Planar coils for Optimal Micromagnetic Brain Stimulation.

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Introduction.

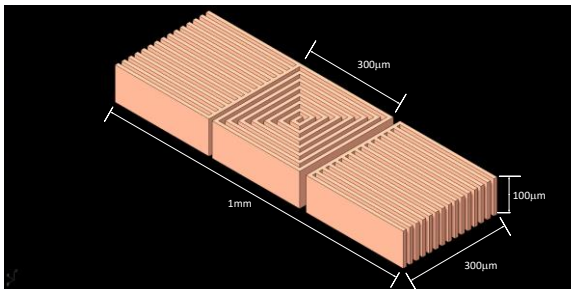

Figure 1: The proposed single layer planar square spiral coil of dimensions which is in copper based on a 10µm trace width. The side structures are simple pads that retain the 10µm trace width structure for uniformity.

Micromagnetic stimulation (µMS) near excitable tissue induces a localized current gradient in both time and space adequate to activate neurons, as demonstrated *in vitro* by activating retinal ganglion cells [1]. We have also shown that µMS is capable of activating neuronal circuitry in-vivo in rodent models, using acutely implanted micro coils to activate neurons of the inferior colliculus [2]. These experiments have shown that µMS has the promise of introducing new paradigms in the stimulation of the human nervous system. µMS has several advantages over the more traditional electrical stimulation: (1) it does not require charge-balanced stimulation waveforms, (2) it can activate neurons with specific axonal orientations, (3) it has improved biocompatibility when coated with implantable grade polymers, and (4) it is a potentially magnetic resonance imaging (MRI) compatible technology. The *in-vitro* studies [1] showed that µMS coils placed parallel to the surface of the tissue can activate neurons differentially based on coil orientation. Furthermore, previous electromagnetic simulations indicate that single-layer planar spiral geometries are more efficient in eliciting neural activation because they induce significantly higher electric fields (up to 5 folds) into the tissue compared to solenoidal or helical coils [3]. The quandary is that µMS technology, which was first developed in our laboratory [1, 2], was entirely based on commercial components off the shelf (COTS), which are readily available to researchers. However, COTS inductors are designed to maximize efficiency (Q-factor), which consists of trapping the generated magnetic field to minimize its losses. Unfortunately, this type of design shields off the tissue from most the magnetic flux and thus greatly reduces the stimulation efficacy. Instead, the magnetic flux of single-layer planar spiral coils may directly expose the tissue without trapping the magnetic flux. In this work, we studied and fabricated preliminary prototypes of a square design version of micro-scale single-layer planar spiral coil structures as µMS devices. Furthermore, we studied electric field **E** distribution over space with numerical electromagnetic simulations. The neural activation function is proportional to the derivative of the **E** along the axon's axis. The Neurons which oriented perpendicular to the long axis of the solenoid are more likely to be activated as they experience higher **E**.

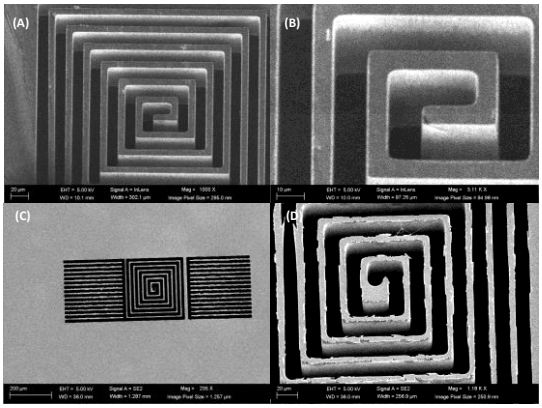

Figure 2: SEM images of the stencil wafer. Top view of the six turns single layer planar spiral coil (A), zoomed in view of the coil from the front (B), overview of the entire coil structure from the back side (C) after flipping the stencil wafer over showing that the DRIE reached through and through, and details of the back side of the stencil wafer that show a bottom footing (D).

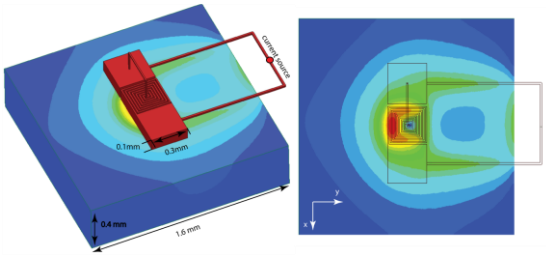

Figure 3: The Finite Elements Simulations. (Left) The geometric model consisting of the coil connected to a current source on top of a slab of tissue. (Right) The electric field induced in the tissue exhibits an asymmetry, which allows stimulating neurons along the direction of the x-axis.

Coil Manufacturing

Fig. 1 illustrates a cad drawing of the proposed single layer planar square spiral coil of dimensions 300µm×300µm×100µm which will be made from copper based on a 10µm trace width design. The two side structures around the spiral are simple pads that retain the 10µm trace width structure for uniformity, the lower pad will be connected to the center of the coil with a microwire.

