Real-time active-tracking of metallic needles during MR-guided radiation therapy: from concept to the first human trial

Wei Wang[1,2], Zion T. H. Tse[3], Wolfgang Loew[4], Charles L. Dumoulin[4], Isaiah Norton[5], Ravi T. Seethamraju[6], Antonio L. Damato[2], Tina Kapur[1], Akila N. Viswanathan[2], Robert A. Cormack[2], and Ehud J. Schmidt[1]

[1]Radiology, Brigham & Women's Hospital, Harvard Medical School, Boston, MA, United States, [2]Radiation Oncology, Brigham & Women's Hospital, Harvard Medical School, Boston, MA, United States, [3]Engineering, University of Georgia, GA, United States, [4]Radiology, Cincinnati Children's Hospital, OH, United States, [5]Neurosurgery, Brigham & Women's Hospital, Harvard Medical School, Boston, MA, United States, [6]Siemens Healthcare, MA, United States

Target Audience: Scientists and clinicians interested in actively-tracked MR-guided interventions

Purpose: MRI is increasingly used for radiation treatment, due to improved visualization of the tumor and its surroundings. In MRI-guided interstitial radiation therapy (brachytherapy), treatment outcomes may improve via placement of catheters (the holder for radioactive sources) into selected regions around the tumor and precise identification of the catheter trajectories after placement. The catheters, which are filled with MR-compatible metallic needles during placement, can be passively tracked, but this process is time-consuming and relatively inaccurate[1]. Active tracking using microcoils is also challenging because 1) the metallic needles within the catheter lead to static (B_0) and RF (B_1) magnetic field inhomogeneities which are exacerbated by the close proximity of 10-20 needles; 2) RF currents induced on metal surfaces can distort imaging and cause heating. The purpose of this study was to develop an active tracking device and dedicated software which enables rapid and accurate real-time tracking of the metallic needle, leading to an improvement in the accuracy and speed of MRI-guided clinical interventions.

Method: Three micro-coils were built using flexible printed circuits and mounted on the surface of a brachytherapy needle (Fig. 1). The microcoils were connected to an 8-channel MR-tracking receiver, allowing simultaneous tracking of up to eight coils. The coil design was optimized by modeling the receive sensitivity (B_1^-) of different coil configurations placed on metal. 3D microcoil positions were measured by an MR-tracking sequence with zero-phase-reference and Hadamard multiplexing schemes[2] implemented on a Siemens 3T scanner (resolution: $0.6 \times 0.6 \times 0.6$ mm^3; 40 updates/sec). Phase-field dithering (PFD) was integrated to suppress the effects of B_1 and background inhomogeneities[3]. The tracked coil positions were continuously transferred to an external workstation for real-time visualization. The system was tested in a phantom and then used in a clinical procedure.

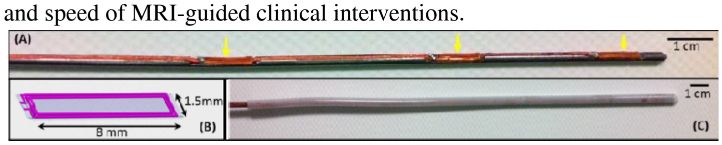

Fig 1: (A) Grooves were carved into the surface of the needle for mounting tracking coils (yellow arrows); (B) Each coil was built on a double-layered flexible printed circuit sheet, consisting of four rectangular conductive loops; (C) The modified needle fits into the original hollow plastic brachytherapy catheter.

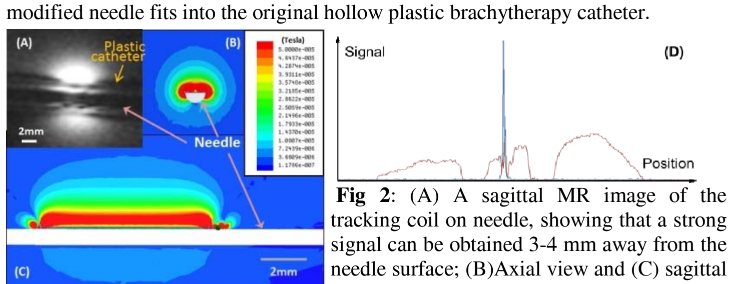

Fig 2: (A) A sagittal MR image of the tracking coil on needle, showing that a strong signal can be obtained 3-4 mm away from the needle surface; (B)Axial view and (C) sagittal view of the B_1^- field simulation carried out with the coil in a saline solution at 123.183 MHz using finite element method; (D) Comparison of tracking signals before (red) and after (blue) PFD was applied.

