Navigation system for interventional MR image guidance in a closed-bore scanner: System setup and estimation of targeting accuracy

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Introduction/Purpose

The ongoing development of powerful magnetic resonance imaging techniques also allows for advanced possibilities to guide and control minimally invasive interventions. Various navigation concepts have already been described for practically all regions of the body [1]. Most diagnostic scanners, however, do not allow the physician to guide the instrument inside the magnet and, consequently, the patient needs to be moved out of the bore. The purpose of this work was to present a concept for real-time navigation with automatic patient registration and interventional control for a closed-bore scanner and to provide first estimates on the overall targeting accuracy in an experimental setup.

Materials and Methods

The navigation system (Localite GmbH, St. Augustin, Germany) is operated from a standard PC (3.2 GHz Pentium 4) outside the MR room and features DICOM image transfer, automatic patient registration, trajectory planning, and real-time navigation in standard or instrument-related views reformatted from intraoperatively acquired 3D data. Instrument tracking outside the bore is accomplished via a recently available optical tracking system (Polaris Spectra, NDI, Waterloo, Ontario) mounted on a rollaway stand. Automatic patient registration is based on a set of three optical and three MR-visible markers [2] mounted in a known geometry onto an MR-compatible arm (Invivo Germany GmbH, Schwerin) attached to the scanner table. The marker errors between the MR and the geometrically measured positions yield a crude estimate for the registration accuracy.

The navigation scene is then fed to an LCD projector outside the MR room and displayed on a large (35"×27") MR-compatible rollaway screen (Localite, Fig. 1). Needle guidance is accomplished by a sterilizable and detachable holder with a ball at its end (Invivo, Fig. 2). A VIBE sequence with 96 1-mm slices and a pixel spacing of 0.334 mm (matrix=512×256, TR/TE=4.3/1.81 ms, FA=15°) was used for planning and control imaging. Ten peas inside a gelatine phantom (Fig. 2) were targeted in a 1.5-T scanner (Siemens Symphony). The needle position was estimated from the respective MR artifact (Fig. 4), verified in situ, and compared with the planned positions. A one sample T-test of the differences between planned and measured needle positions was performed for each individual coordinate assuming +x=left (L), +y=posterior (P) and +z=superior (S).

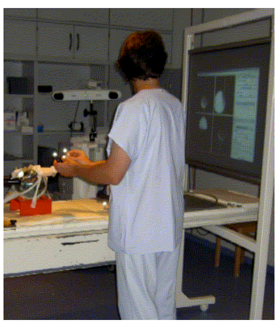

Fig. 1: Overview of the in-room components (see also Fig. 2): instrument tracker and holder, marker holder, tracking camera, and rollaway screen.

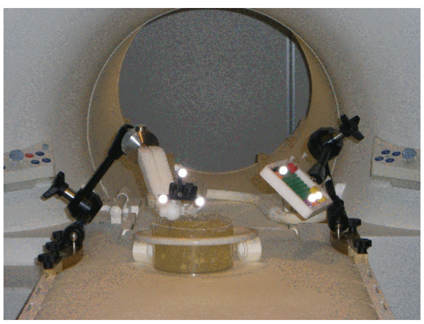

Fig. 2: Close-up of instrument and marker holders. The instrument is attached to a device with 3 optical markers (MR tracker), held and guided by a ball which may be tightened or completely detached by opening the two legs that hold the ball.

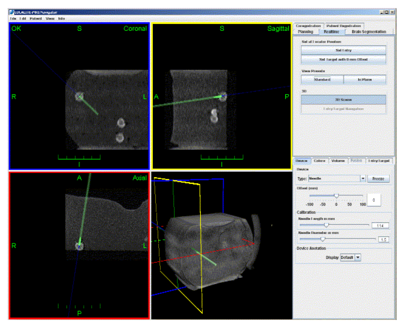

Fig. 3: Screenshot of the Localite PC during real-time navigation relying on three standard views and a 3D rendering of the data to improve mental mapping. Overlaid needle object indicates needle orientation and virtual extension.

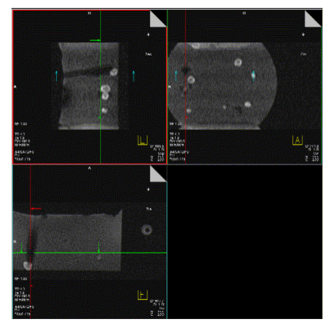

Fig. 4: Estimation of the 3D needle position by analyzing the needle artifact on reformatted sagittal and coronal views of the original transverse data set.

Results

In comparison with a previous camera model (Polaris), the much shorter warm-up time (~10 vs. ~30 min) and the extended measurement volume (up to 3 m) of the model used here (Spectra) were considered to be beneficial for the interventional workflow. The large display of the navigation scene right next to the physician allowed for an excellent hand-eye coordination. Previous efforts with the small display screen of an existing in-room console or with an LCD screen behind the MR window were considered to be inappropriate solutions. The three marker errors during registration were 0.6, 0.7, and 1.1 mm. There was no significant bias in the realized needle position for the L (p=0.704) and P (p=0.285) coordinates and a highly significant bias of ~3.0 (±0.32) mm in the S coordinate (p<.001). This was also confirmed by visual inspection. All realized needle positions were systematically shifted 2.5-3.5 mm towards the S direction. The resulting mean 3D error is 3.17±0.45 mm (95% CL=4.1 mm).

Discussion and Conclusion

Potential error sources for the targeting accuracy of our setup include but are not limited to: (1) Estimation of needle position from artifact on MR image. Reported to be in the mm-range [3]; (2) Gradient nonlinearity of the scanner. Similar to 1, depending on distance from isocenter, here: ~120 mm; (3) Registration error. Mean marker error was 0.8 mm; (4) Tracking accuracy. Given as 0.30 mm RMS and 0.60 mm at 95% confidence level; (4) Motion of the marker holder. Marker positions before and after experiment agreed within ~0.4 mm; (5) Needle bending. Is considered small for gelatine. Although we can not exclude other experimental factors we believe that the most likely sources are (1) and (2). Further investigations are necessary to determine the origin of that bias and a more reliable estimate of the targeting accuracy. For phantom experiments, we believe that a 95% CL of 3 mm will be sufficiently accurate for interventional purposes. Current efforts for an immediate and automatic control scan of the actual needle position relies on an existing socket interface (Siemens) which can directly interact with the graphical slice positioning of an open pulse sequence.

In conclusion, we believe that our approach is a promising alternative for selected MR-guided interventions in many regions of the body. The system provides high-quality images in real time, adequate hand-eye coordination, and seems to be compatible with the clinical workflow.

References [1] Moche M et al., "Navigation Concepts ...", JMRI 2008; in press, [2] Busse H et al., JMRI 2007;26:1087, [3] Blumenfeld P et al. JMRI 2007;26:688.