

**MAGNETIC NERVE STIMULATION:
FIELD FOCALITY AND DEPTH OF PENETRATION**

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MAGNETIC NERVE STIMULATION: FIELD FOCALITY AND DEPTH OF PENETRATION

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Abstract - Magnetic nerve stimulation is a non-invasive method of exciting neural tissue. The major limitation of using magnetic stimulation is the lack of a focused field. At sufficiently high magnetic pulses the diffused field not only stimulates the target population of neurons, but also stimulates adjacent structures as well. Further, for deeply penetrating fields, as is the case in transcranial stimulation, excessively high amplitude current pulses are required in the coils because a significant fraction of the field energy is spread throughout the tissue under the coil.

In this paper we propose two new coil designs that can be used for magnetic stimulation of the peripheral or central nervous system. The purpose of the design was to increase field focality and depth of penetration. The magnetic fields produced by these coils, when driven by biphasic pulses, were simulated using a finite element technique coupled with a transient solver. The resultant field densities and gradients were compared with those obtained from the commonly used Figure-8 coil. Both the air core and the ferromagnetic core designs have superior results when compared to the Figure-8 coil.

Keywords - magnetic nerve stimulation, stimulating coil design, modeling biological tissues.

I. INTRODUCTION

Over the past few years, magnetic stimulation has grown tremendously and moved from diagnoses usage to surgery monitoring, physiotherapy, and psychiatric treatments. The potential advantages of magnetic over electrical stimulation are the key reasons for this growth. Some of these advantages are: reduced or sometimes no pain, access to tissues covered by poorly conductive structures, and stimulation of neural tissues lying deeper in the body without requiring invasive techniques or very high energy pulses [1,2].

The first practical stimulating coil was a circular coil developed by Barker [3]. Hiwacki and Ueno [4] and Maccabee et al. [5] joined pairs of coils to form the Figure-8 coil and the “butterfly” coil respectively. These coils have currents in the same direction at the joint and opposite at the edges. This resulted in a stronger, more focused field under the joint. Larger Figure-8 coils, which could deliver several kilo joules per pulse, were used to stimulate the heart [6,7].

More recently, Knaulein and Weyh [8] decentralized the coil turns to produce an “eccentric coil” in an effort to increase the coil focality. A novel coil, the “slinky” coil, was suggested by Ren et al. [9] and Lin et al. [10]. Their results show that the magnetic field is stronger and more focused when compared to the circular or butterfly coils.

II. THE PROPOSED COILS

The design of the proposed coils is based on modifying the design of the slinky coil of Ren et al. [9]. Both models are composed of three coils, two with windings equally distributed over a 210E arc similar to the slinky model and a third coil tightly wound and positioned perpendicular to the plane of the body surface. The addition of the third set of windings was deduced on the basis of work by previous researchers involving the impact of coil orientation [4,5]. By positioning the third set of windings perpendicular to the joint of the other windings, the field across the targeted nerve will be maximized and weakened outside the targeted area. Accordingly, the combination of the three coils enhances the magnetic field shape and its degree of penetration. The latter can be achieved by independently varying the amount of current supplied to each of the three coils. Further, the introduction of the ferromagnetic core (with relative permeability much higher than that of air) to the second coil design will channel the field outside the body. This results in controlling the magnetic field in the region external to the body and minimizes its diffusion inside the body. Fig. 1 and Fig. 2 show the proposed coils.

Other than the difference in the core material, both proposed coils are similar in shape, size, number of turns, and type of conductors. A detailed description for the coils hardware (coil core material and geometry, coil conductor layout and number of turns) is covered in [11].

