Liu, Mengfei; Tuovinen, Pauli H.; Kawasaki, Yuta; Yedeas, Mohamed Amine; Saitoh, Youichi; Sekino, Masaki

Electromagnetic and mechanical characterization of a flexible coil for transcranial magnetic stimulation

Published in:
AIP ADVANCES

DOI:
10.1063/1.5080148

Published: 01/03/2019

Please cite the original version:
Electromagnetic and mechanical characterization of a flexible coil for transcranial magnetic stimulation

Mengfei Liu, Pauli H. Tuovinen, Yuta Kawasaki, Mohamed Amine Yedeas, Youichi Saitoh, and Masaki Sekino
Electromagnetic and mechanical characterization of a flexible coil for transcranial magnetic stimulation

Mengfei Liu,1 Pauli H. Tuovinen,1,2 Yuta Kawasaki,1 Mohamed Amine Yedeas,1 Youichi Saitoh,3 and Masaki Sekino1,a)

AFFILIATIONS
1Graduate School of Engineering, The University of Tokyo, Tokyo 113-8656, Japan
2Department of Neuroscience and Biomedical Engineering, Aalto University, Espoo 00076, Finland
3Graduate School of Medicine, Osaka University, Osaka 565-0871, Japan

Note: This paper was presented at the 2019 Joint MMM-Intermag Conference.
a)Masaki Sekino: sekino@bee.t.u-tokyo.ac.jp

ABSTRACT
Transcranial magnetic stimulation is a painless and noninvasive method for treating brain disorders. The coil-geometries that match the head topologies achieve more effective stimulation, however it is difficult for a rigid coil to fit the skulls of all patients. We propose and develop a rubber-like flexible coil that can be shaped into different geometries to reduce the inter-individual variabilities in its clinical use. The main challenge is attributed to the fact that the external bending and induced Lorentzian forces cause coil deformation and fatigue. Herein, we investigated the influence of bending on the electromagnetic characteristics of the flexible coil. The magnetic field distribution was calculated and measured using a search coil. The induced Lorentzian force was calculated and the induced eddy current density was simulated using the scalar potential finite difference (SPFD) method. For a mechanical characterization, we fixed the center of the coil, and external bending forces were applied on the two wings of the coil, while Lorentzian forces were applied in a direction normal to the side wall of the wire groove. Fatigue analyses of these forces were also conducted. The results show that the eddy current density induced in the brain by the flexible coil was significantly higher compared to that of the figure-eight and butterfly coils. Fatigue analyses show that the bending force required to achieve a close coil fit on the human head and the generated Lorentzian force would not lead to fatigue problem.

© 2019 Author(s). All article content, except where otherwise noted, is licensed under a Creative Commons Attribution (CC BY) license (http://creativecommons.org/licenses/by/4.0/). https://doi.org/10.1063/1.5080148

I. INTRODUCTION

Transcranial magnetic stimulation (TMS) is a painless and noninvasive technique that uses a coil to stimulate neurons in the brain based on electromagnetic induction.1 It has generated significant interest as a method for treatments of neurological and psychiatric disorders.

TMS stimulator coils that match the target anatomical curvature exhibit the highest filling factor of the coil, indicating that the coil must be positioned as close to the target as possible. However, complicated inter-individual variabilities exist in clinical use as head topologies vary greatly among patients. Additionally, coil positions differ according to the target regions on the brain, leading to considerable differences in the target curvatures. Therefore, our group proposed the concept of a rubber-like flexible coil that can be shaped into different geometries according to the target curvature to reduce inter-individual variability in its clinical use and enhance its performance compared to traditional, rigid coils.

One of the main challenges associated with the use of flexible coils is that in time, repetitive external bending forces may cause plastic deformation and fatigue problems. Furthermore, the induced Lorentzian force within the flexible coil may lead to more severe problems upon its deformation compared to traditional, rigid coils.2

In this study, we investigated the influence of coil bending on its electromagnetic characteristics as well as on the lifetime of a flexible coil exposed to repetitive loading and unloading subject to external bending forces and to an induced Lorentzian force.
II. CONCEPT AND DESIGN

The widely used rigid figure-eight and butterfly coils share the same basic structure but have different angles between the two wings. The figure-eight coil has good performance in terms of focality but has poor penetration characteristics of the induced eddy currents.\(^5\) The butterfly coil has an improved penetration depth because it is bended by a larger angle at the expense of poor focality characteristics.\(^6\) In the butterfly coil, however, the local gap between the coil and the head surface still causes magnetic field attenuation.