Results: The B_1^- field of the optimized micro-coil design was perpendicular to the needle surface, with its profile extending beyond the area where the susceptibility-induced B_0 gradient is greater than the frequency-encoding gradient in the tracking sequence (~2 mm from the surface[4]) (Fig. 2). This field profile is still adequately spatially localized ($3\times3\times8$mm^3), which is essential for tracking precision. PFD provides a sharp signal peak by eliminating the broad signal arising from coupling to neighboring needles (Fig. 3). The SNR of the tracking was ~20-178 in the phantom and ~ 18- 80 in the human.

Eight catheters were tracked inside a prostate phantom. Each coil position was recorded during needle pull-out for the reconstruction of catheter trajectories. The trajectories were compared with the signal voids in the 3D fast spin echo (SPACE) images (minimal susceptibility artifacts) (Fig. 3). Distances from the tracked point to the center of the signal void on the axial images were all <1 mm. The heating by a 4 Watt/kg sequence was <1^{0}C.

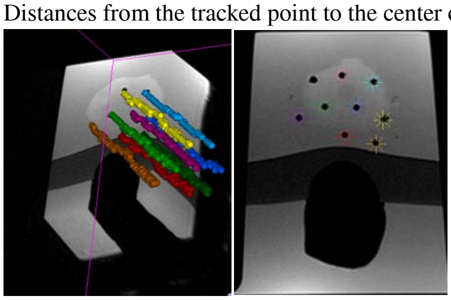

Fig 3: Left: eight needle trajectories in a prostate phantom; Right: axial slice shows consistency of tracking data (starburst markers) with signal voids in high-resolution 3D FSE image.

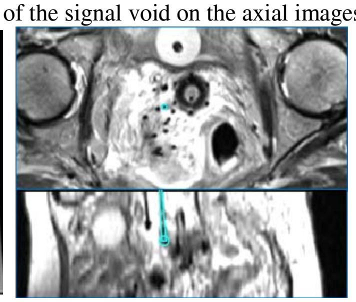

Fig 4:Needle tip position and orientation shown as a vector (cyan) on a pre-acquired 3D MRI data set (upper: axial view, lower: sagittal view)

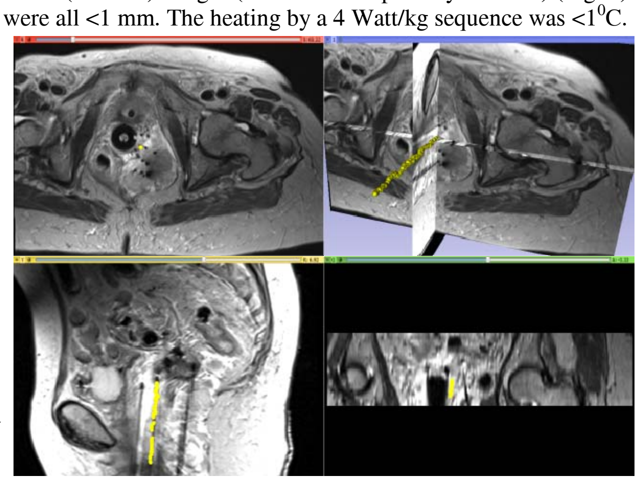

Fig 5: 3D rendering and three orthogonal views of one needle trajectory (in yellow) overlaid on the 3D turbo spin echo images of the patient pelvis. The trajectory was reconstructed by tracking the positions of the distal coil during needle pull-out.

With IRB approval, active MR-tracking was conducted for catheter placement in a woman with recurrent uterine adenocarcinoma. One catheter's location was verified by tracking needle both during the insertion and removal. A second catheter was actively tracked during pull-out. The calculated catheter tip position and orientation was displayed and updated in real time on pre-acquired 3D images (Fig. 4). After the placement of the catheters, the full needle trajectories were reconstructed by continuously tracking the position of the distal micro-coil during pull-out (Fig. 5).

Conclusion: For the first time, an actively-tracked metallic needle with dedicated MR-tracking software was successfully applied in a clinical case. This facilitates accurate and time-efficient catheter insertion and enables improved targeting in radiation therapy. This approach can be generalized to other interventions requiring metallic devices (e.g., guide-wires, cannulas, trocars).

References: 1.Seevinck et al. Magn Reson Med 2011; 65: 146-156. 2. Dumoulin et al. Magn Reson Med 1993; 29:411-15. 3. Dumoulin et al. Magn Reson Med 2010; 63:1398-1403. 4. Müller-Bierl et al. Medical Physics 2004; 31: 579.