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# Human Head Transcranial Magnetic Stimulation Using Finite Element Method

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## Abstract

Transcranial magnetic stimulation (TMS) is a wearable neuromodulation technique. It is approved for several therapies for various neurological disorders, including major depressive disorder, traumatic brain injury, Parkinson's disease, and post-traumatic stress disorder. This method became an alternative neuromodulation technique for such brain-related disorders. However, it has shown significant improvement in this alternative approach. Studies based on this technique have shown limited efficacy. They might be associated with current levels, poor coil locality, optimal coil size, and neuromodulator settings. It has been shown in this research that coil heating is related to higher levels of current. Thus, it is required to analyze the impact of the current levels on the induced magnetic distribution to define the optimal current range for the TMS coils. It is not feasible to investigate this research with experimental tests and analytic methods. Alternatively, using an advanced computational model of the coils and accounting for different human head anatomical layers, coil current capacity can be optimized based on finite element magnetic field distribution. This paper aims to investigate the impact of the coil current levels on the induced magnetic field distribution. The current capacity of the coils can be optimized based on the required magnetic field. In this way, the overheating may be reduced and may result in increased efficacy. As a proof-of-concept, a prototype coil and multi-layered geometrical human head models were generated using geometric shapes. The fundamental human head tissue layers were generated based on their average thickness. The model was simulated based on a finite element magnetic simulation using appropriate boundary conditions and neuromodulator settings. The various coil current levels were applied to analyze the outcome. The models were simulated, and the results were recorded based on these current levels. Results showed that there is a direct relation between applied current levels and induced magnetic flux density in the region of interest.

## 1. Introduction

There is a significant increase in alternative noninvasive brain stimulation methods and a growing domain of research and development for clinical neurophysiology. Thus, many applications have been generated for disease diagnosis and pathophysiologic investigation of cortical excitability changes, and mapping of cortical function (e.g., before brain surgery) [1]– [4]. Transcranial magnetic stimulation (TMS) is a noninvasive technique for neuromodulation that has a therapeutic effect on neurological disorders including major depressive disorder, traumatic brain injury, Parkinson's disease, and post-traumatic stress disorder [5]. TMS has become a promising, safe, and noninvasive alternative to medication for the treatment of brain-related disorders. It works based on the principle of electromagnetic induction. The current is applied through the copper coil wires (as shown in Fig.1) which in turn produces a magnetic field. The magnetic field pulse delivered by a stimulating coil applied on the scalp can pass through the skull bone to generate an electric field. After development of the first TMS [6], other studies have accompanied to improve and optimize the complex design of the TMS for specified purposes [7].

It has been shown that various parameters may influence the efficiency of the TMS. These parameters include the orientation and type of TMS coil [8], and the waveform of the magnetic pulse, which is usually

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monophasic or biphasic [9]. More importantly, the features (shape and size) of the magnetic coils have an important role in determining the focality and depth of stimulation in the brain. Although the coil with type shape of a number eight (called figure-of-eight-coil (FoE)) is widely used, double-cone coil seems to provide a deeper, stronger, and wider electric field [8]. Targeting a wider area with a double-cone coil may be more appropriate for patients [10].