Therefore, we propose a coil design made of a soft material that it is flexible to conform tightly to the surface of the head (Fig. 1(a)). Thus, according to this flexible configuration, we expect that the proposed coil induces eddy currents in the brain more efficiently compared to the figure-eight and the butterfly coils. The conductor winding (Fig. 1(b)) was designed based on the structure of the figure-eight coil and the butterfly coil, and the insulator case was constructed using UV-cured, acrylic, rubber-like resin (Shorehardness Scale A 40–50). It allows bending of 90\(^\circ\) in the upward and downward directions and was assumed to have the ability to be fitted according to the target curvature in all cases. The conductor was made of Kapton insulated (breakdown voltage of 7 kV), tinned, braided copper wire (resistivity of 10.44 \(\Omega/\text{km}\)). The coil parameters were set to achieve a good focality–depth tradeoff\(^5\) with 10 turns on each side and a 2 mm gap between the two conductors. The length and width of the coil were 198 mm and 102 mm, respectively.

Measurement of induced magnetic flux was carried out based on the schematic diagram (Fig. 1(c)). A current monitor was used to measure the generated current running through the stimulator coil. Induced magnetic flux in the center of the coil was measured with a search coil (15 turns and a radius of 3 mm) based on Faraday’s law of induction. The search coil was placed on 20 mm to 60 mm above the center of the coil. An oscilloscope was used to record the signal from both current monitor and search coil. Experiments were carried out both on flexible coil and a 45\(^\circ\) butterfly coil.

In addition, coil heating is a major technical challenge in TMS,\(^6\) especially for the soft cover material. Thus, the cover material was chosen based on considerations of its thermal plasticization characteristics up to the temperature of 48\(^\circ\)C at which the polymer loses the ability to keep its original shape.

III. EXPERIMENTS

A. Induced magnetic field mapping

The variable magnetic flux induced by the flowing biphasic pulsed current could be calculated based on Jefimenko’s equations:\(^10\)

\[
B(r,t) = \frac{\mu_0}{4\pi} \int \frac{\int f\left(r', t_0\right) \frac{1}{|r - r'|^2} \frac{\partial}{\partial t} f\left(r', t_0\right)}{|r - r'|^3} c \times (r - r') d^3r'
\]

where \(B\) represents induced magnetic flux, \(\mu_0\) is the magnetic constant, \(J\) is the total current density, \(r'\) is a spatial point within the charge distribution, \(r\) is a point in space, and \(t_0 = t - \frac{|r - r'|}{c}\) is the retarded time.
B. Magnetic field measurement

The induced magnetic flux in the center of the coil can be measured with a search coil (15 turns and a radius of 3 mm) based on Faraday’s law of induction (Fig. 1(c)).

\[ \varepsilon = -N \frac{d\phi_B}{dt} = -NS \frac{dB}{dt} \]  

(2)
where $\varepsilon$ is the electromotive force, $N$ is the number of turns of the search coil, $S$ is the cross-sectional area of the search coil, and $\frac{dB}{dt}$ is the rate of change of the magnetic flux $B$.

Therefore, the magnetic flux can be calculated with the following equation,

$$B = \frac{1}{NS} \int \varepsilon dt \quad (3)$$

### C. Lorentzian force analysis

The induced Lorentzian force of an element in the coil can be simplified according to the Laplace force,

$$df = l dl \times B = Bldl \quad (4)$$

where $f$ is the Lorentzian force, and $l$ is the current flowing through the coil.

### D. Induced eddy current density simulation

The coil efficiencies were assessed by comparing the flexible, butterfly, and figure-eight coils, after they were placed on the human head with the use of the scalar potential finite difference (SPFD) method. The focality—defined as the area at which the current density is $\geq$ half the maximum value ($J_{\text{max}}$)—was calculated together with the average current densities ($J_{\text{ave}}$) within a spherical area with a radius of 10 mm at a distance of 20 mm below the brain.

