An Active Microelectrode System for Experimental MRI-Guided Intracranial Intervention

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Introduction: Deep brain stimulation (DBS) of the subthalamic nucleus (STN) is an emerging treatment option for patients suffering from severe Parkinson's disease. In DBS, an electrical pulse is delivered to implanted leads and electrodes in direct contact with cerebral tissue in the STN. The leads are subcutaneously connected to a pulse generator implanted in the thoracic cavity. The electrodes are implanted in the STN via stereotactic surgical procedure, wherein a microelectrode system is advanced to the STN, guided only by repeat electrical (EEG) measurements and pre-operative MRI and/or X-ray CT images. Nevertheless, these procedures are very time-consuming (typically 4-8 hrs) and have a suboptimal (~75%) success rate. If they could be performed successfully under MRI guidance, identifying the STN from its MRI characteristics so that the DBS leads could be directly implanted in the STN, then both the procedure time and the success rate could potentially be greatly improved [1, 2]. Thus, the value and efficacy of DBS treatment could be dramatically improved. To evaluate the suitability of MRI for directly guiding DBS lead placement in the STN, we have developed and tested an MRI-compatible cannula and microelectrode system that incorporates an MRI-active loopless antenna. The signal-to-noise ratio (SNR) performance, safety, and preliminary *in vivo* results from a 3T MRI system in a primate are reported.

Methods: We fabricated a cannula coil from three insulated concentric Nitinol tubes, arranged to form a loopless MRI antenna[3]. The innermost tube forms the core of the antenna and provides a conduit to advance an additional MRI-antenna/micro-electrode. The intermediate tube forms the shield of the antenna in combination with the inner tubing, and forms an RF choke with the outer tube via an electrical connection at the proximal end. The inner and outer surfaces are insulated with polymer. The micro-electrode loopless antenna was also fabricated from nitinol tube as the shield, with an insulated gold-plated wire as the core of the antenna. The dipole tip of the antenna is left uninsulated to permit EEG measurements. The cannula and microelectrode antennae are matched and tuned to 123.2 MHz, and decoupled from pick-up during MRI excitation via a decoupling switch analogous to that described previously [3].

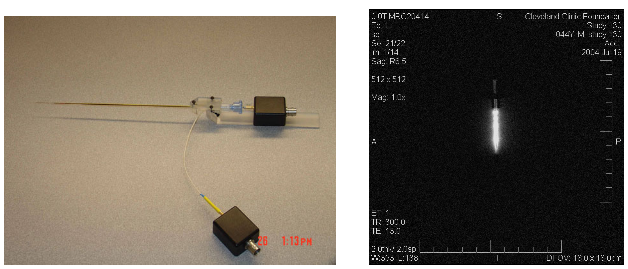

Fig 1: MRI-active cannula, coil (left) and phantom image.

MRI testing of the cannula and coil were performed on a Siemens Allegra® 2.9T scanner. Safety testing was performed with a polyacrylamide gel phantom with a conductivity of 0.9 S/m. A nominal 4W/kg SAR (head) MRI sequence was applied for 3.4 min. Local temperature was measured directly using *FISO* fiber-optic temperature probes. Actual SAR was calculated from the rate of initial temperature rise and the known specific heat of the gel. The MRI SNR of the antenna system was measured in a saline phantom at various insertion depths to simulate clinical conditions.

MRI-guided advancement of the cannula/microelectrode coil system in the brain was tested *in vivo* in a Macque monkey with protocols approved by the Animal Care and Use Oversight Committee. Preoperative MRI and CT images were obtained to design and locate a suitable head mount and to determine the trajectory for micro-electrode advancement. The animal was anesthetized, a burr hole prepared, a head mount assembly fixed to the skull, and the cannula/antenna system advanced under MRI guidance to the location of the STN. To optimize SNR and field-of-view, images from the scanner's external head coil and the internal micro-electrode coil were combined using a custom phased array adapter.

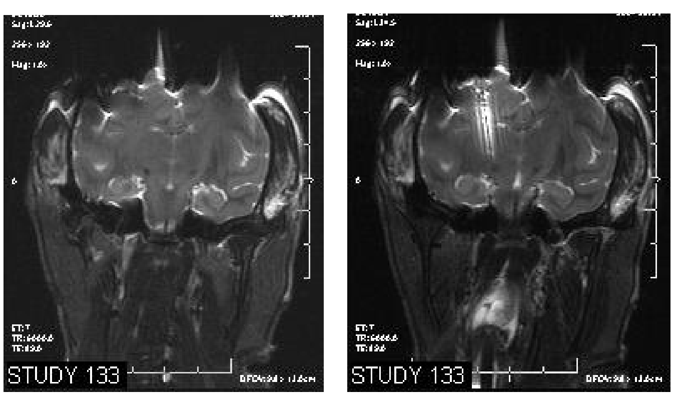

Fig 2: Scout MRI (left), and MRI with probe advanced towards STN

Results and Discussion: The complete cannula and micro-electrode coil is pictured in Fig 1. The maximum temperature increase and SAR was 0.2°C and 4W/kg, respectively, at the microelectrode junction. Thus, the device *can* be effectively decoupled at 3T. The SNR profile of the antenna demonstrated a greater than 50% advantage over the head coil in a circular region of 4 mm radius around the coil. In the primate study, the signal from the scanner's head coil depicts the overall cranial anatomy, while the locally enhanced SNR provided by the MRI-active probe system as it advances into the STN is shown in Fig 2.

We have demonstrated an MRI-compatible active cannula and micro-electrode system suitable for intra-cranial interventionsthat is both safe for MRI and provides a local SNR advantage. The SNR gain should translate to enhanced local spatial resolution. We demonstrated successful use of the device to target the STN in a primate model on a 3T clinical MRI system.

References: 1. Hariz M.I., et al. Stereotact Funct Neurosurg, 2003. **80**: p. 96-101.
2. Rezai, A.R., et al. Invest Radiol, 2004. **39**(5): p. 300-3. 3. Ocali O, Atalar E. Magn Reson Med, 1997. **37**(1): p. 112-8.