

EFFECTS OF TMS COIL GEOMETRY ON STIMULATION SPECIFICITY

Mehran Talebinejad, *Member, IEEE*, and Sam Musallam, *Member, IEEE*

Abstract—Transcranial magnetic stimulation has become an established tool in experimental cognitive neuroscience and has more recently been applied clinically. The current spatial extent of neural activation is several millimeters but with greater specificity, transcranial magnetic stimulation can potentially deliver real time feedback to reinforce or extinguish behavior by exciting or inhibiting localized neural circuits. The specificity of transcranial magnetic stimulation is a function of the stimulation coil geometry. In this paper, a practical multi-layer framework for the design of miniaturized stimulation coils is presented. This framework is based on a magnet wire fabricated from 2500 braided ultrafine wires. Effects of coil bending angle on stimulation specificity are examined using realistic finite element method simulations. A novel stimulation coil with one degree of freedom is also proposed that shows improved specificity over the conventional fixed coils. This type of coil could be potentially used as a feedback system for a bidirectional brain machine interface.

I. INTRODUCTION

TRANSCRANIAL magnetic stimulation (TMS) is a well known method for non-invasive stimulation of neural circuits [1]. The large number of clinical and experimental studies employing TMS have established this method as safe with minimal risk [1-2]. The goal of the current study is to increase the stimulation specificity for use in neural prosthetic systems [3-4] and bidirectional brain machine interfaces.

TMS coil greatly affects the stimulation efficiency, in terms of power consumption and stimulation specificity. The simplest form of TMS coil is a single spiral with 5 to 50 turns and a diameter in the order of 5 to 10 cm. Mathematical approaches for optimizing TMS coil characteristics including stimulation focality have been proposed [5-9] but have not materialized for two major reasons: 1) limitations of proposed mathematical approaches in terms of realistic dimension and material specification and easy to manufacture simulations, and 2) limitations of technology.

Although, several complex TMS coil designs such as

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M. Talebinejad is a postdoctoral fellow at the McGill University, Department of Electrical Engineering, Montreal, Quebec, CANADA (email: mehran.talebinejad@mail.mcgill.ca).

S. Musallam is an assistant professor at the McGill University, Department of Electrical Engineering and associate member of the Department of Physiology, Montreal, Quebec, CANADA (phone: 5143981702, fax: 5143984407, email: sam.musallam@mcgill.ca).

multi-loop coil, double butterfly coil, hatband or tiara coil, and combination of dozens of small coils [10] have been suggested, most of the existing devices still use a double-coil, also known as figure-8. The double-coil is more focal compared to single coil, but consumes more energy and has a larger coil area. Inner and outer diameter of the single coils forming the double-coil and the bending angle between coils also affects the focality. An increased bending angle results in increased focality [11-13]. An increased inner and outer coil diameter also results in higher magnetic intensities at the expense of larger coil area [14].

In this paper, a multi-layer framework for the design of miniaturized TMS coils is presented. A magnet wire with 0.7 mm thickness fabricated from 2500 braided ultrafine wires is presented which is the basis for the coil design. Effects of coil bending angle on specificity of the stimulated area, are examined using finite element method simulations. A coil with one degree of freedom is also proposed that shows improved specificity over the conventional fixed coils for stimulation of different areas of brain.

II. METHODS

A. Wire

TMS coil wire must be rated for high voltages up to 3,000 V, with a medium frequency (< 200 Hz). Round or square wire diameters have minimum requirements at high voltages and it is very difficult to design copper magnet wires with small diameters (< 3 mm). Previously authors have shown rectangular wires with very small thickness could be used for tighter windings that allow more turns in a smaller space and outperform the round and square wires for miniaturized TMS coil [14]. It is also demonstrated that the braided *Litz* wire reduces the undesirable effects of induced eddy currents and skin effects inside the coil [15].