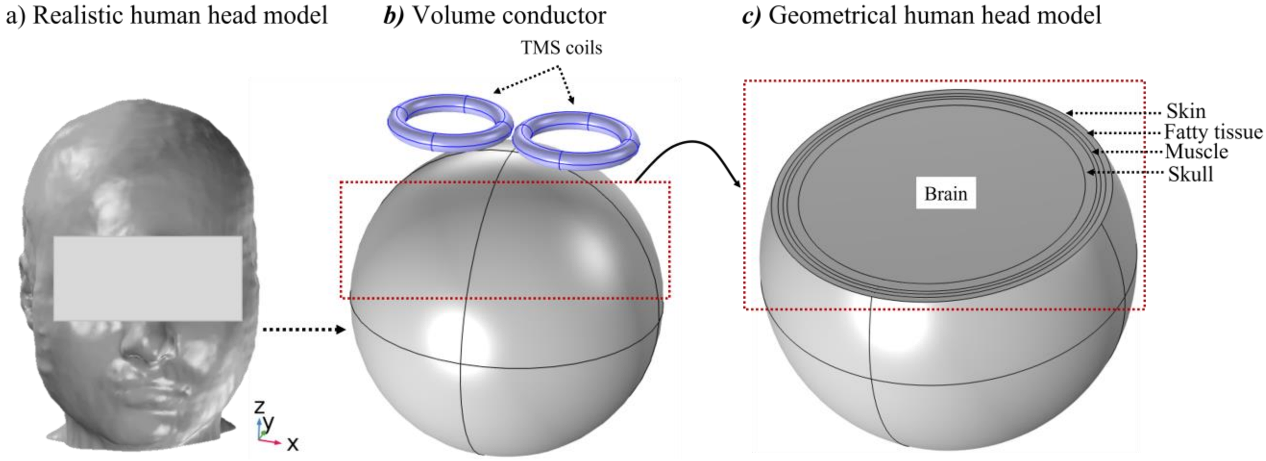


**Figure 1.** The computational model of the human head and TMS coils. (a) shows MRI-based human head model. (b) shows the spherical head model including the TMS neuromodulator. (c) shows human head tissue layers that are considered in this study. The human head was generated based on the average thickness of each tissue.

Although the importance of TMS has been further enhanced with the advent of new devices capable of repetitive stimulation, however, it has been shown that further improvements are required to have optimal TMS [11]. For example, the users are still complaining about poor focality, too rapid coil heating, and too large coil size [12], [13]. These are might be due to relatively applied higher levels of the current levels. Also, it is also shown that the high current levels heat the coil due to resistive losses and exert considerable mechanical forces in the coil windings, reducing their lifespan and causing a loud coil click [13]. In addition, the deeper structures of the human head can be simulated by applying a stronger electrical current to the coil, but this strategy invariably stimulates a larger brain volume which risks seizure in patients and is thus not allowed for safety considerations [13]. Thus, it is vital to investigate the impact of the magnetic field distribution within the human head layers using different levels of the coil current.

It may not be possible to parametrize TMS features (e.g. coil size and shape and current levels) to optimize TMS settings using experimental studies because of the lack of safety in testing them on human subjects. Alternatively, highly advanced computational methods can be used to optimize such parameters without inducing any risks [13], [14]. Such methods are implemented using finite elements (FE) and models (FEM). It consists of a volume conductor model that represents different structures and the electrodes according to their conductivities and appropriate boundary conditions [14]– [16]. Current commercial FEM software packages (e.g., COMSOL Multiphysics, ANSYS) allow calculating electrical magnetic induction in the computational models.

It has been shown that many coils were designed in the last two decades utilizing different geometrical layouts. The typical coil arrangement is figure-of-eight [1], [5], [17]. Other types of coil design may stimulate a wide region of the human head and this may affect neural populations whose simulation procedures have unwanted or interfering effects [13]. Thus, the typical coil arrangement was designed and used in this study as shown in Fig. 1(b). As proof of concept for TMS features optimization, the electrical magnetic distribution was simulated for two identical coil stimulation arrangements to analyze magnetic field distributions across the human head. It has also been shown that the human head can be represented by concentric geometrical shapes as shown in Fig. 1(b), (c) [8]. Thus, the computational model of the human head was generated based on five concentric geometrical shapes including the brain, skull, muscle, fatty tissue, and skin layers as shown in Fig. 1(c). After attaining the required electrical features of these layers and applying appropriate boundary conditions, the current was applied through the coils to calculate magnetic field distributions across the region of interest (e.g. brain). The results suggested that TMS neuromodulator settings (e.g. current level) may be parametrized and simulated by mimicking the anatomical volume conductor of the human head by applying commercially available TMS settings. It was shown that when the applied current was increased the induced magnetic flux density was also increased approximately in the same ratio.

The paper is organized as follows; Section II presents the procedure that was used to design to the model; Section III gives the magnetic field induction on the region of interest based on chosen based on coil arrangement using various current levels, and the discussion and conclusion are given in Sections IV and V, respectively.

