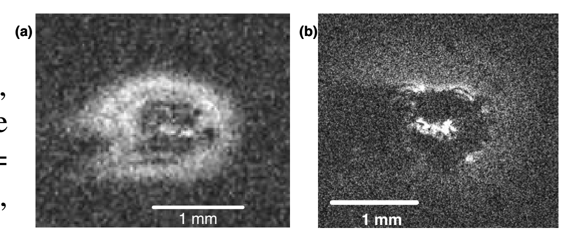
MRI Microcoil and Depth Electrode

D. S. Strick[1], M. S. Cohen[2], and J. W. Judy[3]
[1]Biomedical Engineering, University of California, Los Angeles, Santa Monica, CA, United States, [2]Psychiatry/Neuro/Radiol Sci, University of California, Los Angeles, Los Angeles, CA, United States, [3]Electrical Engineering, University of California, Los Angeles, Los Angeles, CA, United States

INTRODUCTION: Temporal lobe epilepsy (TLE), which is considered the most common form of epilepsy, remains poorly understood, and nearly a third of the patients affected do not respond to anti-convulsion medication [1,2]. Surgical treatment is highly effective when clinicians can correlate the location of structural and functional abnormalities in a patient [2]. Magnetic Resonance Imaging (MRI) is the favored neuroimaging method for identifying structural lesions in TLE [3], while intracranial electrodes reveal waveforms characteristic of seizure generating tissue. Unfortunately, existing intracranial electrodes create magnetic-susceptibility artifacts that prevent imaging of the tissue from which the electrodes record. We aim to design an intracranial electrode for the study, treatment and diagnosis of TLE that transforms the region normally obscured in MRI into the region of greatest spatial resolution. To achieve our goal we must design an implantable MRI antenna, and an artifact-free intracranial electrode.

METHODS: To provide ultra-high resolution MRI near the region of the electrode, we design an implantable radio frequency (RF) coil to integrate with an intracranial electrode. We evaluate several RF-coil geometries and determine a 1-mm-diameter solenoid, known in the field of nuclear magnetic resonance (NMR) as the microcoil, as the optimal coil geometry for this application. NMR microcoils typically exist as glass capillary tubes nested on a silicon chip. We modify the NMR-microcoil design to allow implantation of the RF coil, by winding the microcoil on medical-grade silicone tubing and attaching circuit components on the tubing or at the distal end. We coat copper wire with 25 microns of biocompatible polymer (Parylene C) in order to achieve proper turn spacing. Tuning and matching circuitry insures that the impedance of the RF coil is 50 ohm at the operating frequency for 3-Tesla proton MR applications. We verify the functionality of the RF coil design as an MR transceiver.

RESULTS: The implantable RF coil provided MRI of neural tissue (butcher-grade *Ovis aries*). The images shown here were obtained by using the micrcoil as a transceiver for a 3-T Scanner (Siemens AG, Berlin) Fig. (a) TR/TE = 3000/22 ms, FOV = 26 mm, 256 x 256, slice thickness = 0.40 mm, number of slices 30, NA = 10; Fig. (b) TR/TE = 123/48 ms, FOV = 22 mm, 1024x1024, slice thickness = 0.17 mm, number of slices 15, NA = 3.



DISCUSSION: Obtaining images with the implantable microcoil is a milestone in developing our integrated neural imaging and recording probe for TLE. The next generation of implantable microcoils is underway, including coil modifications and development of circuitry to allow use as a receive-only coil. Future work includes quantification of coil performance, development of an artifact-free electrode, and safety experiments to quantify MR-related heating.

REFERENCES:

[1] Sloviter RS. C. R. Biol. **2005**; 328(2):143-53
[2] Briellmann RS, Kalnins RM, Berkovic SF, Jackson GD. Neurology **2002**;58:265-71
[3] Engel J, New Eng. Journal of Medicine **1996**;334:647-652