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Fabrication of a Prototype Magnetic Stimulator Equipped with Eccentric Spiral Coils

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***Abstract*—The development of compact magnetic stimulators will enable us to treat some intractable neurological diseases at one's home. In this study, we propose eccentric spiral coils which induce sufficient eddy currents in the brain at lower driving currents for the stimulator circuit. Numerical simulations based on the finite element method showed the advantages of the proposed design. A prototype coil and driving circuit were fabricated. The coil generated a magnetic field of 1.41 T at the maximum output level of stimulator.**

I. Introduction

REPETITIVE transcranial magnetic stimulation (rTMS) is effective for treatment of several neurological and psychological diseases. Previous studies by our group demonstrated that rTMS to the primary motor cortex relieves neuropathic pain [1,2]. This new method is not invasive, which is different from the conventional method using implanted electrodes. Since rTMS loses its effect in one day after a therapy session, patients have to undergo rTMS every day. An efficient approach may be to install a magnetic stimulator at the patient's home. Conventional stimulator circuits have high output power and advanced setting options to cover a variety of applications, but are consequently large and heavy. Furthermore, only skilled medical doctors can handle conventional stimulators. We are newly developing a compact stimulator system for use at home. In a previous study, we showed a coil navigation system to help patients locate the coil at an appropriate position on their heads [3].

Magnetic stimulators can be substantially downsized if one uses only single-shot stimulations [4,5]. For rTMS, however, a high-power driving circuit is essential, and downsizing the stimulator requires novel technical advances.

The design of magnetic field coils has been an active area of research in magnetic stimulation. Circular coils have been used ever since magnetic stimulation was invented [6]. Figure-eight coils induce strong eddy currents in a focused region in the brain [7]. In addition to these principal coils, there has been a number of studies on coil design to obtain focused or deeply penetrating eddy currents [8-12].

In this study, we propose a novel coil design which induces sufficient currents in the brain at lower driving currents for the stimulator. Prototype coil and driving circuit were fabricated.

II. Coil Design

Fig. 1 illustrates the winding geometries of a conventional figure-eight coil and our eccentric spiral coils. In the conventional coil, the center of the outer circumference coincides with the center of the inner circumference. In the proposed coil, the inner circumferences are shifted closer together.

When the brain is stimulated using a figure-eight coil, the eddy currents in the brain converge below the middle point of the two spirals. The proposed design has dense conductors in the middle point, which results in a higher eddy current density below these conductors. This means that the proposed coil requires smaller driving currents compared with the conventional coil to obtain the same eddy current density in the target area.

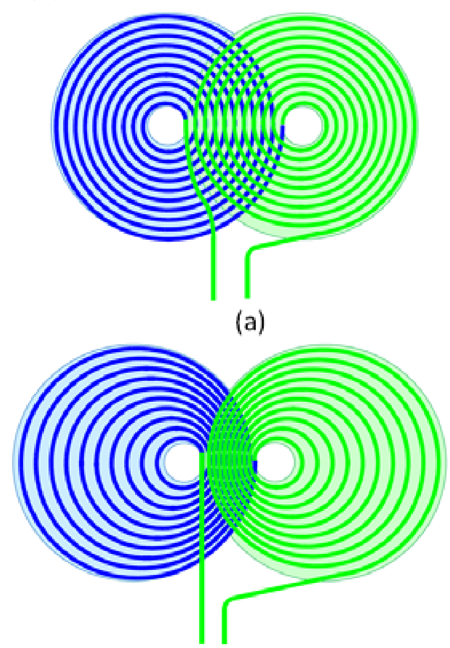

Fig. 1. (a) Conventional figure-eight coil and (b) eccentric spiral coils.

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The driving circuit accounts for most of the total weight of a magnetic stimulator system. The proposed design enables us to use a driving circuit with lower output power and smaller size.

The two spirals in Fig. 1 partly overlap with each other. The overlap further increases the maximum eddy current density below the middle of the coil.

The above discussion is based on the assumption that the coil inner diameter is substantially smaller than the outer diameter. We found in preliminary investigations that the reduction in inner diameter is effective for decreasing the coil inductance and thus decreasing the required voltage for charging the capacitor in the driving circuit.