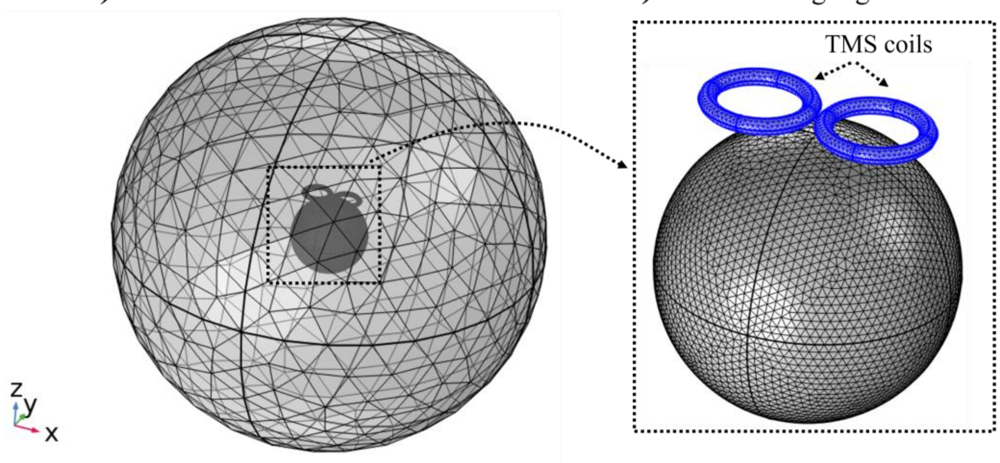


**Figure 2.** Numerical computation model of volume conductor. (a) Shows discretization of the whole model including air (ground) and fundamental anatomical human head layers. (b) Shows the region of interest meshing plot. The coils and anatomical layers are meshed with finer meshing settings to obtain accurate results. The coils are highlighted.

## 2. Materials and Methods

### 2.1. Volume Conductor Model

The computational model was generated using COMSOL Multiphysics (COMSOL, Ltd, Cambridge, UK) based on AC/DC module. The AC/DC module provides a unique environment for the simulation of electromagnetics in 2D and 3D. This module is a powerful tool for detailed analysis of the components of electromagnetism such as coil current and power dissipation. The human-head computational models can be constructed in a range of complexity from concentric sphere models to high-resolution models based on an individual's image data set depending on the clinical question [13]. The anatomically specific image-based head modelling may require extensive prior work on computational modelling. In the studies [12], [11], it has been shown that the geometrical human head (e.g. sphere) can be used instead of an MRI-based highly detailed human head model (as shown in Fig. 1(a)) to analyze the effect of model complexity on the simulation current with less computation cost but more sufficient accuracy. As the clinical question of this study is the same (current range), thus, human head tissue layers and TMS coil electrodes were constructed from geometric shapes in COMSOL Multiphysics (COMSOL, Ltd., Cambridge, U.K.). The head model consisted of five concentric spheres to represent skin, fatty tissue, muscle, skull, and brain as detailed in Fig 1(b). The human head volume conductor was generated based the anatomical layers' average thickness based on Table 1. The brain layer was designed based on the average human head brain diameter. The coils were represented by the relative diameter and merged with the volume conductor. It is noted that the distance between the coils was equal, and the coils were identical. Also, the coils are both rotationally symmetric and symmetric about the z = 0 planes.

**Table 1.** Anatomical layers properties, r: radius.

| Tissue layer | Conductivity (S/m) | Thickness (mm) |
|---|---|---|
| **Skin** | $2e^{-4}$ | 2.8 |
| **Fatty tissue** | $4.24e^{-2}$ | 2 |
| **Musle** | $3.32e^{-1}$ | 1.7 |
| **Skull** | $2e^{-2}$ | 4.5 |
| **Brain** | $4.75e^{-2}$ | r = 77.5 |
| **Coil** | $5.998e^{7}$ | r = 5 |

### 2.2. Simulation Set-up and Boundary Conditions

A large diameter-based sphere was generated to represent air. A boundary layer was defined around the air layer to mimic the infinite domain as shown in Fig.2(a). These boundaries were insulated during the simulation to obtain accurate results.

