

Angle-Tuned Coil: A Focality-Adjustable Transcranial Magnetic Stimulator



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1 Introduction

Transcranial magnetic stimulation (TMS) has been approved by the Food and Drug Administration (FDA) for treatment-resistant major depression [1] and Obsessive-Compulsive Disorder [2]. Its therapeutic effects in other psychiatric and neurological disorders, including drug addiction, are emerging [3, 4]. From both clinical and basic neuroscience perspectives, there has been a strong demand for stimulation tools that can reach deep brain regions with small size targeted stimulations. For example, decades of neuroimaging studies have identified malfunction of dorsal anterior cingulate cortex, insular and amygdala in a range of psychiatric disorders. These structures are 4 cm or more below the scalp. Unfortunately, with current technologies, the stimulation targets are limited to superficial brain regions, or otherwise wide brain areas are stimulated when a deep brain structure is targeted.

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The output of a TMS coil can be treated as field emission from a finite size aperture and follows a specific depth-focality tradeoff rule. Deng et al. theoretically calculated the depth-focality profiles of 50 TMS coils [5]. Two groups of mainstream coils, circular and the figure-8, formed two depth-focality tradeoff curves, respectively. The study concluded that at shorter depth, which is smaller than 3.5 cm, all figure-8 type of coils follows a better depth-focality tradeoff rule and it will be advantageous to use the figure-8 coil. A number of studies have attempted to design TMS coils for enhanced the penetration depth or improved the focality. Rastogi modified conventional figure-8 coil to improve its focality but that, on the other hand, significantly weakened the electric field strength generated in brain tissues [6]. Crowther suggested a “Halo coil” design to improve the penetration depth of the conventional circular coils [7, 8], but this design sacrifices the coil’s focality. Luiz modeled multi-channel coil arrays to improve the focality and penetration depth profile [9]. However, this design involved complicated coil structures, and it required higher efficiency of the coils’ cooling system. Alternative coil design strategy is needed to go beyond the depth-focality tradeoff limitation. Roth et al. have developed the H-coil for human deep brain stimulation, but the design is still limited by the depth-focality tradeoff with a relatively large field spread [5, 10, 11]. We recently reported a multi-layer winding-tilted coil design approach for focused rodents’ brain stimulation that induced unilateral movements [12]. The goal of this study is to extend this novel strategy to design TMS coils for deep and focused human brain stimulation and compare its depth-focality characteristic with other conventional coils.

2 Methods

Figure 1a illustrates our TMS coil design (see reference [12] for details). In this study, the coil dimensions have been extended for human brain stimulation. For each single circular coil, the inner and outer diameters are 8 cm and 9 cm, respectively. The thickness of each single coil is 1 cm and 5 circular coils are accumulated along the central axis in this model. Considering the overall inductance of the coil model, we further extended the coil length while using narrow (lower value of d_o-d_i) but thick (higher value of h) coil windings, so that the total turn number could be low enough to control the coil inductance.

The head model is a homogenous sphere with the diameter of 17 cm and isotropic electrical conductivity of 0.33 S/m^{-1} . The definitions of stimulation depth and focality in Fig. 1b are based on the half-value depth ($d_{1/2}$) and half-value volume ($V_{1/2}$). Modeling frequency was 5 kHz. All conditions are identical to the modeling in the study by Deng et al. [5], except that the software we used for calculation is the COMSOL AC/DC module (finite element analysis software, COMSOL Inc.), which is different from that in their study. To calibrate our calculation using COMSOL, we firstly selected 4 coils, which were already documented in the study by Deng et al., and compared our results with theirs. The selected coil models for calculation calibration were the 50 mm, 70 mm and 90 mm circular and the 70 mm figure-8 Magstim coils.