In Fig. 2 is shown Scanning Electron Microscope (SEM) images (Hitachi SU-8230) of a 100µm thin 100mm Si wafer (undoped, orientation <100> and > 5 kΩ-cm resistance). The

SEM images were acquired after photolithography, and deep reactive ion etching (DRIE) performed at the Center for Nanoscale Systems at Harvard University. Photolithography was performed as follows: a 7μm thick layer of SPR 220-7 was spin coated on the thin wafer after hexamethyldisilazane (HMDS) deposition. This priming step was crucial for adhesion promotion, and the thin wafer was dehydrated through a series of heated (150°C) evacuation and dry nitrogen refills, followed by HMDS vapor inserted into the evacuated chamber forming a monolayer. The coated wafer was then exposed to a laser direct writing system (uPG501, Heidelberg Instruments Inc, Heidelberg, Germany) with a pattern shown in Fig. 1, and developed using AZ 400K for 4 minutes, without stirring. The wafer was then etched with a deep reactive ion etching system (DRIE-Rapier, SPTS Technologies, Newport, UK), which etched the thin wafer using Bosch switched processing for obtaining the steep vertical profiles (Fig. 2). The stencil wafer is still part of work in progress and will then be used to grow copper inside it using copper electroplating deposition.

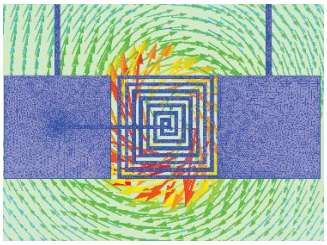

Figure 4: The induced eddy currents in the tissue generated by the μMS coil magnetic flux.

Simulation results

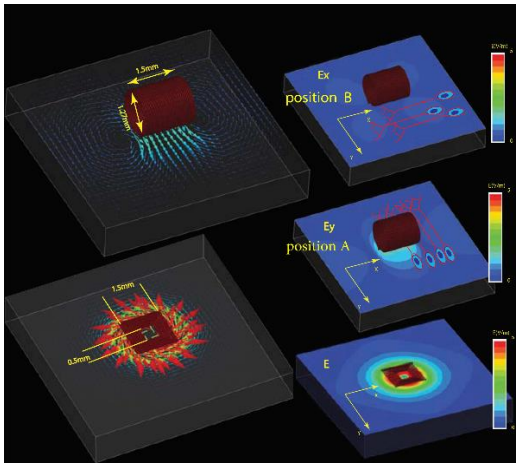

Figure 5: ANSYS Finite Elements Simulations. (Top) E-fields generated by a solenoidal μMS coil, (Bottom) E-fields generated by a planer μMS coil.

Fig. 3 shows the FEM results of the electric field induced in the tissue exhibits an asymmetry along the x-axis. This electromotive force induced by the magnetic flux of the coil (Fig. 4) which has a component along mainly one axis, could be exploited to activate axons with a specific orientation. The current induced by the coil follows a circular path (eddy current) in the tissue, such that current densities in the coil in the tissue have opposite. The E and B distributions below the coil are presented in Figs 5 and 6.

Since neural activation function is proportional to the derivative of the electric field along the axon's axis, neurons oriented perpendicular to the long axis of the solenoid are more likely to be activated as they experience both higher E (Fig. 5) and B fields' (Fig. 6). Spiral coils, on the other hand, induce a symmetric al E field in the tissue. However, our simulations indicate that such geometries are more efficient in eliciting neural activation because they induce significantly higher electric fields (up to 5 folds) into the tissue compared to solenoidal coils.

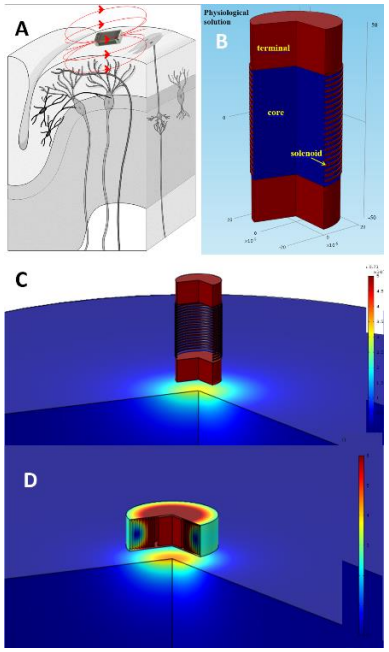

Figure 6: COMSOL Finite Elements Simulations. (A) an example of positioning of μMS coil on cortex for cortical stimulation, (B) Geometry of a realistic solenoidal μMS coil, (C) B-fields generated by a solenoidal μMS coil, (D) B-fields generated by a planer μMS coil.

Conclusions.

Numerical simulations provide a crucial insight into mechanism(s) of micro-magnetic nerve stimulation and should be used both during the design process, and to interpret the neural responses. Based on these results, the long solenoid design optimizes the Q-factor but render the bulk of the magnetic energy inaccessible. An alternative is the use of multiple slimmer coils that maximizes the access of neuronal tissue to the magnetic flux produced by μMS coil to optimize neural stimulation. An initial prototype of the stencil was built using DRIE suggesting that the structure is manufacturable.

References

1. Bonmassar, G., et al., *Microscopic magnetic stimulation of neural tissue.* Nat Commun, 2012. **3**: p. 921.
2. Park, H.J., et al., *Activation of the central nervous system induced by micro-magnetic stimulation.* Nat Commun, 2013. **4**: p. 2463.
3. Bonmassar, G. and L. Golestanirad. *EM fields comparison between planar vs. solenoidal μMS coil designs for nerve stimulation*. in *2017 39th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*. 2017.