We fabricated the rectangular *Litz* wire described in [16] with the standard wire processes as shown in Fig. 1. This wire is 7.1 mm wide, 0.7 mm thickness (minimum requirements for 3,000 V, 200 Hz) and is comprised of 2500 (maximum number of wires that could be used in these dimensions) ultrafine $2 \mu\text{m}^2$ round wires. Each wire is individually insulated with film (*Litz* configuration). The thickness specially makes this wire suitable for horizontal-spiral coils that are designed to have small areas such as TMS coil [14].

B. Coil Geometry

TMS coil is conventionally approximated as a cylinder in the mathematical models [11-13]. We used 3D models drawn in AUTOCAD with realistic geometrical parameters including, wire length and winding spaces. The 3D models are later modeled using finite element method in COMSOL Multiphysics. Fig. 2, shows a 3 layer double-coil. Multi-layers enhance the overall magnetic field intensity in this framework [14]. Current runs in the same direction at the adjacent points of side by side coils. The wire length in each single coil is 774.169 mm. The angle between the coil surface and central axis (bending angle) is 0° in Fig. 2. This coil is the basis of the experiments presented here with a varying bending angle along the central axis.

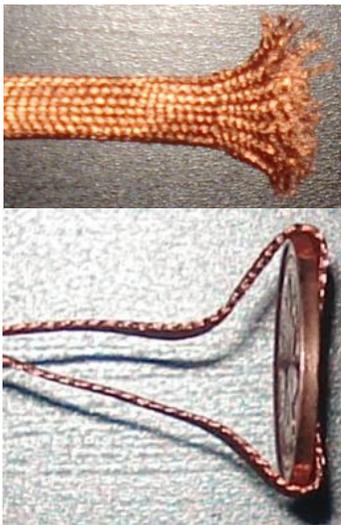


Fig. 1. Braided Litz wire fabricated from 2500 braided ultrafine round wires with film insulation. Width is 7.1 mm and the thickness is only 0.7 mm compared to 1 mm thickness of a coin.

C. Stimulator Circuit

A single phase stimulator circuit is conventionally comprised of a high voltage source that charges a capacitor which is discharged into a coil via a heavy gauge high conductive axial cable. This combination behaves like an oscillating RCL circuit. The oscillatory current in the coil $I(t)$, is determined by the capacitance of the discharging capacitor C , inductance of the coil L , and lumped resistance of the circuit and connecting wires R as,

$$I(t) = (V_C/L\omega)e^{-\alpha t} \sin \omega t, \quad (1)$$

where, $\alpha = R/2L$, $\omega^2 = (LC^{-1}) - \alpha^2$, and V_C is the initial charge of the capacitor [15]. In this work, following [15], capacitance and initial charge are set to $C = 100 \mu\text{F}$ and $V_C = 3,000 \text{ V}$. The inductance and resistance of the coil are set based on the measurements made from the magnet wire described above which are $15.39 \mu\text{H}$, and $35.34 \text{ m}\Omega$ respectively. There are other inductive and resistive contributions from the circuit and connecting wires.

D. Materials

The simulation framework consists of 6 different materials summarized in Table I with their electrical properties. Following [16], the head was simulated with 4 multiply: 1) 1mm scalp, 2) 3 mm skull, 3) 4 mm cerebrospinal fluid (CSF), and 4) brain. The wire properties are set according to [15] for standard 99 % copper with film insulation. The head and the coil are surrounded by air in the simulations. In this framework a 3,000 V input results a magnetic field greater than 3 Tesla on the coil surface which is sufficient to induce currents for stimulation up to 1cm inside the head model.



Fig. 2. Isometric, three layer double coil with bending angle 0° , inner diameter 8 mm, and outer diameter 30 mm. There is 0.1 mm space between the turns and 1 mm space between the layers.