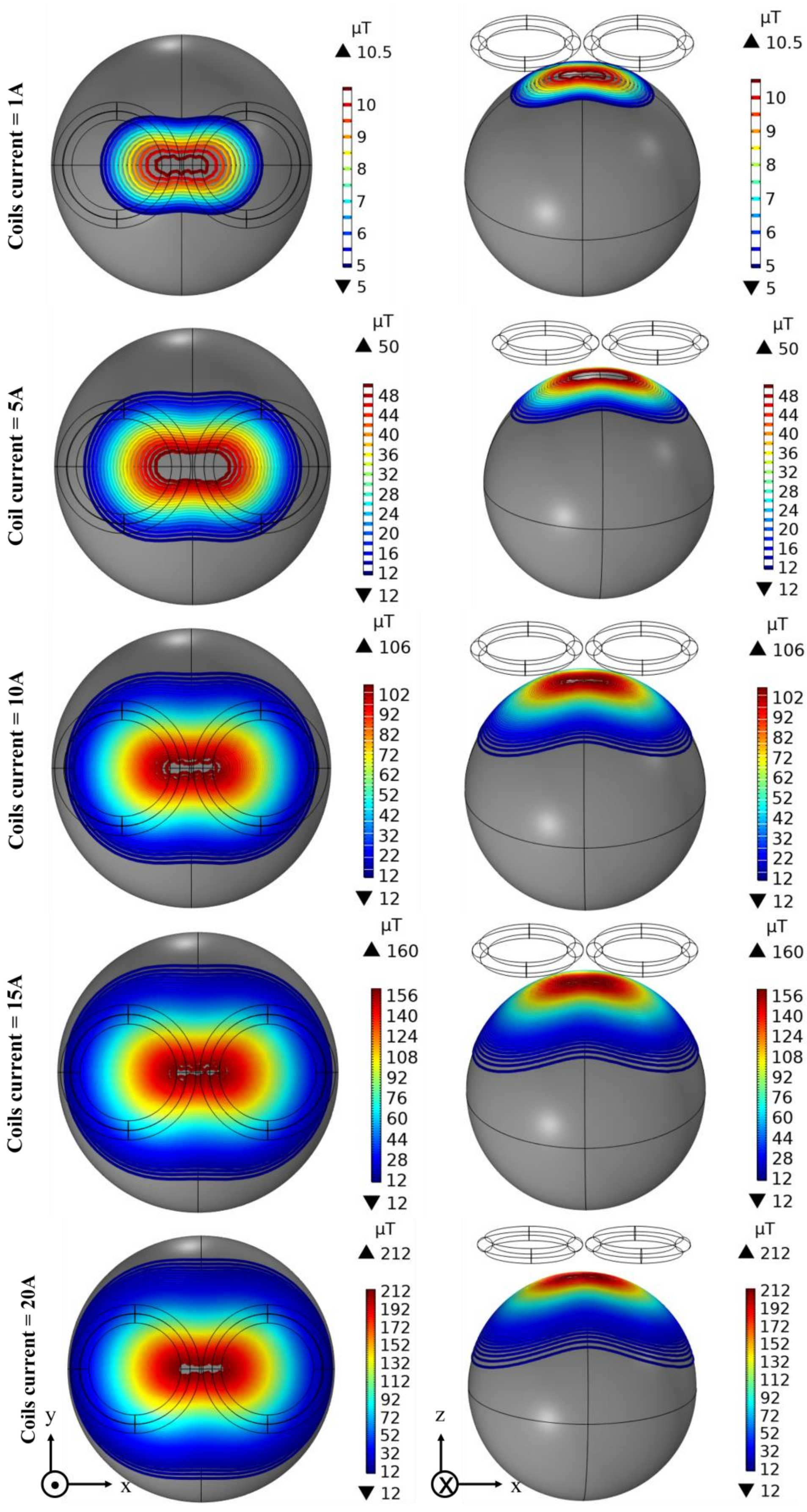


**Figure 3.** Magnetic flux distribution based on various coil current levels. The results are shown on the brain layer.

Since the TMS neuromodulator device works based on the low-frequency range, the dielectric properties of the anatomical layers were attained based on low frequency using Table 1. Since the results were calculated based on the quasi-static approximation, the tissue permittivity was neglected. After generating a volume conductor, the dielectric properties of each tissue layer were attained. Then, various current levels (1 A, 5 A, 10 A, 15 A, 20 A) were defined for coils. The volume conductor simulated each current level by defining the current direction in the coil in COMSOL. The magnetic flux density was calculated for each current level to examine the impact of the current levels on the magnetic flux distribution on the human head. It is noted that the direction of the coil current was defined for each coil to simulate magnetic simulation. The *coil domain* feature was added to the model to specify the direction of the current within the coil. The current flow direction was defined by including a *geometry analysis* sub-feature of the coil domain. The distance between the coils and the human head was set at a constant value to obtain a fair comparison. It is noted that the same coils were used for all current levels.

