in even complex sequences; tracking and geometry updates were interleaved every 48 k-space phase-encodes (400 ms of image data acquisition). Two cases were tested, i) with the volunteer at rest, and ii) performing a reproducible ±5º left-right headshake motion throughout the entire scan. For each case, two scans were acquired – with correction ON and OFF – in order to evaluate improvements in image quality under similar motion conditions.

Results: Comparison of Fig. 2c vs. 2b demonstrates the dramatic improvement in image quality achieved in the presence of motion, while comparison of Fig. 2b with 2a illustrates virtually perfect correction. A quantitative metric of image quality based on the signal power in the image at high spatial frequencies (9) was calculated for each image volume (Fig. 2a-c), and represented as a percentage Q by normalizing to the at rest, uncorrected volume (images not shown). Results established substantial improvement for both motion corrected scans (Q > 100%), quantitatively confirming positive visual inspection of image quality. It is interesting to note that this metric detected an improvement in image quality with correction ON even with the volunteer at rest, suggesting correction for physiological motion; active marker tracking information during at rest scans contained a frequency cycle matching the respiratory bellows monitor, confirming that the dominant motion was respiration in these cases. The current package has also been tested in both standard/turbo spin-echo and gradient-echo sequences, with very similar results.

Discussion & Conclusion: This work demonstrates utilization of active markers in a successful motion detection and real-time correction package for rigid-body motion. High quality images are acquired in the presence of motion, clearly showing the benefits of this approach. The correction scheme allows for intra-image motion compensation, i.e. between single or multiple k-space lines. In contrast, a real-time image-based correction approach (PACE) (1), is only capable of correcting for inter-image motion, such as between image volumes in an EPI time series. In both PACE and 3D spherical navigator techniques (2,3), long temporal delays (~ 2 s) are required for complicated motion detection measurements and registration calculations, limiting factors in real-time applications. In contrast, active markers require only a short tracking pulse-sequence and computationally simple correction calculation. The total delay is 37 ms, on-par with cloverleaf (4) and optical (5) strategies (18.2 ms and 32 ms, respectively), but without the need for cloverleaf's 12 s initial reference scan (during which the subject must remain motionless), or the additional hardware requirements of optical tracking. Unlike 2D-PROPELLER (12), our approach is not confounded by through-plane motion and is likely to provide significant improvement in image quality in virtually all situations, even when no excessive motion is present. Utility in other MR applications, for example EPI-based sequences such as DWI, DTI, ASL, and fMRI, will be avenues of future investigation.

Fig. 1

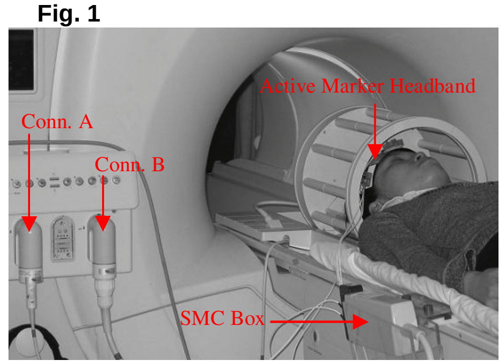
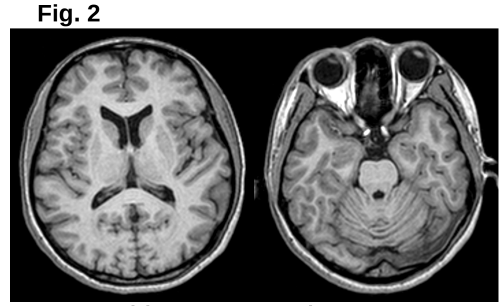
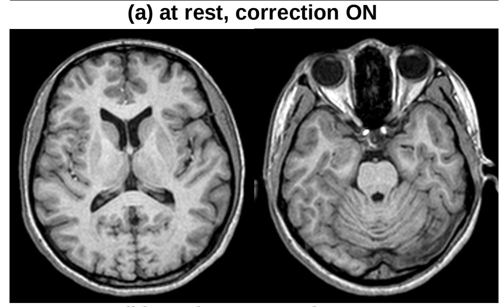


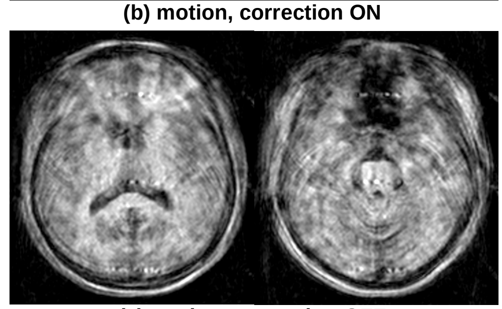
Fig. 2



(a) at rest, correction ON



(b) motion, correction ON



(c) motion, correction OFF

Grant sponsor: National Institute of Health; Grant number: R21EB006877

References: [1] Thesen S et al., MRM 2000;44(3):457-65. [2] Welch EB et al., MRM 2002;47(1):32-41. [3] Ari N et al., Proc. 14th ISMRM 2006;3195. [4] van der Kouwe AJW et al., MRM 2006;56(5):1019-32. [5] Zaitsev M et al., NEUROIMAGE 2006;31 (3):1038-50. [6] Ackerman J et al., Proc. 5th ISMRM 1986;1131-32. [7] Derbyshire JA et al., JMRI 1998;8(4):924-32. [8] Krueger S et al., Proc. 14th ISMRM 2006;3196. [9] Ooi M et al., Proc. 16th ISMRM 2008;209. [10] Krueger S et al., IEEE Trans Med Imaging 2007;26(3):385-92. [11] Umeyama S, IEEE PAMI 1991;13(4):376-80. [12] Pipe JG, MRM 1999;42(5):963-69.