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Design of a magnetostimulation head coil with rutherford cable winding

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Abstract

Magnetic Particle Imaging (MPI) uses sinusoidal drive fields to excite the magnetic nanoparticles. These time-varying magnetic fields form electric fields within the body, which in turn can cause peripheral nerve stimulation, also known as magnetostimulation. In this work, we propose a design for a human head-size magnetostimulation coil with a Rutherford cable winding. This design achieves 12-fold decrease in the voltages needed to generate a given magnetic field, facilitating the safety of human subject experiments. With electromagnetic simulations, we determine the electric field patterns on a human head model to determine the potential primary locations of magnetostimulation.

I Introduction

Time-varying magnetic fields are subject to human safety constraints on magnetostimulation and specific absorption rate (SAR). In magnetic particle imaging (MPI), studies have shown that the main safety constraint for drive field (DF) frequencies up to 150 kHz is magnetostimulation [1-6]. In addition, stimulation thresholds decrease with frequency [1], and depend on the direction of the applied field [2-4], as well as its duration and duty cycle [5-6].

There has been a growing interest in head-size MPI systems for brain perfusion and functional imaging purposes [7-8]. There is a need to determine the safety limits of these systems before applications on humans. Here, we present the design of a human head-size magnetostimulation coil with Rutherford cable windings. Designed to be used in human-subject magnetostimulation experiments, this setup achieves 12-fold reduction in the voltages needed to generate the required magnetic fields.

II Materials and Methods

II.I Coil Design

First the theoretical threshold for the human head was calculated based on the following model [1]:

$$\Delta B_{\text{min}} = \frac{\lambda_{\text{fit}}}{r_{\text{eff}}} \tag{1}$$

Here, ΔB_{min} is the asymptotic threshold (approximately equal to thresholds at 25 kHz), r_{eff} is the effective radius of the body part, and $\lambda_{\text{fit}} \approx 285$ mT-pp·cm [1]. For the human head, we assume $r_{\text{eff}} = 10$ cm, which yields $\Delta B_{\text{min}} = 28.5$ mT-pp. Note that this is the expected mean threshold for the subjects. To induce stimulation on the

Table 1: Comparison of coil specifications.

Coil Specifications	Coil with Regular Winding	Proposed Coil with Rutherford Cable Winding
Inner Diameter	28 cm	28 cm
Coil Length	25.5 cm	27.2 cm
Number of Layers	2	1
Number of Windings per Layer	84	14
Inductance (L)	6.44 mH	0.044 mH
Coil Sensitivity	528 μT/A	44 μT/A
Voltage at 40 mT-pp	38.2 kV	3.14 kV

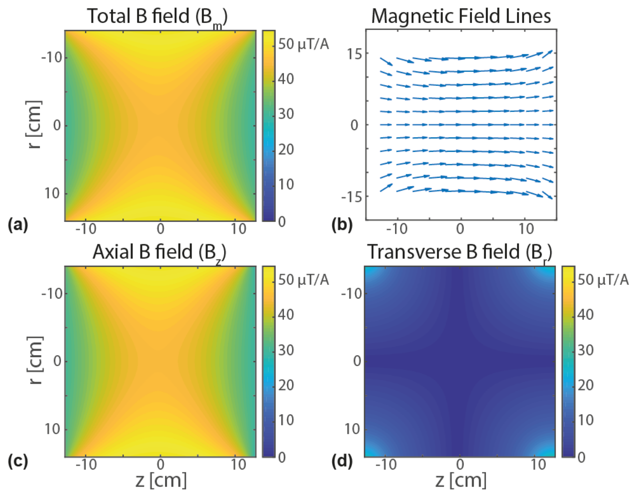

Figure 1: (a) Total B field pattern, (b) B field lines, (c) axial B field pattern, and (d) transverse B field pattern inside the coil. Magnetic field is dominantly along the axial direction.

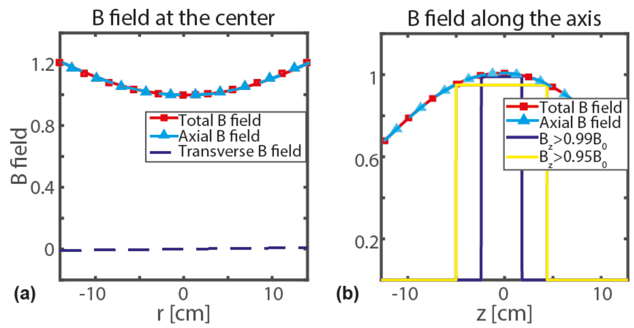

Figure 2: (a) Magnetic field at the center. No transverse magnetic field is observed. (b) Magnetic field along the z-axis and homogeneity levels for different levels.

majority of the recruited subjects, we choose the required magnetic field capability for the coil as $B_{\text{req}} = 40$ mT-pp.

