A Comparative Study on a New Coil Design with Traditional Shielded Figure-of-Eight Coil for Transcranial Magnetic Stimulation

Yinan Li[1], Student *Member, IEEE,* and Babak Fahimi[1], *Fellow, IEEE*
[1]Department of Electrical Engineering, University of Texas at Dallas, Richardson, TX 75080, USA

Transcranial magnetic stimulation (TMS) is a non-invasive neuro-stimulation and modulation technique based on the principle of electromagnetic induction that uses brief, strong magnetic pulses of electric current delivered to a coil placed on the subject's head to induce an electric field in the brain. This electric field could stimulate and modulate neural activity and repetitive E-field inside brain generated by repetitive TMS (rTMS) can produce changes in neural activity that extends beyond the period of stimulation. Therefore, rTMS can be used as a probe for exploring higher brain functions and as potential treatment technique for psychiatric and neurological disorders. However, based on different excitation coil shape, a trade-off exists between the excitation focality on the brain and the depth to which the field penetrates. In this paper, a new design of TMS coil is proposed to compare with the traditional figure-of-eight (FOE) coil. Three-dimensional (3-D) finite-element method (FEM) has been used to simulate and compare the electric field induced inside the model of a human brain under the influence of both coil design.

***Index Terms*— coil design, electric field, focality, penetration depth, TMS**

I. Introduction

Transcranial magnetic stimulation is a novel non-invasive method to stimulate a small region of brain by using brief but intense pulses of electric current delivered to a coil on top of the subject's head [1]. This induced field can have sufficient magnitude and density to depolarize neurons, modulate the neural transmembrane potentials and cortical excitability even beyond the duration of the stimulation when TMS pulses are applied repetitively, which has behavioral consequences and therapeutic potential [2]. As a result, rTMS is now emerging as one of the basic new treatments in psychiatric practice for a considerable number of years [1]. In a given test of TMS, two features are of fundamental importance, the focality of the field and the penetration depth of the field. However, while the induced field can be controlled to confine in a small area by using small coils, the penetration depth is limited by the rapid attenuation of the electric field through human skull and brain tissue. Although coils with larger dimensions generate electric field that travels deeper, it is less focal [3]. In this paper, a new coil configuration is proposed which can induce an electric field that has a deeper penetration depth than the commonly used FOE coil while maintaining the focality at the same time. The specific absorption rate (SAR) and heat induced in the brain are at safe levels, but the risk of unintended neuro-modulation and seizures with such coil has to be further evaluated.

II. Methods

Traditionally, TMS has been restricted to superficial cortical targets because the dorsolateral prefrontal cortex has been the most common target in depression in cortical-subcortical-limbic network [3]. However, recent evidence suggests that the subgenual anterior cingulate cortex (sACC) is also central to this network, and other targets including the ventral portion of the anterior limb of the internal capsule, adjacent dorsal ventral striatum, nucleus accumbens and habenula are not superficial as well [4]. Therefore, accessing them with TMS would require the development of new coil designs in which the induced E-field is able to penetrate deep enough.

A. *TMS Coil Design Overview*

Although not generally acknowledged, it is usually the case that coils with larger dimensions generate electric field that can penetrate deeper than those with small coils whose electric field generated is more focal. As can be seen in the first TMS system, the coil used is small circular coil [5]. However, the field induced is circular ring-shaped and the penetration depth is very shallow. Until now, the milestone improvements of TMS coil design in terms of focality is the FOE coil [6], which consists of a pair of circular wire loops placed adjacent to each other in which current flows in opposite directions so that it can produce a relatively focal E-field right under the location where the two loops meet. The field focality can be further improved by introducing a metal shield below the coil with a window cut at the location where the two loops meet, as shown in Fig. 1. In order to improve the penetration depth, a group of coil designs called H coils has been proposed in [7]. Those H coils have a fairly complex winding patterns with large dimensions. Therefore, they are expected to have less E-field attenuation with depth.

B. *New Coil Design*

By integrating the features of both H coils and FOE coil, a new coil design is proposed, which is shown in Fig. 2. It consists of six FOE coils of different sizes overlap with each other at different angles. The tilt angle of each coil is designed so that it does not touch the skull surface. It retains the feature of FOE coil by concentrating the current flow at the desired location and amplifies the effect by more wires. In the meantime, similar to the H coils, it has a large dimension compared to conventional FOE coil. Therefore, it is expected to have slower E-field attenuation and concentrated focality point. The electric field E induced in the tissue is obtained by the virtue of Faraday's law of induction:

$$\nabla \times E = -\frac{\partial B}{\partial t} \qquad (1)$$

where B is the magnetic field generated by the coils, given by Biot-Savart Law:

$$B = \frac{\mu_0}{4\pi} \iiint_V \frac{(JdV)\times \mathbf{r}'}{|\mathbf{r}'|^3} \qquad (2)$$

where the integration is with respect to dV along the volume of the coils, and μ_0 is the permeability of free space which is the same as that in the biological tissue. Generally, the total electric field generated in the tissue is comprised of two parts: the primary and secondary. They are given by:

$$E = -\frac{\partial A}{\partial t} - \nabla\phi \qquad (3)$$

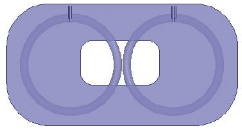


Fig. 1. Figure-of-eight coil with shield plate

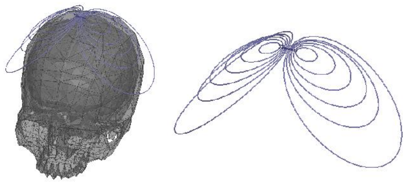


Fig. 2. New coil design with human head model

where A is the magnetic vector potential of the coils, and ϕ is the electrostatic potential which results from the non-uniformity of electric charges caused by conductivity changes along the path of current. This electrostatic potential follows Laplace's equation:

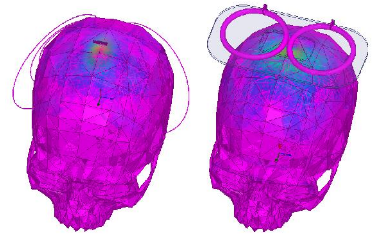
$$\nabla^2\phi = 0 \quad (4)$$

C. *Electric Field Simulation*

Using this method, both coil configurations were simulated in this paper. The human head model was obtained from the human body toolbox supplied by ANSYS which consists of two parts, skull and brain, with an isotropic electrical conductivity of 0.42S/m and 1.16S/m, respectively. The other distinct head tissue layers were not differentiated because the magnetically induced E-field in a nearly spherical shape is insensitive to radial variations of conductivity [8]. Fields are assumed to be time-harmonic with a frequency of 1.5GHz. The key features of electric field, penetration depth and focality, are quantified by half-value depth, d and half-value area, A, respectively. Half value depth is the distance from the surface to the point where the E-field strength is half of its maximum value on the surface and the half-value area is the area of the brain cortex where the E-field is greater than half of the maximum E-field strength.

III. Results and Conclusion

In the simulation, the point where the two loops meet in the FOE coil is set at 17mm from the head, the same as the point where the several loops meet in the new coil design. The focality was calculated on both the skull surface and the brain cortex. As can be seen in Fig. 3, the new coil design has a smaller focal area on the skull surface. Fig. 4 demonstrates that the field attenuates slower in the new coil design, where its half-value depth is 7.7mm, compared to the 3.9mm in the shielded FOE coil. Table I summarizes the performances of both coils. Because of the new coils' large dimension, the field starts to scatter inside the object and slowly lose its focality advantage over the FOE. As one may notice on the brain cortex surface, the focal area is larger in the new coil design. However, it still maintains this advantage in a certain range of depth. At 3.9mm (half-value depth of FOE) below the skull surface, the field is more focal on the new design as shown in Table I compared to shielded FOE. Overall this new coil design achieves the goal of increasing of the penetration depth while limiting the area of field focality. After coil optimization in terms of tilt angle and number of coils, it will provide another coil option for medical technician when implementing TMS operations in terms of different trade-off considerations.

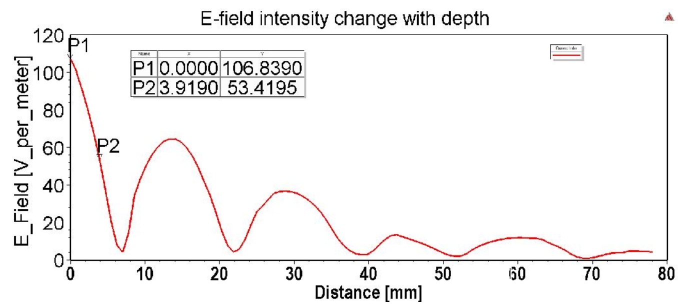


(a) (b)

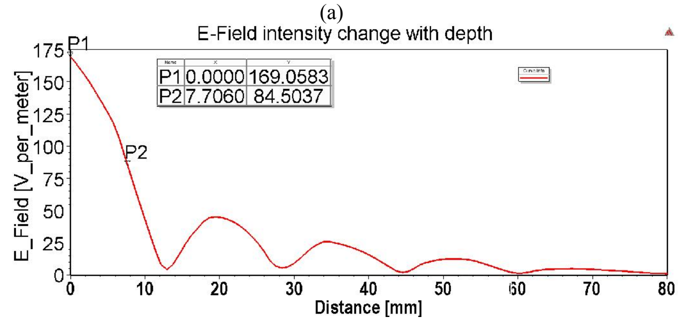
Fig. 3. E-field distribution on human head under (a) the new coil design and (b) the shielded figure-of-eight coil

TABLE I. FOCALITY AND PENETRATION DEPTH COMPARISON

Parameters	Shielded FOE coil	New coil
Half-value depth (mm)	3.919	7.706
Half-value area on head (cm^2)	16.62	12.56
Half-value area on cortex (cm^2)	19.63	30.19
Half-value area at 3.9mm from skull	17.02	15.69



(a)



(b)

Fig. 4. E-field intensity change with depth on (a) the shielded FOE coil and on (b) the new coil design

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