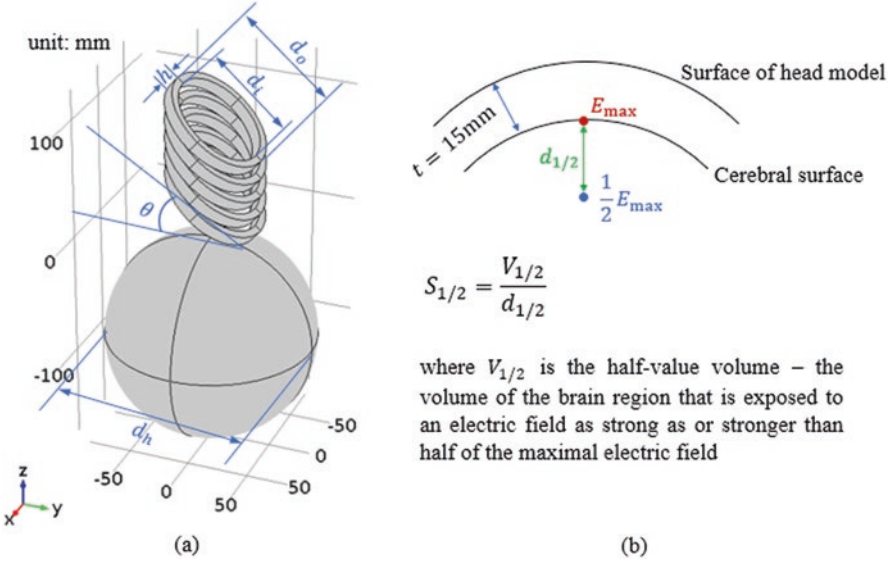


Fig. 1 (a) Simulation model and induced electric field distribution in the human head model; (b) The definitions of the stimulation depth and focality

Table 1 Comparison of the stimulation depth ($d_{1/2}$) and focality ($S_{1/2}$) calculated by COMSOL with data recorded in Deng’s study

Coil type	50 mm circular magstim coil	70 mm circular magstim coil	90 mm circular magstim coil	70 mm figure-8 magstim coil
$d_{1/2}$ (by COMSOL) / cm	1.31	1.49	1.75	1.45
$d_{1/2}$ (by Deng et al.) / cm	1.29	1.44	1.74	1.41
$S_{1/2}$ (by COMSOL) / cm ²	53.1	65.8	87.8	13.8
$S_{1/2}$ (by Deng et al.) / cm ²	53.7	66.0	87.4	14.8

3 Results

3.1 Cross-Validation of Theoretical Simulation

The stimulation depth and focality calculated by COMSOL of the 4 selected coils (50 mm, 70 mm and 90 mm circular and the 70 mm figure-8 Magstim coils) as calibrations are listed in Table 1, compared with the data in Deng’s study [5]. Both the stimulation depth and focality from the two finite element analysis software matched reasonably well with minor differences.

Theoretical simulation of multi-layer winding tilted coil design.

For our multi-layer winding tilted coil design, we firstly investigated how the tilt angle θ affected the stimulation depth and focality. The angle was adjusted from 0 degree to 70 degrees with a step of 10 degrees without changing any other parameters in the modeling. The values of the stimulation depth and focality were marked in the same plot summarized in Deng et al. in Fig. 2 [5]. Coils with 0-degree tilt angle (or flat coils) are located along the circular coil curve. As the tilt angle increases, the location of the coil in the depth-focality tradeoff plot moves from the circular coil curve towards the figure-8 coil curve. If the number of the winding layer is set to 5, the coil location drops on the figure-8 coil curve when the tilt angle reaches 50 degrees. Further enlarging the tilt angle enables the curve, which is plotted from the trace of our coil designs (formed by the blue square dots in Fig. 2), to penetrate the figure-8 coil curve. For example, when the tilt angle is adjusted to 60 or 70 degrees, the locations of the coils in the plot are below the figure-8 coil curve, and that indicates a better depth-focality characteristics than the existing TMS coils. The number of the winding layers are also adjusted to 2 and 9. We finalize that a smaller number of winding layers moves the curve of the coil in the plot towards the left side. For a certain tilt angle, both the stimulation depth and focality decrease. However, when the winding layer number increases from 5 to 9, the stimulation depth is not considerably improved. This phenomenon may be cause by the longer distance from the stimulation target to the few top layers, which significantly weakens the electric field strength at the stimulation target contributed by those layers. Figure 3 presents the induced electric field distribution on the human head model surface by the proposed coil designs with various design parameters.