E. Finite Element Simulations

The COMSOL Multiphysics, AC/DC Module v.3.4, was employed for finite element method simulations. Current density in the coil was specified azimuthally as,

$$J_\phi = (j\omega\sigma - \omega^2)A_\phi + \nabla \times (\mu^{-1}\nabla \times A_\phi), \quad (2)$$

where ω is the angular frequency consistent with C and L , σ is the electric conductivity, μ is the relative permeability and A_ϕ is the magnetic vector potential. Computing Eq. 2, leads to the solutions for induced currents, magnetic field intensity, flux density, and resistive heat [16]. These parameters are computed on a uniform 0.01 mm triangle mesh using the finite element method. In this work, we consider the flux density inside the head model as a measure of stimulation area to compare stimulation specificity among different coils.

F. Simulation Setup

The TMS coil (Fig. 2), was placed on the model head (Fig. 3), at the top 2 elliptic areas E_1 , E_2 . In each area coil was placed horizontally and vertically with 0.1 mm distance from the surface of scalp at the coil center [2]. Then, the bending angle was set for each area individually to have an equal distance between the two ends of each coil and the surface similar to realistic placement. Total of 6 different sets with empirical bending angles were simulated. In E_1 , the coil was placed horizontally and vertically (E_{1H} and E_{1V} , 2 sets). In E_2 , the coil was placed horizontally and vertically with its center at the front E_{2F} , and at the side E_{2S} (E_{2FH} , E_{2FV} , E_{2SH} , and E_{2SF} , 4 sets). For comparison 2 fixed angle coils at 0° and 10° (similar to commercially available coils) is also placed at E_1 and E_2 , horizontally and vertically (E_{1H} , E_{1V} ,

E_{2FH} , E_{2FV} , E_{2SH} , and E_{2SV} , 2×6 sets). A single 3 layer coil and two reverse engineered coils from MagVenture (B35), with 1 and 5 cm, inner and outer diameter and square wire (3×3 mm) windings single and double-coil are used for comparison in this setting.

TABLE I
MATERIAL ELECTRICAL PROPERTIES

TYPE	RELATIVE PERMEABILITY	CONDUCTIVITY (S/m)
AIR	1	0.000
COPPER	1	60×10 ⁶
SCALP	1	0.500
SKULL	1	0.001
CSF	1	2.000
BRAIN	1	0.500

G. Stimulation Specificity

Focality of stimulation is usually defined as a function of flux density on the stimulation area. A more focal stimulation means a smaller area is stimulated with the maximum magnetic flux density. On the other hand the coil area also affects the stimulation area and specificity.

In this work, a specificity index S_I is defined as,

$$S_I = \left[A_F \times A_C (A_F + A_C) \right], \quad (3)$$

where A_F is the area stimulated with 90 to 100% of the flux density at a given depth, and A_C is the coil area both with the same metric (e.g., mm). Smaller A_F and A_C give smaller S_I which indicates more specificity and is unitless.

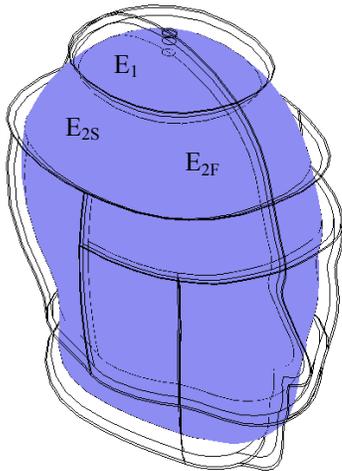


Fig. 3. Four multiply head including, scalp, skull, CSF, and brain. Areas E_1 , E_{2S} and E_{2F} are indicated within the top 2 elliptic areas.

III. RESULTS

Table II summarizes all the results. Single coil shows the highest specificity (lowest index). An increased bending angle gives an increased specificity as it decreases A_F and A_C which is consistent with [11-13]. The higher specificity of the novel coils with one degree of freedom compared to the

fixed coils is also due to the higher bending angles. Among novel coils, specificity is the highest at E_1 and E_{2F} , which is due to the higher bending angle in these areas when coil is vertically positioned. The commercial coils have a higher index which is expected from a higher diameter. Double coil is only placed at E_1 horizontally; the specificity index is consistently higher for this 5 cm coil regardless of the position compared to the 3 cm coil presented. The new wire provides 13 turns for a 1 and 3 cm, inner and outer diameter coil; but the square wire only allows 9 turns for a 1 and 5 cm, inner and outer diameter coil.

