Student Paper

New Model of the Portable Transcranial Magnetic Stimulation Apparatus

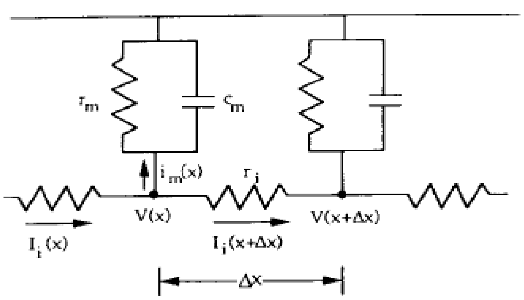
Hanmin Qian[a], Qinglei Meng[b], Fow-Sen Choa[b]

[a] *Centennial High School, USA, qianhanmin99@gmail.com* [b] *Department of Computer Science and Electrical Engineering, UMBC, USA*

There is a large body of evidence demonstrating altered brain circuit function and behavioral output with invasive technologies, such as deep brain stimulation and optogenetics. These invasive technologies in humans, non-human primates, or rodents have proven successful in altering neuronal plasticity and behavioral plasticity. The ability to modulate neuronal activity in precise brains regions, without surgery, can increase the accessibility of neuromodulation testing, without the adverse effects with brain penetration or other confounds that occur with optogenetic or deep brain stimulation technologies. Non-invasive brain modulation techniques, like transcranial magnetic stimulation (TMS), can not only provide a detailed assessment of the overall effects of non-invasive stimulation on neural plasticity, behavioral plasticity, and whole brain activity but also bring in new treatment of many neural diseases like essential tumors, Parkinson's disease, dystonia, depression, migraine, Alzheimer's, and Schizophrenia, ...etc.

The purpose of this investigation is to study an optimal structure for TMS stimulation within 2cm distance away from the stimulator. This distance is ideal for doing TMS research in mice, but can also applied to humans for stimulation in the cortex region, which is responsible for a number of important cognitive and physiological functions. The purpose of this investigation is to find an optimal structure for TMS for stimulating a region at 2cm away. This distance is ideal for doing TMS research in mice, but can also applied to humans for stimulation in the cortex region, which is responsible for a number of important cognitive and physiological functions.

In this research, a two-dimensional finite element analysis software 'FEMM' is used to model the structure of the coils and plot graphs related to specific coil arrangements. FEMM presents the cross section of the coil when it is cut along an axis. The magnetic flux density distribution graph is then compared between the current commercial model and the new design.



$$\lambda^2 \frac{\partial^2 V}{\partial x^2} - V = \tau \frac{\partial V}{\partial t} + \lambda^2 \frac{\partial \epsilon_x}{\partial x}$$

Figure 1. the neuron membrane equivalent circuit and the cable equation. Where V is the membrane potential, λ is the length constant of the axon, τ is the time constant of the axon, and x is directed along the axon.

As shown in figure1 the neuron membrane equivalent circuit and the neuron stimulation cable equation [1], we can find that the driving force comes from the $\partial\varepsilon_x/\partial x$ term and not from the magnetic field or the induced electrical field. So, to maximize the stimulation effect, we shall design stimulator coils that can produce maximum *gradient electrical field*. With the Maxwell equation: $\nabla \times E = -\partial B/\partial t$, assume the B field is along the z direction, then we can simplify the equation to $\partial Ey/\partial x = -\partial Bz/\partial t$ or $\partial Ex/\partial y = -\partial Bz/\partial t$. Therefore, for a symmetric case like a ring-shape coil, the curl electric field equals to $-\partial Bz/\partial t$ and the electrical field gradient (*dE/dx*) distribution can also be proportional to the gradient of the transient magnetic field. In the following we will try to optimize coil design by maximizing *dB/dx* with the FEMM simulation.

The commercial model consists of two coils with opposite directions of current load in a figure 8 formation. Figure 2 below shows the cross section of the commercial model in FEMM, in which the coils are 0.5cm in height, and for each coil the inner and outer diameter is 2cm and 4cm, respectively. A new TMS stimulator design is proposed by using two coils, running current along opposite directions for obtaining maximized magnetic field gradient. A larger radius coil and a relatively smaller radius coil, with radius ratio of 2 to 1, were top-down overlapped. By shifting the smaller coil along its horizontal direction to change their relative positions and adjust current flow ratio between them we can optimize the magnetic field gradient which is closely correlated to grad(E).