III. Numerical Simulations of Eddy Currents

The advantage of the proposed coil design was evaluated using numerical simulations of eddy currents. Fig. 2 shows numerical models of the stimulating coils and the human brain. The human brain model was constructed as a fourth part of a sphere with some modifications to reduce the amount of calculation. The brain model had a uniform electric conductivity of 0.106 S/m, which corresponds to the conductivity of the gray matter at 3 kHz. The brain model was surrounded by an insulating layer for incorporating the effect of the skull, which is not shown in Fig. 2. The number of elements in the brain model and the insulating layer was 34320.

The coil consisted of two spirals; 10 turns each, an outer diameter of 100 mm, and an inner diameter of 20 mm. The conductors had a rectangular section of 6 mm × 2 mm.

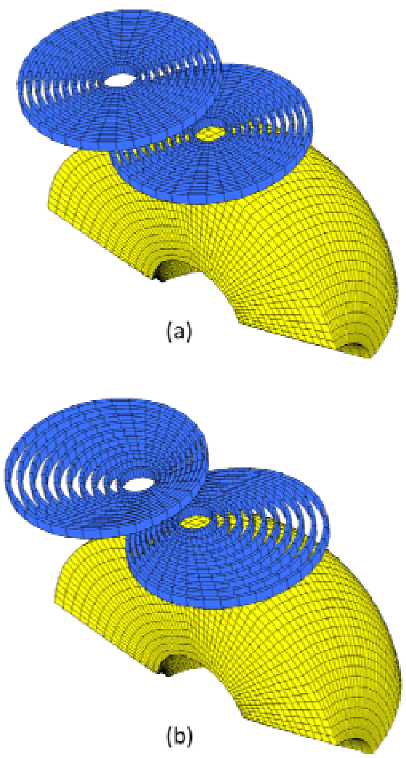

Fig. 2. Numerical models of coils and the human brain for electromagnetic field analyses. (a) Conventional figure-eight coil and (b) eccentric spiral coils.

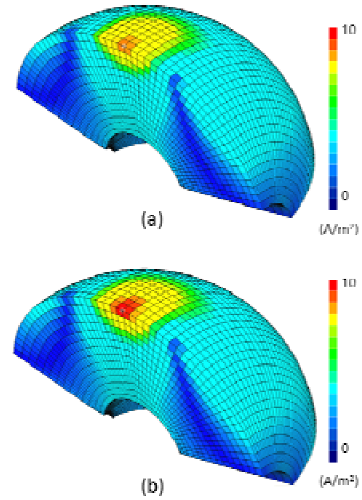

Fig. 3. Eddy current distributions on the surface of the brain model for (a) the conventional figure-eight coil and (b) the eccentric spiral coils.

While the actual coils have spiral windings, we constructed the coil models as a set of ring conductors for simplifying the procedure. In the eccentric spiral design, we gave maximum shift to the rings so that they touched each other in the middle of the coil. This does not mean that the rings were electrically connected with each other.

AC currents were applied to each ring conductor with an intensity of 4860 A and a frequency of 3.3 kHz, which corresponds to the inverse of a pulse width of 300 μs.

Analyses were carried out using commercially available software, PHOTO-EDDYjω. The vector potential of magnetic field produced from the coil was obtained by numerically integrating the Biot-Savart law. The eddy current distribution in the brain model was obtained using the finite element method.

Fig. 3 shows the eddy current densities for the conventional and proposed coils. The current density exhibited the maximum value below the middle of the two coils. The maximum current densities of the conventional and proposed coils were 8.53 A/m^2 and 9.27 A/m^2, respectively. The proposed design caused an increase of eddy current density. On the other hand, the maximum current density was 8.1kA when winding turns were pushed to the outer circumference. According to our analyses of magnetic fields, the inductances of the conventional and proposed coil were 8.88 μH and 9.80 μH, respectively. This would cause an increase in the required capacitor voltage by 5 % to obtain the same coil current.

These results indicate that the proposed coil requires an intensity of driving current approximately 10 % smaller than the conventional coil for obtaining a comparable eddy current density in the brain, thus contributing to downsizing the stimulator system. Because heat generation in the coil is proportional to the square of the driving current, the contribution to a reduction in heat should be larger than 10 %.

IV. FABRICATION OF A PROTOTYPE STIMULATOR

A magnetic field coil was fabricated based on the proposed eccentric spiral design, as shown in Fig. 4. The design parameters of the windings were the same as those used in the numerical simulations except for the degree of decentering. The pathways of the windings are expressed using the following functions:

$$
\begin{aligned}
x &= -\left(10 + \frac{40}{2\pi \times 10}\theta\right)\cos\theta \\
y &= \left(10 + \frac{40}{2\pi \times 10}\theta\right)\sin\theta
\end{aligned}
\qquad (1)
$$

where θ ranged from 0 to 20π, and the units of x and y were in mm. The total length of the windings was 3.7 m. The minimum gap between the adjacent conductors was 0.5 mm in the middle of the two windings. Within the overlapped region, the gap between the two winding layers was 1 mm.