TABLE II
COIL GEOMETRIC PROPERTIES AND SPECIFICITY INDEX

COIL TYPE	PLACEMENT	BENDING ANGEL	SPECIFICITY INDEX AT 2 cm	# OF SIDES /LAYERS /TURNS
SINGLE 3 LAYER	N/A	N/A	12	3×13
FIXED	E_{1H}	0°	42	2×3×13
DOUBLE	E_{1V}	0°	48	2×3×13
FIXED	E_{2FH}	0°	42	2×3×13
DOUBLE	E_{2FV}	0°	48	2×3×13
FIXED	E_{2SH}	0°	42	2×3×13
DOUBLE	E_{2SV}	0°	48	2×3×13
FIXED	E_{1H}	10°	37	2×3×13
DOUBLE	E_{1V}	10°	39	2×3×13
FIXED	E_{2FH}	10°	37	2×3×13
DOUBLE	E_{2FV}	10°	39	2×3×13
FIXED	E_{2SH}	10°	35	2×3×13
DOUBLE	E_{2SV}	10°	36	2×3×13
1°FREEDOM	E_{1H}	14°	33	2×3×13
1°FREEDOM	E_{1V}	16°	29	2×3×13
1°FREEDOM	E_{2FH}	13°	32	2×3×13
1°FREEDOM	E_{2FV}	16°	29	2×3×13
1°FREEDOM	E_{2SH}	12°	34	2×3×13
1°FREEDOM	E_{2SV}	14°	33	2×3×13
B35 SINGLE	N/A	N/A	35	3×9
B35 DOUBLE	E_{1H}	10°	98	2×3×9

IV. DISCUSSIONS

The rectangular braided *Litz* wire with 0.7 mm thickness is suitable for horizontal-spiral and miniaturized TMS coils; this wire allows tighter windings (more turns) compared to the round and square wires. This will lead to greater magnetic field intensity which is reduced when the coil diameter is decreased.

Results show the double-coil provides more focality, but increased area of two coils lowers the overall stimulation

specificity compared to a single coil. Hence, single coil is the best candidate for specificity. Adding layers to the coil is one way to increase the overall magnetic intensity while the coil area is constant; however, adding more than 3 layers does not show significant contribution to the overall induced currents.

The novel coil with a degree of freedom clearly outperforms the conventional fixed double-coil; smaller coil area and higher focality due to a larger bending angle increase the specificity. Thus enhancing double-coil with this new feature will improve the specificity in cases where stimulation of double-coil is needed; induced current of a double-coil is characteristically different from a single coil. The double-coil with one degree of freedom could be designed with a displacement system between the two single coils.

Our goal is to use TMS for a bidirectional brain machine interface, to reinforce volitional movements, providing feedback only when movements of external devices are willed. However, current coils are not well suited for this application due to the inadequacy of stimulation specificity; meanwhile, specificity of the proposed single coil with braided *Litz* wire is 3 times higher than B35.

V. CONCLUSIONS

This work shows smaller coils and magnet wires are a good solution while instrumentation barriers for realization of multi-channel stimulators with arrays of dozens of coils are removed.

The proposed novel coil with one degree of freedom is potentially a useful enhancement to improve the stimulation specificity of a double-coil.

Application of the novel coil could be extended to small animals; it is expected to see higher specificity when the head is smaller as the bending angle could be increased further.

During stimulation period the coil experiences physical changes that result into acoustic noise as high as 140 dB in some cases [2]. For a bidirectional brain machine interface this acoustic noise is not desirable. Preliminary results show the braided *Litz* wire produces less acoustic noise compared to single conductor and stranded wires.

Future research will further examine acoustic noise and actual coils in practice.

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