The magnetic field distributions on the brain layer's edges may provide useful guidance for neuromodulator design. Thus, the magnetic flux density on the edges of the brain layers was analyzed.

To investigate the impact of the coil vertical distance to the skin layer, the coil was placed at the different distance (*4d, 3d, 2d, d*) to the skin layer and the magnetic field distributions over the brain layers was measured. It was noted that the coil current was kept constant to solely analyze the effect of the coil distance to the anatomical layers based on magnetic field distributions.

### 2.3. Finite Element Magnetic Simulation

The FEM was used to solve the electrical magnetic distribution in each medium. Each completed head model was simulated by dividing the geometry between the model into a mesh of small elements and solving the underlying equation for each element separately but in relation to each other in the COMSOL Multiphysics modelling environment. The domains in the volume conductor were discretized using free tetrahedral meshing settings. The region of interest was more finely meshed, while the rest of the region was relatively coarsely meshed to obtain magnetic distributions on head tissue layers at a reasonable time. In particular, the outermost layer was coarsely meshed using *Normal* meshing settings. The remaining layers were meshed using *Finer* meshing setting by applying a minimum element size of 1 mm. This resulted in about 1 million degrees of freedom. All the simulations were carried out using COMSOL while considering the quasi-static approximation of Ampère's Law's equations (1-3) where $H$ is the magnetic field strength, $J$ represents current density, $B$ is magnetic flux density and $A$ shows magnetic vector potential. $\sigma$ shows conductivity and $E$ electrical field. It is required as a pre-process to include *coil geometry analysis* in the study to first compute the current direction through the structure. After successfully solving the model, the *Stationary* study method was applied to the volume conductor to extract magnetic flux density on the anatomical layer. Since the brain layer is the region of interest, the simulation results were recorded on this layer as shown in Fig. 3 and 4.

## 3. Results

The magnetic flux density variation on the brain layer based on various current levels is shown in Fig. 4. The variations are represented with contours.

There is a proportional relation between current levels and induced magnetic flux density. When the current level is increased, the induced magnetic flux density on the anatomical layer shows an identical trend. Also, an increment in the current levels resulted in widespread magnetic flux density over the anatomical layer. It is shown that the maximum magnetic flux density is induced using 20 A while the lowest value is recorded for 1 A. Also, this is valid for the current spreading area on the brain layer. When the distance between the coils and the anatomical layer is increased, the magnetic induction on the brain layer is reduced. Thus, the maximum induction is observed just beneath the coils.

The magnetic flux density on the brain edges is shown in Fig.4. The magnetic flux density is recorded based on arc length in the *xz* and *xz* directions. It is shown a proportional increment with current levels for both directions. The magnetic induction in the *x* direction relatively shows higher variation compared to the *y* direction. The maximum magnetic induction on a certain arc length is about the same for both directions. The magnetic flux density is reduced when it is recorded far away from the coil arrangement.

$$\nabla \times H = J \tag{1}$$

$$B = \nabla \times A \tag{2}$$

$$J = \sigma E \tag{3}$$

The results for the various coil distance to the anatomical layers are shown in Fig. 5. It is shown that when the vertical distance between the coils and the anatomical layers is reduced, the magnetic flux density on the brain layer is inverse-proportionally increased. Although the induced value of the magnetic flux is increased, however, this is not spreaded to the large region of the brain.

The magnetic flux density variation on the target edges based on different directions is shown in Fig. 6. The magnetic flux density is recorded based on arc length in the xz and xz directions. It is shown that when the distance between the coils and skin layer is reduced, the magnetic flux density inversely increases.