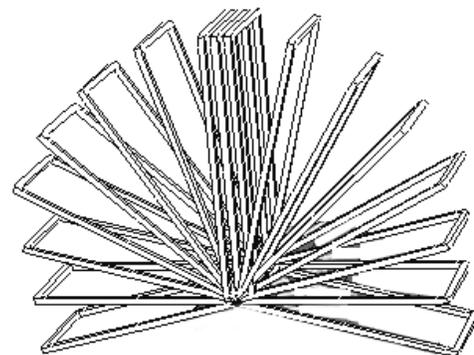


Fig. 1. The proposed air core coil.

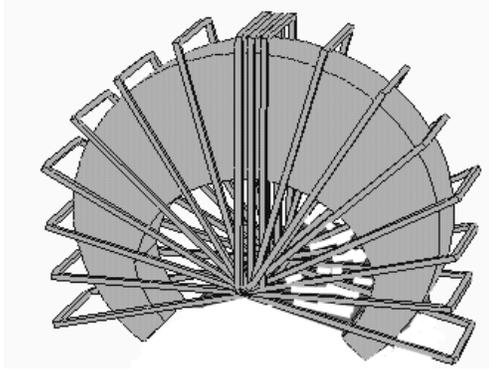


Fig. 2. The proposed magnetic core coil.

III. MODELING AND SIMULATIONS

To model and calculate the electrical field in the tissue due to magnetic stimulation is not an easy task. The difficulties involved are due to the transient state of the supply, the complex shapes of the regions of interest, and the heterogeneous and non linear electrical characteristics of these regions. Considering the above factors, it becomes impossible to solve this problem with precise results by an analytical method. Instead, a numerical method is needed. The finite element method is the most suitable method as it has the flexibility to handle complex geometry for various regions with different types of boundaries and conductivities. The general formula for the electric field induced during magnetic stimulation can be represented by equation (1)

$$\vec{E} = -\nabla V - \dot{\vec{A}} \quad (1)$$

Equation (1) shows that the induced electric field consists of two components. The first component ($-\nabla V$) generates an electrostatic field E_M which is relatively small when compared to the electric field due to magnetic induction E_A . Therefore, to calculate the electric field of the proposed coils, we will only consider the component E_A . To apply the finite element method to this problem, the software package **Magnet** [12] was used. Both proposed coil models were simulated using static and transient field analysis, and the resulting fields were compared to those achieved with a Figure-8 coil (the most commonly used commercial coil).

As stated earlier, magnetic nerve stimulation can be applied to both the peripheral and central nervous systems. The simulations conducted for this study, with the objective of evaluating the proposed coils, used the median nerve as the targeted area. Since the median nerve is well defined within its surrounding tissue, and is the region of interest, a model with distinctive boundaries can be developed. The alternative choice

is a model that represents the cranial region. However, using a cranial model results in a problem that is too difficult to solve or requires tremendous simplifications which leads to unrealistic results. Fig. 3 illustrates the computer model used for our simulations.

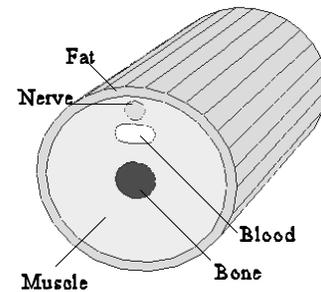


Fig. 3. A model representing a cross section of the upper arm.

The magnetic characteristics (relative permeability) of the heterogeneous tissues included in this model can be considered similar or equal to that of free space. However, the induced potential and the resulting currents will vary depending on the rate of change of the magnetic flux and the different tissue conductivities (after [13]). This model represents a cross section of the upper limb at the junction of the proximal and middle thirds of the humerus. The outside diameter was modeled according to an average adult male's upper limb (10 cm).