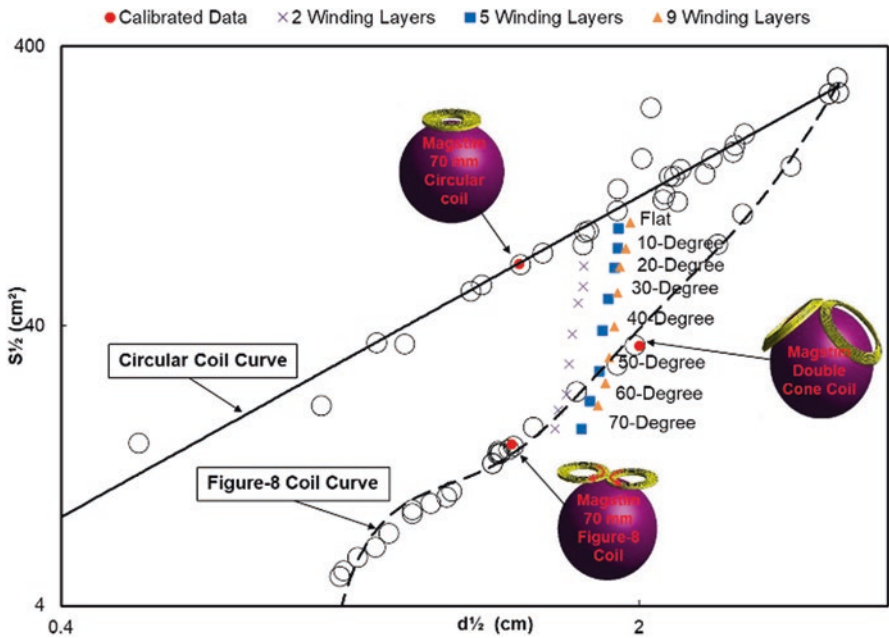


Fig. 2 Calculations of stimulation depth and focality for the multi-layer winding-tilted coil design with air core and tilt angle ranging from 0 to 70 degrees (green curve), the number of winding layers of 2, 5 and 9

To further demonstrate the advantage of our coil design, we compared the elec-

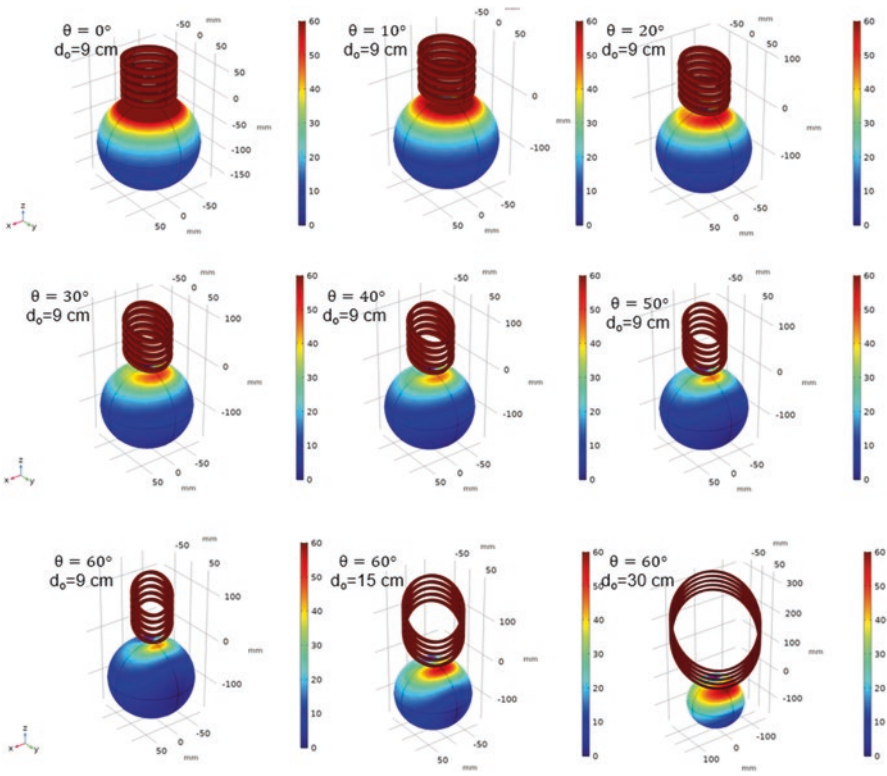


Fig. 3 Induced electric field distribution on the human head model surface by coil designs with different design parameters

tric field decay rate in the brain model for 6 coil structures, as shown in Fig. 4, and experimentally demonstrated the tilt angle could improve the focality of the induced electric field distribution. The electric field decay rate curves in the human head model in Fig. 4 indicate that neither the application of ferromagnetic core nor the tilt angle θ is able to improve the field decay rate, but the winding accumulation along the coil’s central axis considerably improves it. For example, at the depth of 3 cm, a 5-layer winding accumulation improves the electric field decay rate by 4–5%.

