Real-time automatic tracking with a dedicated 3D Hall-effect integrated circuit for MRI-guided interventions

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TARGET AUDIENCE – Interventional radiology community

PURPOSE – The alignment of the MR scan planes to the interventional instrument (e.g. biopsy/ablation needle, catheter) in order to monitor the procedure is a demanding and time-consuming task. Gradient-based tracking techniques, that do not require NMR signal, have already been proposed based on small orthogonal inductive coils[1] to measure dedicated gradients along the X, Y and Z MRI-axis. Position and orientation of the sensor are then computed based on gradient maps. The use of a 1D Hall sensor was proposed as an alternative to inductive coil[2]. However, a 3D real-time tracking system was not developed probably due to the unavailability at that time of an accurate and monolithic 3D Hall sensor. In this work, a dedicated 3D Hall chip designed in standard CMOS technology and its localization algorithm are evaluated for real-time tracking inside the MRI.

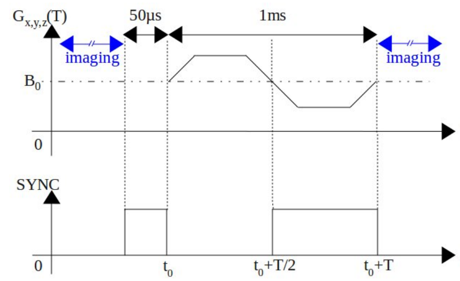

Fig. 1. Dedicated bipolar gradient and sync signal added to the sequence for the device localization.

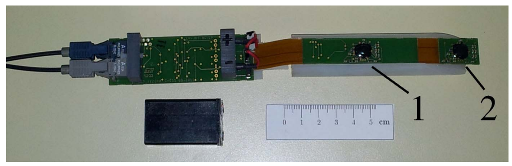

Fig. 2. Tracking device with two 3D Hall probe**s.**

METHODS – 3D Hall probe and tracking device: A 3D Hall sensor and its specific amplifier circuit have been designed and integrated on a CMOS chip[3]. The horizontal Hall device, sensitive to the magnetic field component perpendicular to the chip plane, has a 20 µT resolution while the 2 vertical Hall devices, that each sense one of the magnetic field components in the chip plane, have a 50 µT resolution each. A synchronization (sync) signal is played by the MR system to trigger the 3D measurement of a dedicated 1ms bipolar gradient (fig. 1). Two of this 3D Hall probes have been embedded in a tracking device (fig 2) including a 9V non-magnetic battery and 2 optical fiber connectors: one for the sync trigger from the MRI hardware and a second one to transmit the digital gradient measurements to the processing computer outside the MRI room.

Pulse sequence: A spoiled gradient echo sequence designed for interactive multi-slice real-time monitoring[4] is modified to add the localization bipolar gradient. The localization gradient is played along one direction, each other TR (between the readout and the next RF excitation pulse), successively alternating between the X, Y and Z-axis of the MR system, with no localization gradient played every other TR for main magnetic field (B0) suppression. With this scheme, 6 TR (30.6 ms, see below) are thus needed for a 3D localization of the chip. Relevant imaging parameters are: FoV 350 mm, matrix 192×154, TE/TR 1.6/5.1 ms, FA 15°, ST 8 mm, image acquisition time 780 ms.

Real-time localization and scan planes alignment: Experiments are carried out in a 1.5T scanner (MAGNETOM Aera, Siemens). An external computer is set up to receive the gradient measurement and compute the position and orientation of the tracking device. The localization algorithm requires a comparison between the measurement and a 3D map of the magnetic field from each gradient coil. In the present study, a linear model of the gradient coil fields was identified during a calibration procedure of a 10cm^3 3D grid with a 16 mm step. The algorithm is implemented with the Matlab™ software for prototyping and requires around 240 ms to compute the 3D device localization. Based on the calculated pose, the parameters of the next 2 orthogonal real-time MR slices to acquire -centered and aligned on the main axis of the tracking device- are sent to the MR scanner console through an Ethernet protocol.

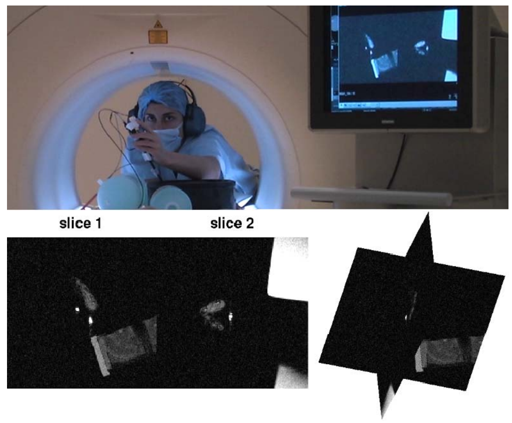

Fig. 3. Real-time tracking of the Hall probe with 2 orthogonal MR slices aligned to its axis. Fiducial markers (bright dots) placed on the device remain visible during free-hand motion.

RESULTS – The static accuracy of the tracking device was evaluated at 32 known positions in the calibrated workspace with an averaging of the gradient measurements over one image acquisition (42 measurements per MRI axis). Mean position and rotation errors are respectively 1.06mm ± 0.52 and 2.92° ± 3.42 for a device orientation similar to the one during the calibration. However, they become respectively 2.90mm ± 1.57 and 6.12° ± 6.62 when the horizontal Hall sensor is orthogonal to its orientation during the calibration. Finally, the effectiveness of the tracking during free-hand motion was qualitatively evaluated by adding fiducial markers on the 3D Hall probe and checking that they stay visible during moderate speed motion of the device in the real-time MRI images (fig 3).

DISCUSSION – First experimental results show a 3D localization accurate on the order of 1mm, and validate the real-time tracking setup. Increase of the error when the orientation of the horizontal Hall sensor is different from the gradient mapping procedure, may result from a minute deviation of the magnetic field lines due to small ferromagnetic components on the tracking device board (e.g. resistors and capacitors) that are not yet integrated inside the chip.

CONCLUSION – These results validate the concept of a 3D Hall sensor on chip for real-time tracking and image-plane automatic alignment in MR-guided procedures. Increased accuracy is expected from the full integration of the circuit on the chip. The small form factor of the chip (2.3×3.4mm) opens the way to its integration inside most surgical instruments.

REFERENCES – 1. E. Nevo et al., An electromagnetic 3D Locator System For Use in MR Scanners, ISMRM, 2002; 334. **2.** K. Scheffler, J. G. Korvink, Navigation with Hall sensor device for interventional MRI, ISMRM, 2004; 950. **3.** J.-B. Schell et al., Towards a Hall effect magnetic tracking device for MRI, EMBC 2013; 2964-2967. **4.** Pan L et al., An Integrated System for Catheter Tracking and Visualization in MR-Guided Cardiovascular Interventions. ISMRM 2011.