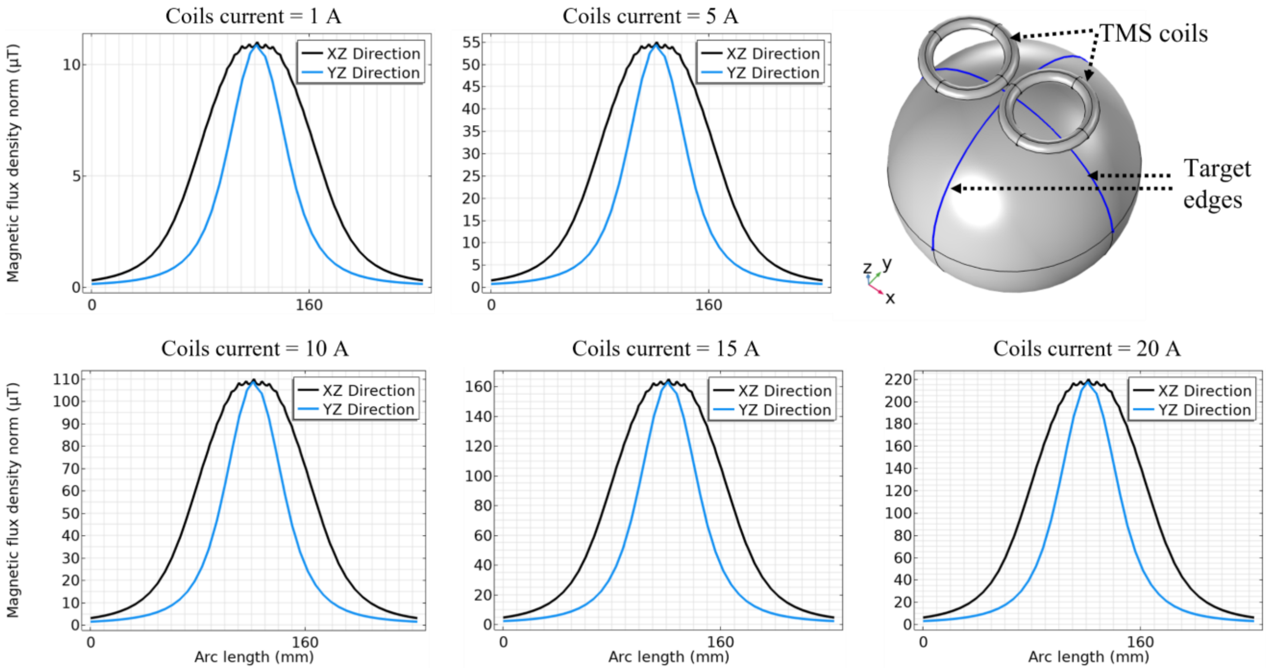


**Figure 4.** Magnetic flux density variation across target edges based on various current levels. Coil current levels are shown and target edges on the brain layer are highlighted. The edge length is highlighted.