3.2 Experimental Validation

To verify how the angle θ affected the focality, we fabricated 3 coil prototypes with winding’s tilt angles of 20, 10 and 0 degree. They shared the same coil length, inner and outer diameters, which were 4.4 cm, 3.8 cm and 7.5 cm respectively. Each coil

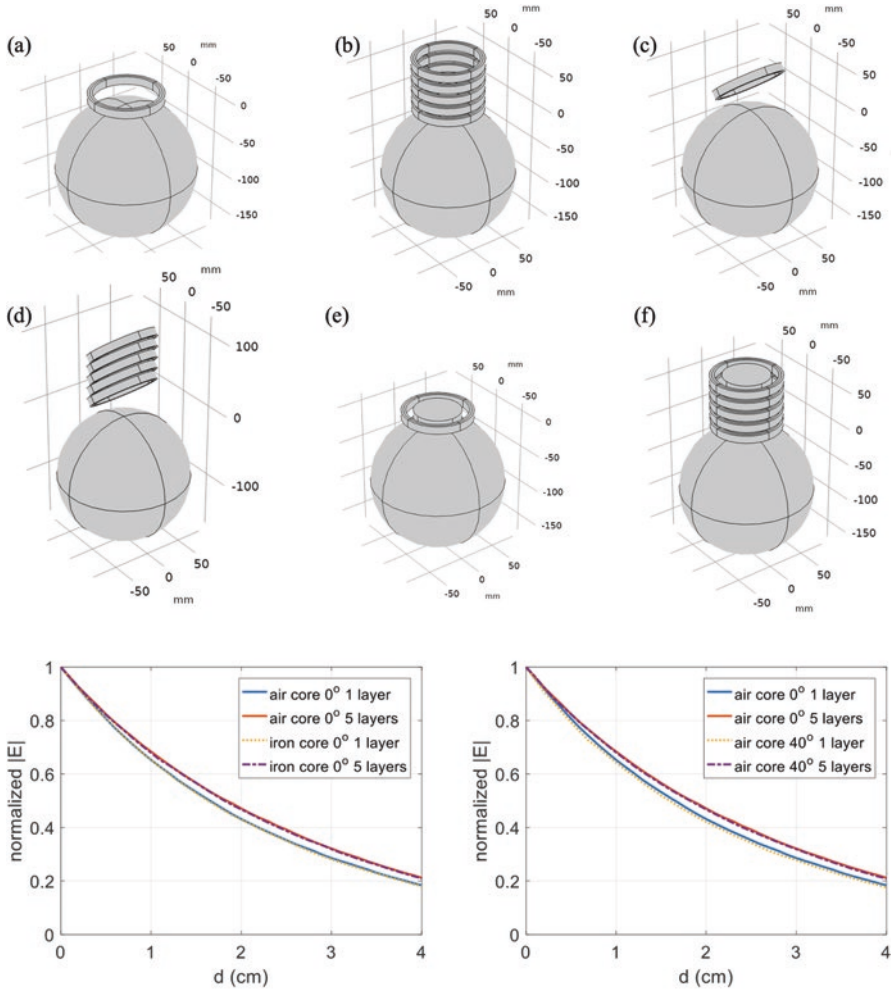


Fig. 4 Comparison of field decay rates for 6 different coil structure models: (a) single-layer flat circular coil with air core; (b) 5-layer flat circular coil with air core; (c) single-layer circular coil with air core and 40-degree tilt angle; (d) 5-layer circular coil with air core and 40-degree tilt angle; (e) single-layer flat circular coil with ferromagnetic core; (f) 5-layer flat circular coil with ferromagnetic core

was wrapped by 20 turns of the litz wires, and each turn contained a bundle of 135 piece of AWG30 wires. The TMS coil was driven by a customized driving circuit, in which an insulated gate bipolar transistor (IGBT) was used as the switch to control TMS pulses and a capacitor bank charged by a power supply was the main current source [13]. The charging voltage of the capacitor bank was set to 100 V and the pulse duration was 250 μ s. The induced electric field was measured with a modified Rogowski coil electric field probe customized in our lab [14]. The electric field was mapped in the medium of air within a plane 2 cm away from the coil surface. Since