Due to its size, a head coil can have large inductance and voltages. Considering 3 kV/mm electrical breakdown threshold of air [9], the voltage V on the setup need to be constrained to ensure the safety of the subjects:

$$V = I j\omega L = \frac{B_{\text{req}}}{B_0} j\omega L \tag{2}$$

Here, I is current amplitude, B_0 is the coil sensitivity, $L = A_L N_L^2$ is the inductance, A_L is the inductance constant, and N_L is the number of turns. The voltages on a head coil with regular coil winding (i.e., using Litz wires) can easily exceed 30 kV at 25 kHz (see Table 1). One potential solution is to use a thicker wire to reduce N_L. However, such thick wires can be difficult to wind and can also degrade field homogeneity. To reduce the voltage while preserving homogeneity and ease of manufacturing, we propose using a Rutherford cable winding, where N_R Litz wires are twisted together to form a single cable with a rectangular cross-section. With this topology, the number of windings is $N_L' = N_L/N_R$, and the total current for each winding is $I' = I N_R$. Hence, assuming A_L remains approximately the same, N_R^2-fold reduction in inductance and N_R-fold reduction in voltage can be achieved, i.e.,

$$L' \approx A_L N_L'^2 = A_L \left(\frac{N_L}{N_R}\right)^2 = \frac{L}{N_R^2} \tag{3}$$

$$V' = I' j\omega L' \approx I N_R j\omega \frac{L}{N_R^2} = \frac{V}{N_R} \tag{4}$$

II.II Coil Parameters and Simulations

For the head coil designed in this work, we chose $N_R = 12$, where 12 Litz wires were twisted together with 6x2 rectangular cross-section (equivalent to having two layers of winding on a regular coil), to form a single Rutherford cable. Table 1 lists the physical properties of the coils with regular winding and with Rutherford cable winding. The coil with regular winding a total of $N_L = 168$ windings in 2 layers, which was reduced to $N_L' = N_L/N_R = 14$.

MATLAB simulations were performed to determine the magnetic field map and sensitivity for the designed coil. In addition, electromagnetic simulations were performed in COMSOL to determine the electric field patterns on a 3D human head model inserted into the coil model. For these simulations, DF was at 28.5 mT-pp and 25 kHz. The head model, obtained from an online library [10], was assigned uniform dielectric properties, based on the mean properties of brain tissue at 25 kHz: conductivity=0.2 S/m, density=1090 kg/m3, relative permittivity=$1.2 \cdot 10^4$, relative permeability=$2 \cdot 10^4$ [11]. These simulations were performed on an Intel i7 7800X 3.50 GHz CPU, 128 GB RAM computer.

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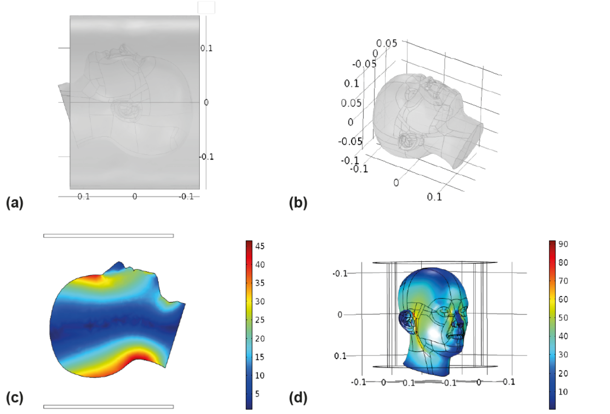

Figure 3: (a) 3D head model placed inside the coil. (b) The dimensions of the model. (c) Electric field map on the central sagittal plane and (d) on the surface of the model (in V/m).

III Results

As given in Table 1, the designed coil with Rutherford cable winding achieves 12-fold reduction in voltage compared to a coil with regular winding. Magnetic field profile for the coil is given in Fig. 1, where the field is dominantly along the axial direction and is homogeneous over a large region of interest. As shown in Fig. 2, the coil has greater than 99 % homogeneity in a 4.2-cm long region, and greater than 95 % homogeneity in a 9.4-cm long region at the center. These features are particularly important for consistent and repeatable measurements of the magnetostimulation thresholds.

Figure 3 shows the electric field map on the 3D human head model. Accordingly, the highest electric field forms on the forehand, behind the head, around the nose, and behind the ear. Hence, magnetostimulation is expected to occur primarily in these regions.

IV Conclusions

In this work, a magnetostimulation head coil is designed for the purposes of determining the thresholds in the human head. The use of Rutherford cable winding achieves 12-fold reduction in voltages compared to regular winding. facilitating the safety of the human subjects, as well as the setup itself.

Author's Statement

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