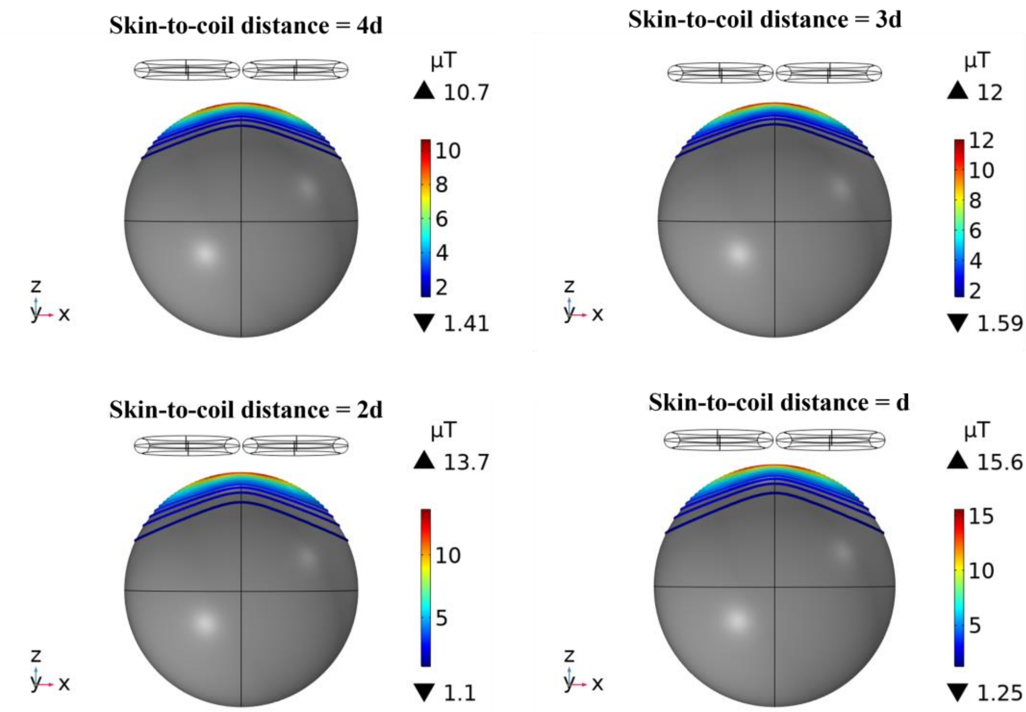


**Figure 5.** Magnetic flux density variation based on various coil vertical distance to the skin layer. The skin-to-coil distance represents with *d*.

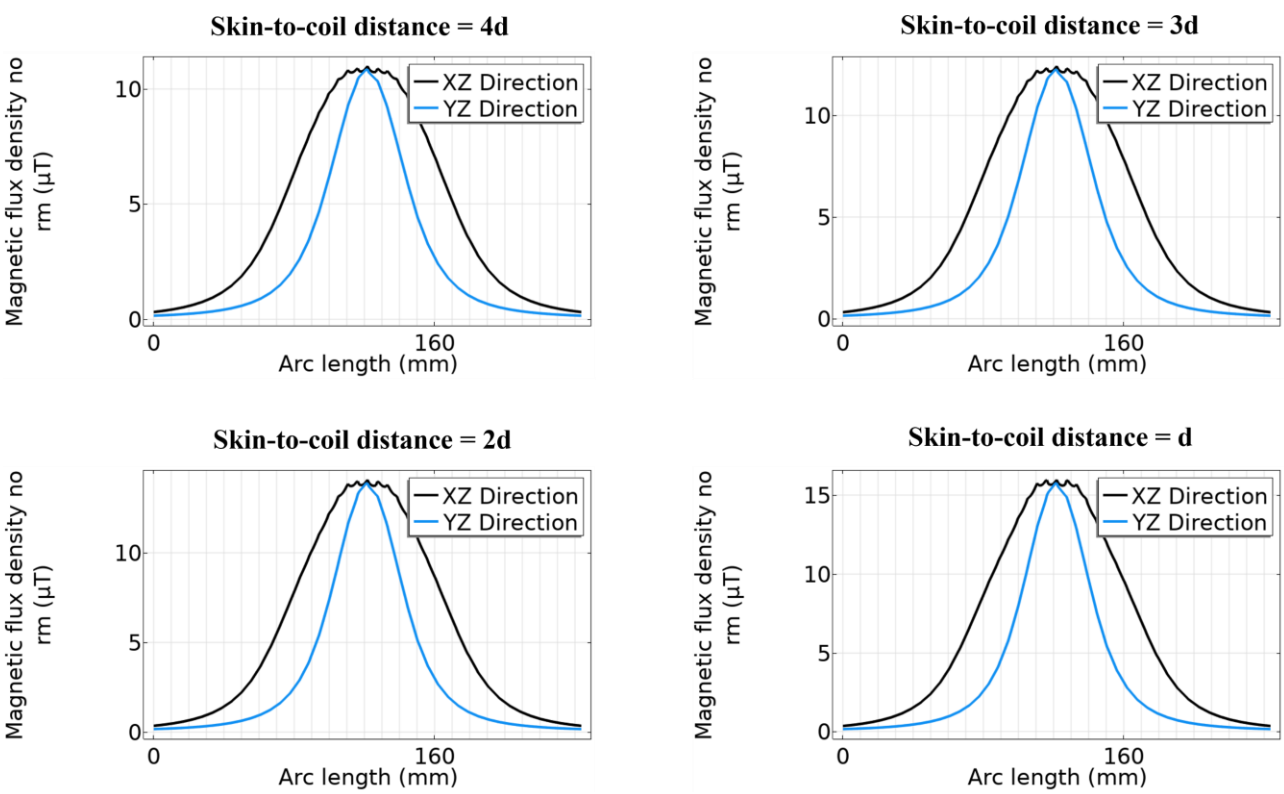


**Figure 6.** Magnetic flux density variation based on various coil vertical distance to the skin layer for target edges. The skin-to-coil distance represents with *d*.