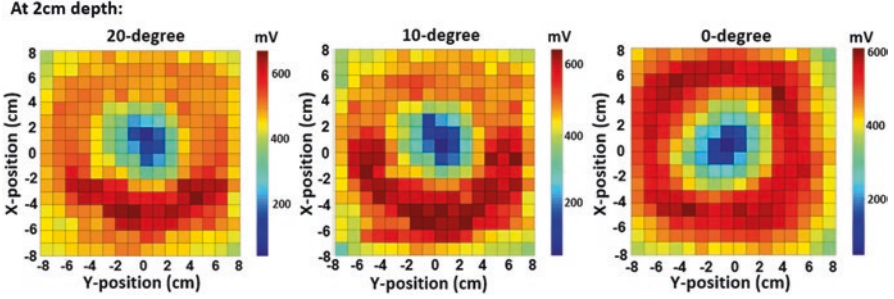


Fig. 5 Induced electric field measurements using modified Rogowski coil probe for the 3 coil prototypes with 20 degrees, 10 degrees and 0 degree wire wrapping tilt angle at the depth of 2 cm in air medium

the electric field components along the Z axis (in parallel with the coil's central axis) was small enough to be negligible, only the X and Y components of the electric field were measured. Considering the probe size, we mapped the electric field at a step size of 5 mm within the X-Y plane. An area of 8 cm \times 8 cm was scanned for each coil.

Figure 5 shows the heat maps of the electric field distributions for the 3 scanned coils. For the coil with a tilt angle of 20 degrees, the area with the field strength over or equal to 80% of its peak value (E_{peak}) is only 14.5 cm²; while for the other 2 coils, the values have reached 22.5 cm² (for 10-degree tilt) and 30.75 cm² (for 0-degree tilt), respectively. It is also found that a larger tilt angle of the coil windings would slightly increase the coil's inductance. For example, the inductances of the 3 scanned coils were measured to be 52.2 μH , 51.21 μH and 49.4 μH , respectively.

4 Discussions

The combination of air-core accumulated windings along the coil's central axis and the tilt angle of the windings provides a significant innovation to the depth-focality profile of TMS coils. The tilt angle technique is a mild symmetry breaking method. It does not reduce the equivalent field emission aperture size or speed up the electric field decay rate along the coil's central axis direction, but significantly distorts the ring shape electric field distribution, resulting in a much smaller focal spot. The only limitation of this coil design, to our knowledge, is its possible larger inductance compared with conventional TMS coils. The current flowing inside the coil $I(t)$ can be expressed as.

$$I(t) = \frac{V_c}{\omega L} \sin(\omega t) \exp(-\sigma t), \quad (1)$$

where ω and σ are related to the charging of the capacitors (C) in the stimulator circuit, the inductance of the coil (L), and the resistance in the LC circuit. V_c is the voltage to charge the capacitors [15]. The induced electric field $E(t)$ can be expressed as

$$E(t) = \alpha \frac{dI}{dt} \cong \alpha \frac{V_c}{L}. \quad (2)$$

So, E is inversely proportional to L . The Magstim human TMS coils always have an inductance of around 20 μH (from the Magstim Rapid [2] system manual) [16]. Our prototype human TMS coil design may have a higher inductance due to its length. This would require a stimulator of higher power output to drive the coil and enhance the current load in it, and on the other hand, the turn number of the coil can be reduced to limit its inductance to a reasonable value.

The curves we plotted for our multi-layer winding-tilted coil design in the depth-focality tradeoff profile in Fig. 2 demonstrate better focality than the H coils. The half-value depth values are comparable with the H coil designs. The half-value depth values are larger than the conventional figure-8 coils. Our coil design has achieved better depth-focality characteristic than a large sized double cone coil for human brain stimulation [5]. However, the double cone coil is known to induce scalp and facial pain and has limitations on its stimulation targeting site due to its geometry [17]. Our coil design has the advantage of much smaller contact area with the human head during the stimulation, and that may reduce or even exempt the scalp or facial pain. Moreover, with our coil design, it is feasible for users to conduct multisite stimulation, which cannot be accomplished by the double cone coils. Our design has provided the current best method for multisite human brain stimulation than all the conventional TMS coils considering both the stimulation depth and focality. The stimulation depth and focality are adjustable by the geometry of the coil design, for example, by tuning the tilt angle of the windings and the coil's outer diameter. This novel design provides a promising solution for the future deep and focused multisite human brain stimulation.

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Declaration of Competing Interest The authors declare that there is no conflict of interest for this manuscript.

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