High Precision Measurement of Micro-Coil Locations for Real-Time Tracking Applications

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Introduction: The ability to track the location of multiple point-sources in the MR scanner is of great utility in various applications. Monitoring the position of a micro RF-coil using magnetic resonance was first introduced (1), and later developed as a means of actively tracking medical devices, such as the tip of a catheter or a surgical tool, during MR-guided invasive procedures (2-4); with multiple "active markers" attached, both position and orientation of the device can be determined. Marker-based methods have also been used to detect movement and compensate for motion-induced artifacts both retrospectively (5-7), and prospectively (8). Recent work has demonstrated a real-time, intra-image motion correction package employing active marker tracking (9), with results showing image quality improvement in both phantom and brain scans acquired in the presence of substantial motion (10).

The success of these marker-based applications is dependent upon the quality of micro-coil position measurements. The current work presents a method of tracking multiple active markers with high accuracy (0.30 ± 0.40 mm, mean ± SD) and precision (0.01 mm), at a temporal resolution (20 ms) suitable for real-time applications.

Methods: Experiments were performed on a 1.5T Philips Achieva (Philips Medical Systems, Best, The Netherlands). Each marker (Fig. 1a) is a glass sphere (3.25 mm internal diameter) filled with 10 mM Gd-doped water solution. The spherical symmetry minimizes orientation-dependence of the fiducial, and its short T1 allows full signal recovery in the presence of rapid RF-pulse excitations. A miniature RF receive-coil surrounds the sphere and is actively detuned via PIN-diodes during RF-transmit. The resonance circuit is attached via a transmission line to a custom interface box (Fig. 1b), which is connected to a vendor-provided Synergy Multi-Connect box for third-party coil designs.

Fig. 1a

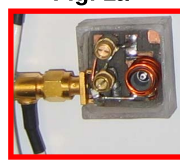
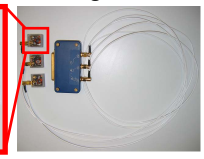


Fig. 1b



Markers are localized (2) by the tracking pulse-sequence in Fig. 2. A non-selective RF-pulse (θ = 3º) is followed by gradient-echo projection readouts in three orthogonal directions (total duration = 20 ms), thereby defining the position of each active marker uniquely in 3D-space. The markers are measured simultaneously, and since the receive channels are independent, signals from each can be separately and unambiguously identified. The signal from each marker's channel is displayed in real-time on the scanner console using an external monitoring GUI. Fig. 3 shows an example of the monitoring display of three markers after the first projection readout G_x. The time signal from each coil (Fig. 3, lower display) is dominated by the frequency f_0 corresponding to its location, x_0, given by the linear relationship: $x_0 = f_0 \bullet 2\pi/\gamma/G_x$. Fourier transform (Fig. 3, upper display) of the time signal yields its location along the projected axis. To refine its location, a quadratic fit to the intensities of the peak and two surrounding points is made. Since the SNR is typically >1000, this yields sub-millimeter resolution of each marker's location, well below the raw digital resolution. The following experiments used a sequence-defined tracking resolution of 1 mm, with the quadratic fit resulting in 0.01 mm resolution. Spoiler gradients (Fig. 2) designed to dephase the magnetization in large volumes while preserving signal from the smaller spherical samples (4) are used to further enhance the spherical sample's signal relative to any signal from the subject. If necessary, measurement errors caused by B0 inhomogeneities can be corrected by repeating the tracking pulse-sequence with the readout gradients reversed, and averaging the result (2,4).

A standard planar grid phantom was used for validation tests. The grid was secured to the scanner bed at a coronal orientation, and aligned with the y and z gradient axes. A marker was moved to 15 evenly-spaced positions, over a range of -112 to 112 mm, along both y and z axes. The B0-corrected tracking sequence was run five times at each position, resulting in a set of measurements to be used for evaluation of accuracy and reproducibility along the two gradient axes. Due to similarity of the physical x and y gradients, behavior in the x direction was not investigated.