When using the software **Magnet**, the first step is to define the problem geometry within a plane of two dimensions. For a three dimensional problem a set of two dimensional planes are required to create a model in three dimensions. The next step is to apply an appropriate mesh to the problem. For this analysis careful mesh distribution was considered with high node densities applied around the nerve and the interface between the coil and the arm. The mesh constructed for this analysis consisted of 101,805 nodes and 175,070 bricks. After applying the mesh, the variables related to the problem were defined including the region electrical characteristics, the coils and their forcing functions (supplies). For the Figure-8 coil, the current was divided equally between both sides of the coil, while for the proposed coils, the current was divided into three equal parts. Two parts were divided between fourteen windings (on both sides of the coil) while the third part was divided equally between the three windings of the third perpendicular coil. The boundary in the XY-plane for this problem was defined as a circle with a radius of 50 cm from the centre of the arm, while the boundary in the Z-axis (depth) was defined 20 cm away from the front and back surface of the arm. Assuming that the magnetic flux diminishes at the boundary, Dirichlet's boundary conditions were applied to this analysis.

IV. RESULTS

Fig. 4 shows the flux density along the Y-axis (the flux starts from the coil interface with the arm and penetrates through the tissues to the center of the bone).

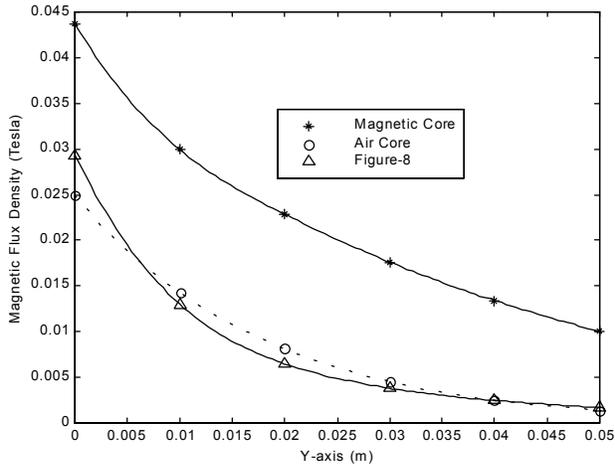


Fig. 4. Flux density along the Y-axis (field penetration depth).

Fig. 5 illustrates the flux density along the perimeter of a cylinder with a radius of 3.75 cm and its center defined at the center of the arm. The radius of this cylinder represents the distance between the center of the arm and that of the nerve.

Fig. 6 shows the flux density along the Z-axis of the previous cylinder.

From Fig. 4, 5, and 6 it can be seen that the flux density produced by the magnetic core coil is higher (almost double) compared to the flux generated by the air core coil or the Figure-8 coil. Further, Fig. 5 and 6 show that the field gradients produced by the magnetic core coil is higher than that generated by the other two coils.

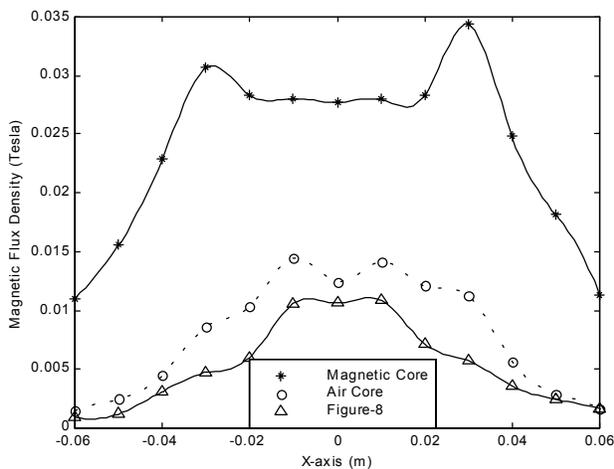


Fig. 5. Flux density along the X-axis (field focality).

