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NOTE

Determination of optimal electrode positions for transcranial direct current stimulation (tDCS)

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Abstract
The present study introduces a new approach to determining optimal electrode positions in transcranial direct current stimulation (tDCS). Electric field and 3D conduction current density were analyzed using 3D finite element method (FEM) formulated for a dc conduction problem. The electrode positions for minimal current injection were optimized by changing the Cartesian coordinate system into the spherical coordinate system and applying the (2+6) evolution strategy (ES) algorithm. Preliminary simulation studies applied to a standard three-layer head model demonstrated that the proposed approach is promising in enhancing the performance of tDCS.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Transcranial direct current stimulation (tDCS) is a technique to stimulate the brain with direct current (dc) generated at few (usually two) electrodes attached on the subject's scalp (Nitsche and Paulus 2000, Priori 2003, Holdefer *et al* 2006, Miranda *et al* 2006). Compared to transcranial magnetic stimulation (TMS) (Salinas *et al* 2007, Miranda *et al* 2007), tDCS stimulates relatively wide brain regions, but is thought to be appropriate for the stimulation of relatively deeper brain areas than TMS, because the magnetic induction current elicited by TMS is generally distributed at shallow cortical areas (Im and Lee 2006). tDCS has various potential applications in areas such as depression (Fregni *et al* 2005), electro-analgesia (Mignon *et al* 1996), epilepsy (Fregni *et al* 2006) and so on (Fregni and Pascual-Leone 2007).

In the practical application of tDCS, the amplitude of the injection current should be considered carefully because large injection current density near the electrodes may cause side effects such as skin burn and muscle spasm. Moreover, it is obvious that excessive current injection causes unnecessarily widespread brain activation. Considering these two aspects, it is desirable to reduce the injection current to a minimal value, while maintaining the desired amount of current at a target brain area. Since for a given head model the current density distribution is influenced only by the scalp electrodes locations, injecting the minimal current can be realized by optimizing the electrode locations. This problem is equivalent to a simpler problem searching for two electrode locations which can generate maximum current flow at the target brain area with a fixed injection current.

In the present study, the positions of the two scalp electrodes were optimized by transforming the position parameters in a Cartesian coordinate system into angle parameters in a spherical coordinate system and applying the (2+6) evolution strategy (ES) algorithm (Im *et al* 2004). 3D finite element method (FEM) formulated for a dc conduction problem was used for the electric field analysis (Jin 2002). We applied the proposed approach to two case studies, for which a standard three-layer head model was used.

2. Methods and materials

2.1. Problem definition and finite element method (FEM)

Analysis of conduction current density is essential in the analysis of electrical stimulation systems. Conventional studies have used a simplified head model (Miranda *et al* 2006) or a single slice brain model (Holdefer *et al* 2006) for the numerical analysis of the 3D conduction current. In the present study, full 3D analysis was performed using a 3D steady-state FEM solver. Figure 1 shows a head model used for the present simulation study, which consists of scalp, skull and cerebrospinal fluid (CSF) regions. Since the inhomogeneous electrical conductivity distribution of a human head cannot be estimated accurately even with the currently best imaging modalities, we used well-known effective electrical conductivity values of the head structures. The effective electrical conductivity values for the scalp, skull and CSF regions were set to be 0.22, 0.014 and 1.79 (S m^{-1}), respectively (Haueisen *et al* 1997). The finite element model was extracted from structural MRI data of a standard brain atlas (http://www.mrc-cbu.cam.ac.uk/Imaging/Common/mnispace.shtml#evans_proc) and tessellated into 118,433 tetrahedral elements with 19,981 nodes, using CURRY5 for windows (Compumedics, Inc., El Paso, TX) and a free mesh generation software package *Tetgen* (http://tetgen.belios.de). Considering very low frequency (<2 Hz) current conduction, the following quasi-static Laplace equation was used as the governing equation of FEM:

$$\nabla \cdot (\sigma \nabla V) = 0, \tag{1}$$

where σ and V represent the electrical conductivity and electric potential, respectively. We have used a first-order finite element formulation and an ICCG matrix solver (Jin 2002). The two scalp electrodes were modeled as two nodes having different Dirichlet-type boundary values (0 V and 1 V).