Due to the complicated geometry, it is not easy to wind up the coil from a single rectangular-section wire. We cut the windings out of copper plates 6 mm in thickness. The use of a wire electrical discharge machine enabled us to cut out the windings without considerable distortion. This machine was applicable only to planar geometries. Windings with an overlap should be divided into multiple parts for machining. The conductors shown in Fig. 4(a) consist of four parts, including the two spirals, resulting in three joints between the parts (one joint is not visible in the photo). Plastic chips were inserted for electric insulation where the winding conductors came close to each other in the middle of the coil.

The conductors were located inside a plastic casing. The casing was filled with epoxy for mechanical support and electric insulation. The outer size was 200 mm × 120 mm × 21 mm. The thickness of casing was 1 mm on the side attached to the human head. A cable of 1 m in length was connected to the terminals of the coil.

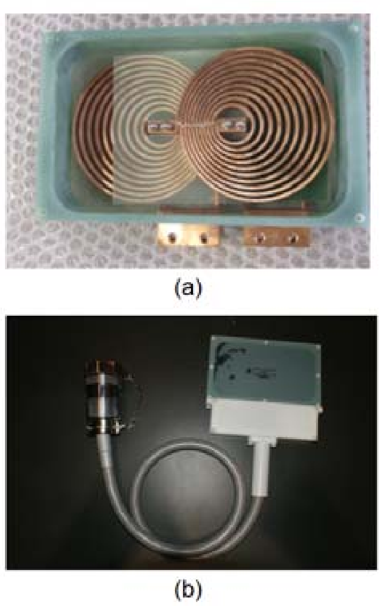

Fig. 4. A prototype coil based on the eccentric spiral design. (a) Windings and (b) completed coil with a cable.

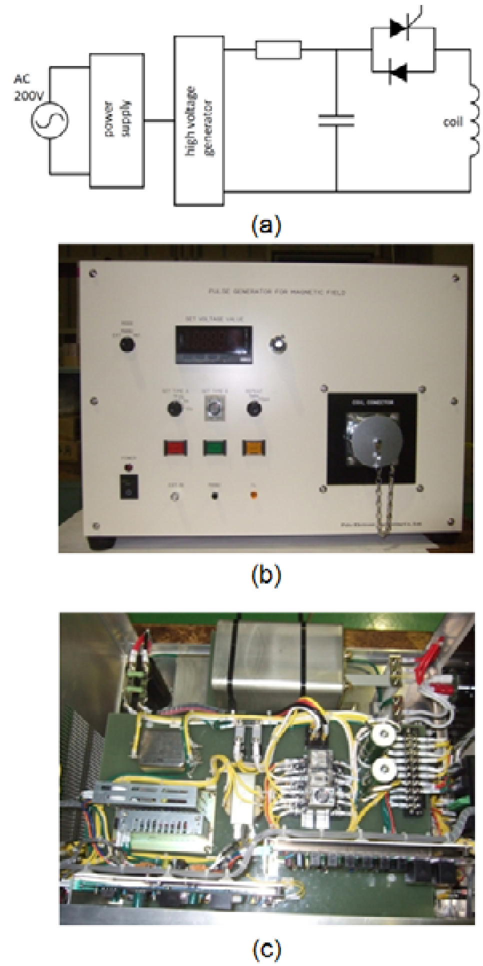

Fig. 5. Originally developed driving circuit for rTMS. (a) Simplified circuit diagram, (b) exterior and (c) internal structure of driving circuit.

As illustrated in Fig. 5(a), the basic configuration of the driving circuit is composed of a power supply generating DC voltages from AC of 200 V, a high-voltage generator for charging the capacitor, a resistor for limiting the charging current, and a semiconductor switch for controlling the coil current. The combination of a thyristor and a diode allows the coil current to flow with a biphasic waveform.