## 4. Discussing

The development of neuromodulator therapy systems for neurological disorders has accelerated with improved technologies and with expanding understanding of the effect of electrical stimulation on neural tissue. Computational methods are widely used for advancing and optimizing electrode design, stimulation parameters, and understanding the mechanism of action of these neuromodulator devices. The optimization methods of the TMS are gradually improved [1], [18]–[20]. The results are mixed results. In particular, it has been shown that relatively higher current levels cause coil heating. The advanced computational modelling facilitates a depth and scale of the investigation that may not be possible in experimental tests. Numerical methods have been used as a tool to study electrical stimulation within the volume conductors. The neuromodulator can be designed and developed using these sophisticated computational methods [21].

In this study, the impact of the TMS coils current levels and their distance on the magnetic induction of the human head layers was investigated using such computational methods. The human head was generated based on concentric smooth shapes to examine the magnetic field distribution on the human brain layer. The various coil current levels were applied and the volume conductor was simulated and results were recorded for each current level. Then, the impact of the coils distance to the skin layer was also investigated.

The results suggested that there was a direct relation between current and induced magnetic flux density. When the current increased ten times, the induced magnetic flux density also approximately increased ten times as shown in Fig. 3 and 4. The same trend was observed for all the applied current levels. As shown in Fig. 4, the induced magnetic flux density was relatively higher in the *x* direction compared to the y direction. This is maybe associated with the coil design as coils occupied more area in the *y* direction.

Theoretically, it has been approved that the magnitude of the induced magnetic field was decayed far away from the coil. The computational study also concluded that the magnetic flux density was significantly reduced when it was recorded far away from the coil arrangements as shown in Fig. 3 and 4. It was shown in Fig. 5 that the spreading of the magnetic flux did not change after a certain distance (e.g., the spreading results are similar for *2d* and *d* distances)

Overall, the results of this study suggested two significant deductions. i) Since the current level is proportional to the induced magnetic flux density, coils should be designed based on higher current levels if the larger region of the brain is targeted. ii) Coils should be designed based on axes based on the region of interest. For example, if the region of interest is in the *x* direction, then the coils should be designed in the *x* direction, iii) It was shown that magnetic flux density was not change after a certain coil distance to the skin layer.

This study investigated a range of current levels and impact of the coils distance to the anatomical layer. It may be required to analyze more current levels to conclude the results. Also, the distance between the volume conductor and coils and the diameter of the coils may have a significant impact on the results. The results of the associated parameters may have a significant impact on the neuromodulator design. Thus, the future of current study is to analyze the impact of these parameters on the optimization of the neuromodulator settings using computational modeling.

## 5. Conclusion

TMS is a non-invasive neuromodulator that has been approved for neuropsychiatric disorders. The therapeutic efficacy of TMS treatment has been modest, despite decades of research. Many potential reasons cause the limitation of these procedures. One prominent example is using relatively higher current levels that may cause discomfort or activate unwanted neuroanatomical structures. Thus, the human head model based on geometric shapes was generated and merged with the conventional coil arrangements to investigate the impact of the current levels on the region of interest. The result showed that the applied current levels and the distance between the simulating coils and the skin layer have a substantial impact on the outcomes.

More accurate results and detailed conclusions may be drawn by modelling the neuromodulator settings including coils size, different simulation frequency range and considering neuroanatomical layer variations. The results can in turn be used to design a more effective neuromodulator based on the specific effects of the variations.

## Declaration of Ethical Standards

The authors of this article declare that the materials and methods used in this study do not require ethical committee permission and/or legal-special permission.

## Conflict of Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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