Fig. 2

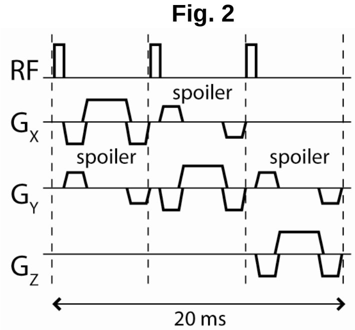


Fig. 3

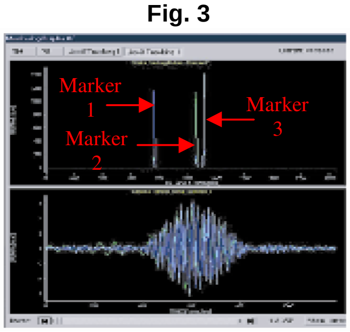


Fig. 4

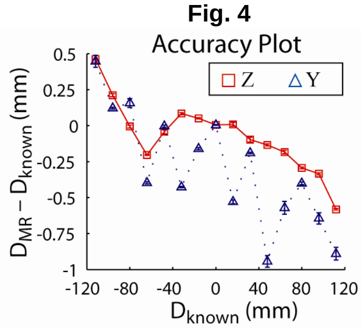
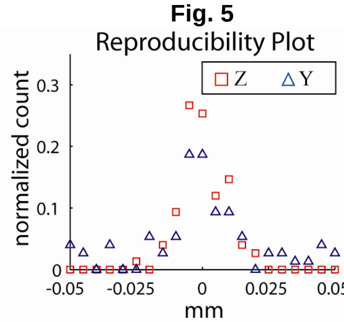


Fig. 5



Results: To measure absolute marker accuracy, the distance from each MR measurement to a reference point at the center of the grid was calculated. The resulting set of mean MR-measured distances at each position (D_{MR}) were compared with corresponding distances measured by digital caliper (D_{known}) at the same marker positions. Figure 4 plots $D_{MR} - D_{known}$ vs. D_{known}, (error bars are SD); overall localization errors (mm, mean ± SD) were 0.30 ± 0.40 and 0.07 ± 0.24 along the y and z axis, respectively. To evaluate marker reproducibility, deviation of each MR-measurement from its corresponding mean value D_{MR} was calculated. Figure 5 plots these distributions as normalized histograms; FWHM < 0.01 mm for both y and z gradient directions, illustrating high measurement precision consistent with our estimate of the quadratic fit resolution based on peak SNR.

Conclusion & Future Direction: A rapid position tracking method has been demonstrated that is suitable for high-precision, real-time applications, including medical instrument tracking and motion detection/correction. As a gating mechanism, a simple active marker may prove a competitive alternative to linear navigators. For applications related to head motion, 3D navigator-based tracking (11,12) procedures use the subject's head itself as the reference object. Because the head is large and complex, a much more complicated gradient scheme (i.e. trajectory through k-space) is needed to determine its orientation and position. In contrast, since each active marker is a small spherical sample with a designated receive coil, simple balanced readout gradients are all that is needed to determine their absolute positions. Three active markers attached to the head are sufficient to uniquely solve for the six degrees of freedom (three rotations and three translations about an orthogonal coordinate system) that fully describe any arbitrary rigid-body transformation; this transform can then be applied to update the image-plane orientation, thus compensating for subject movement (8-10). With modern scanners, multiple coil detection – for simultaneous device tracking and imaging – is straightforward, making the RF tracking-coil scheme a very competitive tool for motion detection.

References: [1] Ackerman J et al., Proc. 5th ISMRM 1986;1131-32. [2] Dumoulin CL et al., MRM 1993;29(3):411-15. [3] Zhang Q et al., MRM 2000;44(1):56-65. [4] Krueger S et al., IEEE Trans Med Imaging 2007;26(3):385-92. [5] Korin HW et al., MRM 1995;33(5):663-69. [6] Bernstein MA et al.,MRM 2003;50(4):802-12. [7] Shu Y et al., MRI 2006;24(6):739-49. [8] Derbyshire JA et al., JMRI 1998;8(4):924-32. [9] Krueger S et al., Proc. 14th ISMRM 2006; 3196. [10] Ooi M et al., Proc. 16th ISMRM 2008; 209. [11] Welch EB et al., MRM 2002;47(1):32-41. [12] van der Kouwe AJW et al., MRM 2006;56(5):1019-32.