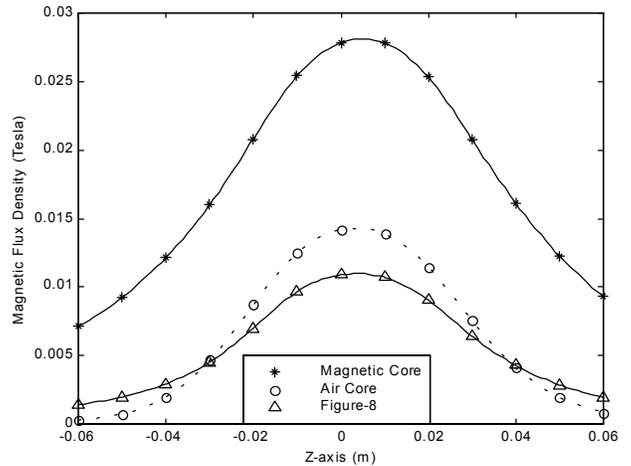


Fig. 6. Flux density along the Z-axis (field focality).

These findings agree with the work done by Lin et al.[10] and Davey et al. [14].

By applying the transient solver, provided by the software **Magnet** [12], the induced currents in the nerve were calculated. The simulations conducted for the three coils used the same energy per pulse with a pulse duration of 300 Fsec. The current waveforms (forcing functions) applied to the three coils were biphasic. Fig. 7 shows the current waveforms (induced in the nerve) produced by the three coils: magnetic core coil, air core coil, and Figure-8 coil.

From Fig. 7 it can be seen that the magnetic core coil generated an induced current five times higher than the air core coil. Further, the air core coil generated an induced current twice as high as the Figure-8 coil. It is crucial to mention that the inductance of the magnetic core coil was assumed to be 25 FH, while the inductances of the air core coil and the Figure-8 coil were considered to be 10 FH.

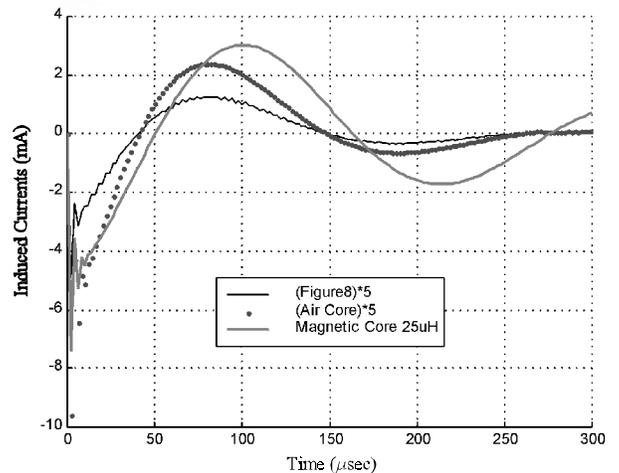


Fig. 7. Induced currents produced by the three coils.

The resultant of the inductance difference can be noticed in Fig. 7. From Fig. 7 it can be seen that the rate of change of the induced current is lower for the magnetic core coil than that of the other two coils. This is a valid result as increasing an inductor will decrease its current rate of change. Further, it can be seen that the induced currents produced by the magnetic core coil carry more multi phasic sections compared to the induced currents produced by the other two coils. This is another valid result as the energy coupled between the arm and the magnetic core coil is higher than the energy coupled by the other two coils.

V. CONCLUSIONS

This paper outlines a new coil design for magnetic nerve stimulation with the objective of enhancing the stimulating coil focality. The finite element method was applied, using the software **Magnet**, to compare the proposed coils with the Figure-8 coil. The simulation results clearly show a large improvement in flux strength and focusing when using the magnetic core coil. The use of the ferromagnetic material enhances the field delivered by the stimulating coil to the targeted area. These findings agree with our previous findings [15,16] and with the results presented by Davey et al. [14] and their suggestion in using a partial toroid as a core for their stimulating coil. However, the unique design proposed in this study for the core geometry adapts and controls the flux lines produced by the coil and consequently improves its performance. Further, these findings agree with Lin et al. [10] regarding the superior performance of slinky-type coils compared to Figure-8 and circular coils. In the proposed coils the introduction of the ferromagnetic material coupled with the inclusion of a tightly wound third set of windings with its own power supply, give greater flexibility for generating a more focussed and deeper field.

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