The homogenous half spherical head model (radius of 75 mm) was used with a 0.1 S/m conductivity. Simulations on developed personalized brain models of healthy subjects were carried out to analyze whether there were inter-individual variabilities on the coil efficiency. These simulations targeted the primary motor cortex areas for face, foot, and hand. A biphasic pulse current with an intensity of 3 kA was applied at a frequency of 4 kHz, which was equal to the current generated by the driving circuit used in this experiment.

### E. Fatigue analysis

Repetitive applications of external bending and induced Lorentzian forces in the coil may cause its plastic deformation. Over time, force loading and unloading may even lead to the fracturing of the coil. Therefore, fatigue analysis is essential for testing the safety and lifetime of the rubber-like flexible coil. We used the ANSYS workbench to conduct fatigue analyses. The fatigue lifetime of a material can be reflected by the characteristic curve of stress vs. number of cycles ($S$–$N$). If the cycle number achieved is in the order of $10^5$–$10^6$, the object can be regarded to have an infinite lifetime in terms of fatigue characteristics and would not experience fatigue problems when used in practice under the same conditions.

We suppose a treatment of neuropathic pain using TMS. A previous study showed that a TMS session of 500 pulses/day was effective. A total of $5 \times 10^5$ cycles is equivalent to 3 years of use.

Polypropylene was used as the test material, and simulations were conducted on the external bending force ranging from 1–10 kN. Loading was applied on the cover and the applied pressure by the induced Lorentzian force (normal to the side wall of the wire groove) ranged from 0.1–10 MPa. The frequency of the loading bending force was 0.5 Hz, while the Lorentzian force was 5 Hz according to the frequency of the repetitive TMS (rTMS) trial.

![FIG. 3 SPFD simulation results for the butterfly, figure-eight, and flexible coils.](image)
IV. RESULTS AND DISCUSSION

A. Induced magnetic field mapping

According to the results (Fig. 2(a–d)), a maximum magnetic field was induced in the center of each circular coil. In the center of each circular coil, the magnetic field induced by the flexible coil yielded increased field intensities along the x- and y-directions but lower intensities along the z-direction compared to the figure-eight coil. As the bending angle increased, the total magnetic field decreased. This means that the bending of the flexible coil leads to a decrease in the maximum induced magnetic field. However, given that the coil matched the head topologies tightly (unlike the large gap between the figure-eight coil and head surface), the filling factor of the magnetic stimulator coil was improved, and the efficiency of eddy current induction was improved. The increased utilization rate of the magnetic field energy helped the flexible coil induce a stronger eddy current inside the brain.

B. Magnetic field measurements

The results of the induced magnetic flux measured below the center of the entire structure of the flexible coil prototype and the 45° butterfly coil showed that the flexible coil induced a stronger

<table>
<thead>
<tr>
<th>Coil</th>
<th>$J_{ave}$ (A/m$^2$)</th>
<th>$J_{max/2}$ area (mm$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Butterfly</td>
<td>11.22</td>
<td>3826</td>
</tr>
<tr>
<td>Figure-eight</td>
<td>8.425</td>
<td>2287</td>
</tr>
<tr>
<td>Flexible</td>
<td>12.56</td>
<td>2944</td>
</tr>
</tbody>
</table>

TABLE I. Numerical simulation results using SPFD.

<table>
<thead>
<tr>
<th>$J_{ave}$ (A/m$^2$)</th>
<th>E</th>
<th>K</th>
<th>Q</th>
<th>S</th>
</tr>
</thead>
<tbody>
<tr>
<td>Butterfly</td>
<td>23.23</td>
<td>7.035</td>
<td>23.47</td>
<td>5.491</td>
</tr>
<tr>
<td>Figure-eight</td>
<td>10.89</td>
<td>11.10</td>
<td>11.07</td>
<td>7.508</td>
</tr>
<tr>
<td>Flexible</td>
<td>26.56</td>
<td>15.44</td>
<td>11.91</td>
<td>10.91</td>
</tr>
</tbody>
</table>

TABLE II. Average current density on the LT hand area.