Figures 5(b) and (c) show the exterior and the internal structure of our originally fabricated driving circuit. The front panel has a connector for the coil cable, a power switch, buttons for controlling pulse generation, a dial and an indicator for adjusting the voltage. The total weight of the driving circuit is 41 kg. There is an input connector on the rear panel for externally triggering the pulse generation. The above driving circuit basically follows conventional designs. However, the development of driving circuit in parallel with the coil is important for downsizing the total system. We are reconsidering the specification of each element in the driving circuit.

The intensity of stimulation is adjusted so that the risk of inducing seizure is sufficiently low while providing substantial therapeutic effects. The TMS is equally safe as the electric stimulation which has already been carried out at home.

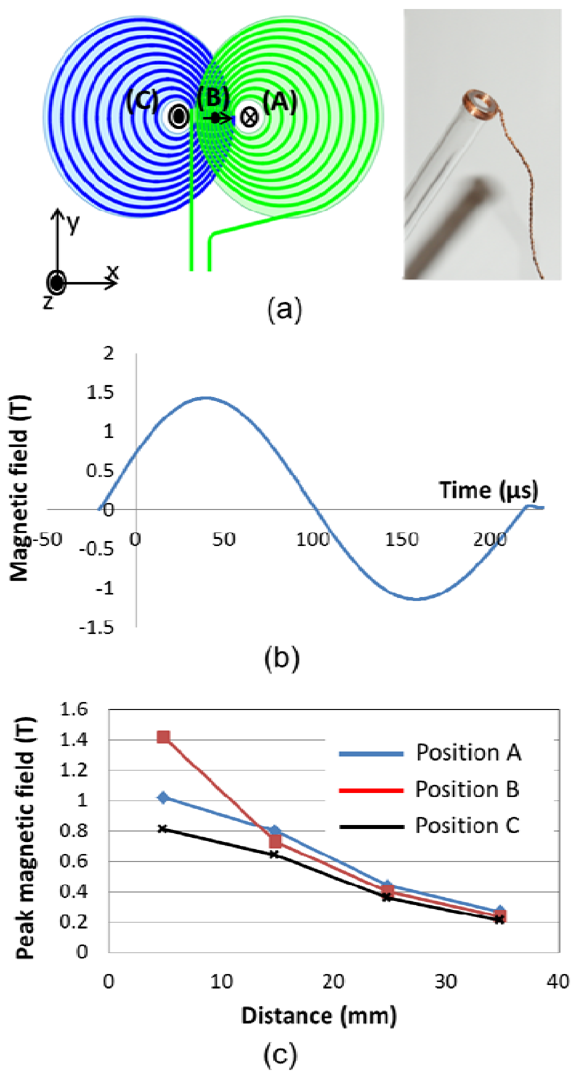

Fig.6. Measurements of magnetic fields generated from the prototype coil. (a) Measuring positions and search coil. (b) Measured waveform of magnetic field. (c) Peak magnetic field

V. Measurements of Magnetic Fields

Magnetic fields were measured using a search coil located above the stimulating coil. Time-varying magnetic fields arising from the stimulating coil induce voltages in the search coil. The voltage waveforms were recorded using an oscilloscope. The driving circuit gave the maximum output. The search coil consisted of five turns of inner loops 6.3 mm in diameter, and four turns of outer loops 7.1 mm in diameter. These parameters resulted in an effective area of 314.3 mm^2. As shown in Fig. 6(a), the search coil was located in the following three positions:

Position A: Center of the right winding
Position B: Middle of the two windings
Position C: Center of the left winding

For each position, the gap between the center of the search coil and of the upper stimulating coil (green part in Fig.6(a)) ranged between 4.8 mm and 34.8mm in 10-mm steps. The directions of measured field component are shown in Fig.6 (a). The stimulating coil lies in the x-y plane. The z component of magnetic field was measured for positions A and C, whereas the x component was measured for position B. The magnetic flux density $B(t)$ was obtained by numerically integrating the voltage of the search coil:

$$B(t) = \frac{1}{S}\int_{t_0}^{t} V(t')dt' \qquad (2)$$

where S is the effective area of the search coil. As shown in Fig. 6(b), the peak magnetic flux density was 1.41 T at position B, gap 4.8 mm. The measured pulse width was 240 µs.

Fig. 6(c) shows the dependence of the peak magnetic flux density on the distance of the center of the search coil from the center of the upper stimulating coil. The magnetic flux density attenuated with an increase in distance. The result of position C was similar in measured values to that of position A. At position B, the peak flux density was 1.41 T at a distance of 4.8 mm. The intensities of the magnetic fields were comparable to those of conventional magnetic stimulators.

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