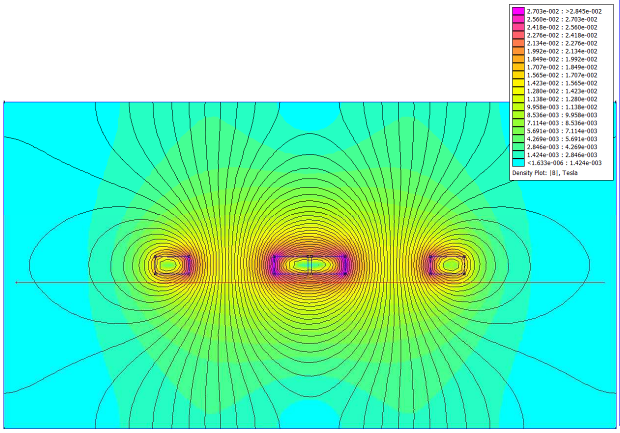


Figure 2. Figure-8 TMS coil magnetic field distribution

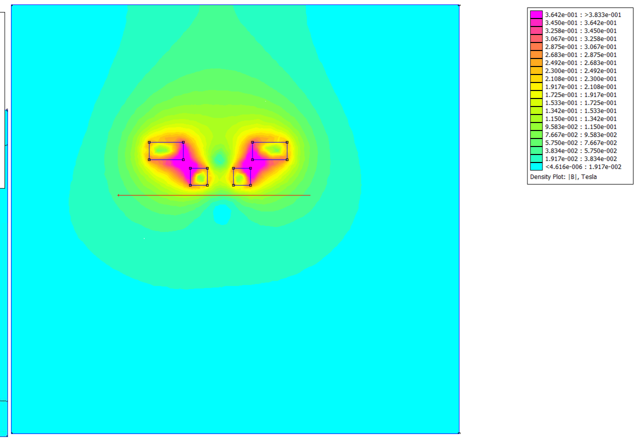
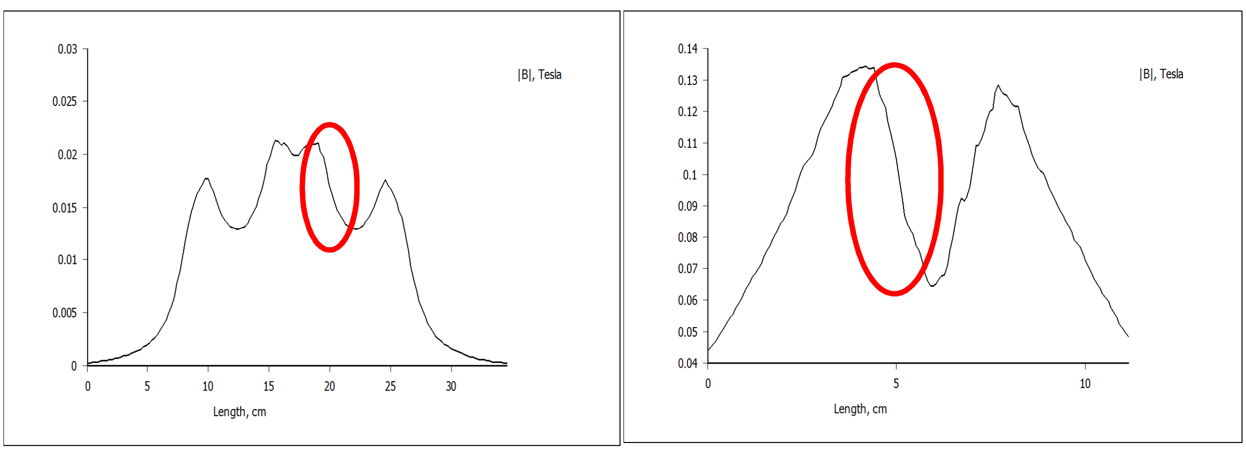


Figure 3. Coil design in FEMM for higher fiend gradient

We found that the highest magnetic field spatial gradient was able to be achieved when the inner diameter of big coil wiring is aligned with the outside diameter of the smaller coil. Figure 4 compares the magnetic flux density distributions of the figure-8 coil and our design along a cutline 0.5cm below the downside surface of the coils, which indicates a much higher magnetic field changing rate of the new design along the cutline than the commercial figure-8 coil. Further study on the current ratio of the up and down side coils in our design proves that by adjusting the current excitation in the large coil and keeping the current in the small coil 100A as a constant, we found that the optimized current ratio (big coil current to smaller coil current) was from 1.5:1 to 1.7:1 in this case. With the two coil in antiphase operations we can achieve the same grad(E) amplitude at a lower total current compared with the conventional structure, like the figure-8 structure case.



a b

Figure 4 a. Magnetic flux density distribution 0.5cm below figure-8 TMS coil; b. Magnetic flux density distribution 0.5cm below the coil designed in this study. Red circles show high gradient regions.

Reference:

[1] Roth, B. J. and P. J. Basser "A model of the stimulation of a nerve fiber by electromagnetic induction." IEEE Trans Biomed Eng **37**(6): 588-597, 1990.