<table>
<thead>
<tr>
<th>$J_{ave}$ (A/m$^2$)</th>
<th>E</th>
<th>K</th>
<th>Q</th>
<th>S</th>
</tr>
</thead>
<tbody>
<tr>
<td>Butterfly</td>
<td>12.86</td>
<td>6.767</td>
<td>10.48</td>
<td>7.872</td>
</tr>
<tr>
<td>Figure-eight</td>
<td>10.15</td>
<td>10.87</td>
<td>9.209</td>
<td>8.630</td>
</tr>
<tr>
<td>Flexible</td>
<td>20.49</td>
<td>15.62</td>
<td>16.51</td>
<td>14.18</td>
</tr>
</tbody>
</table>

TABLE III. Average current density on the LT face area.

<table>
<thead>
<tr>
<th>$J_{ave}$ (A/m$^2$)</th>
<th>E</th>
<th>K</th>
<th>Q</th>
<th>S</th>
</tr>
</thead>
<tbody>
<tr>
<td>Butterfly</td>
<td>3.175</td>
<td>3.552</td>
<td>5.117</td>
<td>4.505</td>
</tr>
<tr>
<td>Figure-eight</td>
<td>5.223</td>
<td>5.265</td>
<td>5.152</td>
<td>4.344</td>
</tr>
<tr>
<td>Flexible</td>
<td>7.333</td>
<td>7.508</td>
<td>9.021</td>
<td>7.794</td>
</tr>
</tbody>
</table>

TABLE IV. Average current density on the LT foot area.
magnetic field compared to the 45° butterfly coil at the same distance from the center (red dot in Fig. 2(e), which is regarded as the center of head surface). As the bending angle increased, the magnetic field below the center of the coil increased.

Combined with the calculated results, it was shown that bending led to decreased maximum magnetic field intensities in the center of each circular coil. Accordingly, this effect may reduce the induced Lorentzian forces—a finding that has direct relevance to the coil’s safety profile. Additionally, bending led to an increase of the magnetic field induced in the middle of the two windings, and helped increase the eddy current density induced inside the brain.

C. Lorentzian force analysis

The maximum induced Lorentzian force also decreased as the bending angle increased (Fig. 3(g)). Therefore, the Lorentzian force induced by the flexible coil was smaller than that induced by the figure-eight coil (17 N).

D. Induced eddy current density simulation

1. Simulation on half-sphere head model

According to the results of simulations on the half-spherical head model (Fig. 3(a–f) and Table I), the flexible coil induced the highest current density among the coils (10.67% higher than the butterfly coil and 49.08% higher than the figure-eight coil). The distribution area was 23.05% smaller than the butterfly coil and 28.72% larger than the figure-eight coil.

2. Simulation on realistic head model

Coils were fitted on the brains (magnetic resonance imaging data) of different subjects (E, K, Q, S), as well as different stimulation target regions (left hand, left foot, and left face area of the motor cortex) (Fig. 3(g–h)).

According to Tables II–IV, in most cases, the flexible coil induced the strongest eddy current density among the three types of coils. The flexible coil induced much larger eddy current densities than those induced by the figure-eight and butterfly coils in different subjects in different target regions.

E. Fatigue analysis

First, analyses were conducted on the bending force on the cover structure (Fig. 4(a, b)) and the Lorentzian force induced on the bottom part (Fig. 4(c, d)). According to the results, repetitive forces less than 2250 N on the cover structure would not lead to a fatigue problem over time. Additionally, generated pressure by the Lorentzian force less than 10 MPa would not lead to a fatigue problem. Since the maximum Lorentzian force was calculated to be 17 N, the pressure was approximately 0.1 MPa. Since the maximum Lorentzian force was calculated to be 17 N, the pressure was approximately 0.1 MPa.

Simulations on the entire body of the flexible coil were then performed with the combination of the external bending force and the pressure generated by the Lorentzian force. The results (Fig. 4(e, f)) show that (1) the assembled structure of the flexible coil was much more stable and less fragile than its individual components, and (2) an external bending forces less than 10000 N under a Lorentzian pressure of 0.1 MPa, would not lead to fatigue problems.

Therefore, bending force of less than 1000 N exerted by humans and pressures induced by the Lorentzian force of approximately 0.1 MPa can be considered durable for daily use.

ACKNOWLEDGMENTS

This study was supported by Teijin Pharma Limited.

REFERENCES