Automatic device tracking in a closed-bore MRI: principle and initial experimental results on a robotically driven needle

G. Thörmer[1], N. Garnov[1], J. Haase[2], T. Kahn[1], M. Moche[1], and H. Busse[1]
[1]Diagnostic and Interventional Radiology, Leipzig University Hospital, Leipzig, Germany, [2]Physics and Geosciences Department, Leipzig University, Leipzig, Germany

Introduction/Purpose

An image-based localization of interventional tools or devices inside the magnet requires adequate, MR-visible markers and a powerful image processing tool to automatically detect them. In contrast to active tracking techniques, a substantial amount of time is needed for the acquisition of the marker images alone. In this work, a marker sequence at low spatial resolution was used to reduce the acquisition and hence the localization time. Inductively coupled radiofrequency (ICRF) miniature coils were used as markers to achieve efficient background suppression at very low flip angles. The parameters of an existing morphologic image processing tool [1] were adapted to allow 3D marker localization on poorly resolved MR images. The goal was to evaluate the performance of such an approach in a needle tracking experiment involving a robotic manipulator inside the magnet and to demonstrate the capabilities for potential applications in device tracking.

Materials and Methods

Three ICRF coils tuned to the resonance frequency (≈63.8MHz) of a standard 1.5T MRI (Siemens Symphony) served as MR visible markers (coupling to the integrated body coil) [2]. A custom-made board with three of these markers and a port for the instrument (here a 16G coaxial needle from Invivo Germany, Schwerin) was attached in a known geometry to the application module of a robotic manipulator (Innomotion, Innomedic, Herxheim, Germany) [3]. A dedicated software tool provided by the manufacturer was used to reproducibly move the robotic arm between start and finish positions with the needle immersed in a water phantom (Fig. 2). During this approach, the board markers were continuously scanned with a fast balanced SSFP marker sequence (TR/TE=2.1/0.9 ms flip angle FA=0.3°, acquisition matrix AMX=64x64, FOV=300 x 300 mm^2, slice thickness ST=300 mm, partial Fourier factor PF=6/8, receiver bandwidth RBW=1395 Hz/pixel). Automatic 3D localization was based on a 2D Gaussian template fitting and a subsequent matching of the fitted peak positions from three standard projections (sagittal, coronal, and transverse) [1]. The measured marker positions were used to calculate the geometries of the scan planes that contain the needle axis at each localization time point. A b-SSFP sequence with a "normal" flip angle (FA=70°, TR/TE=4.3/2.14 ms, FOV=150 x 150 mm^2, ST=3 mm, FA=70°, AMX=128x128, PF=4/8, RBW=1085 Hz/pixel) was then used to capture needle and surrounding "anatomical" structures during the repeated robotic motion of the needle along the same trajectory. The anatomical images were acquired at every fourth localization time point (n=20) because the standard sequence took an extra 900 ms for preparation and local shimming (actual imaging time was 300 ms only).

Fig. 1: a) MR-visible (ICRF) marker on a glass tube. **b)** Robotic manipulator (Innomotion) mounted on patient table. **c)** Detail of Fig. 1b (frame) showing custom-made board with three MR-visible markers and model instrument (16G coax needle).

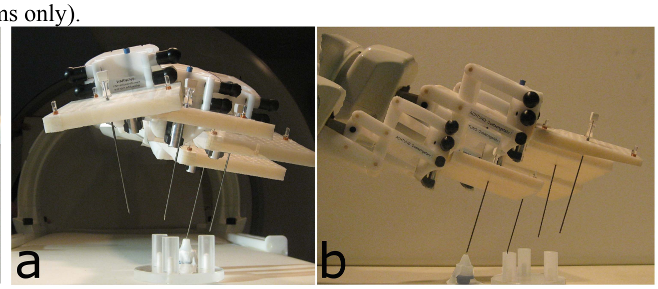

Fig. 2: Head-on **(a)** and lateral **(b)** view of the robotic arm with the attached needle during "target approach". The water phantom has been removed to clearly depict the entire needle length.

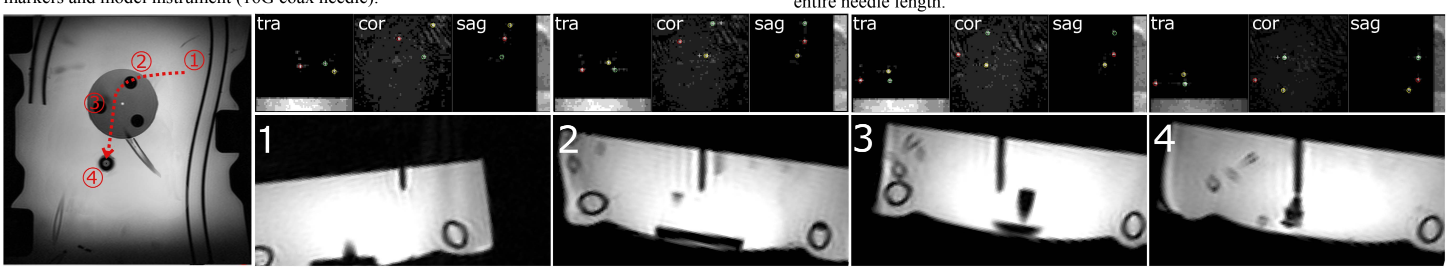

Fig. 3: Left image: Coronal roadmap image with overlaid needle trajectory (dashed line) realized by robotic manipulator. **Top row (right):** Marker positions (colored circles) are automatically overlaid onto each 2D view of the marker images. Note the faint anatomical background at a FA of 0.3°. **Bottom row (right):** The needle artifact (average FWHM diameter of 6 mm) is clearly depicted in the "anatomical" control images (detail) acquired with a slice thickness of 3.0 mm. Numbers correspond to device positions on Figs. 2 and 3.

Results

Figure 3 shows a selection of the resulting images during a 24 s robotic needle motion. Localization images were acquired every ≈300 ms. Three-dimensional marker detection was successful in 75 of 80 cases (≈94%). Accurate automatic definition of the scan plane geometry is demonstrated by the clear depiction of the needle artifact on all (20/20) "anatomical" control images.

Discussion and Conclusion

These results demonstrate that image-based, near-realtime tracking of interventional devices is possible and does not necessarily require highly resolved marker images. With pixel sizes of ≈4.7 mm and partial k-space sampling a rate of ≈3 position updates per second could be achieved. Faster position updates may be obtained by applying other acceleration techniques such as parallel imaging. For an inline device tracking, the suggested marker localization (marker imaging and analysis) needs to be integrated into a custom-made pulse sequence and interleaved with the anatomical imaging part. In conclusion, the presented approach is considered a flexible alternative for the localization of MR markers with potential applications in marker and device tracking.

References [1] H. Busse et al., JMRI 2007;26:1087, [2] M. Burl et al., MRM 1996;36:491, [3] A. Melzer et al., IEEE Eng Med Biol Mag 2008;27:66.