2.2. Optimization of electrode locations

The aim of the present study was to search for two optimal electrode positions which can generate a desired current vector toward a certain direction at a target brain area, with minimal current injection. This problem is equivalent to a simpler problem searching for two electrode locations which can generate maximal current toward a certain direction at the target brain area, with fixed current injection. As aforementioned, we imposed fixed boundary conditions (0 V and 1 V) to two electrode locations in order to solve the latter problem. The simplest way to determine the optimal electrode locations is to evaluate the current value at the target position

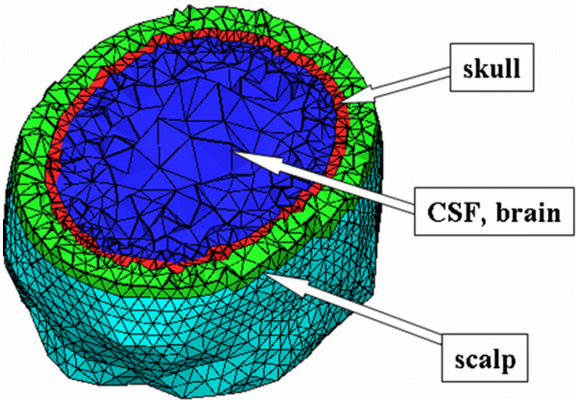

Figure 1. A finite element head model used for the present simulation study: the effective conductivity values for scalp, skull and CSF regions were set as 0.22, 0.014 and 1.79 (S m^{-1}), respectively. The finite element model was extracted from structural MRI data of a standard brain atlas and tessellated into 118,433 tetrahedral elements and 19,981 nodes.

for the every possible combination of two nodes on the outmost boundary. Considering the large number of boundary nodes (2342 nodes) and heavy computational cost of FEM, however, such exhaustive scanning was not practical.

Inspired by the fact that the shape of the scalp surface can be roughly approximated as a sphere, we transformed the location parameters in a Cartesian coordinate system into angle parameters in a spherical coordinate system. First, the center of the head was approximately determined by searching for the best-fitting sphere. Then, two angles in the spherical coordinate (θ and φ) with respect to the center of the sphere were evaluated for every boundary node. We used four angle parameters (θ_1, φ_1), (θ_2, φ_2), representing two scalp electrodes, as the optimization variables. Once the four angle parameters had been determined during the optimization processes, fixed boundary conditions (0 V and 1 V) were imposed at two boundary nodes which have the most similar angle parameters. Then, the (2+6) evolution strategy (ES) with an improved selection scheme was applied to the optimization (Im *et al* 2004). The population sizes 2 and 6 were determined empirically after some preliminary simulations. The range of θ was set as (0°, 120°), in order to prevent electrodes from being attached on the bottom of the head model. The range of φ was set as (0°, 360°). The ES iteration continued until the quality of solutions was not improved any more or the mutation range decreased to be smaller than a predetermined level (0.1% of the initial mutation range).

3. Simulation results

For the simulation model presented in figure 1, we first placed a test target point at a central posterior region (around the occipital area). The target point was arbitrarily selected without considering any physiological meanings. The optimization processes elucidated in the previous section were then applied. Figure 2 shows the location of the target position and the two electrode locations obtained after 123 ES iteration, where the results were viewed from two different viewpoints. In the simulation, the current direction at the target position was assumed as superior–anterior to inferior–posterior direction. The streamlines show the directions of the current and their colors represent the magnitude of the current density. To check the convergence characteristics of the optimization process, we applied the ES

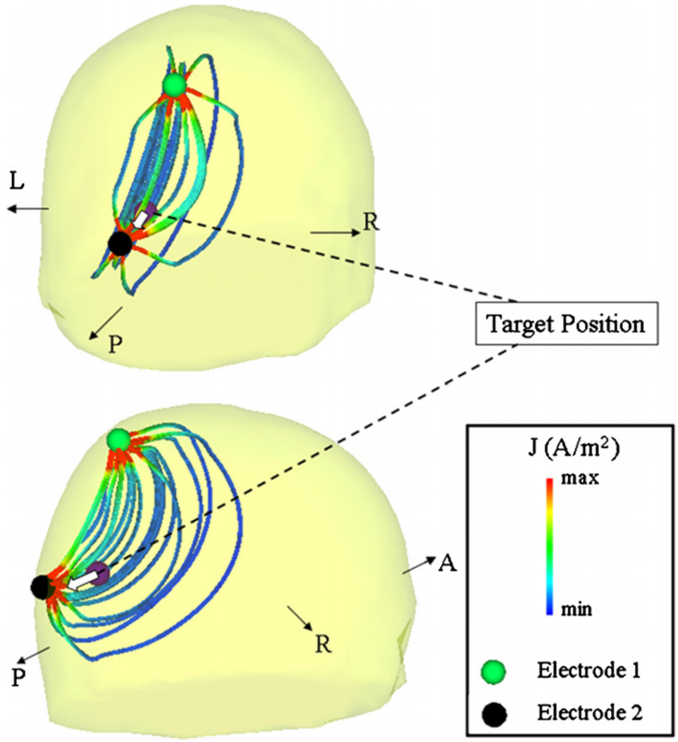

Figure 2. Optimized electrode locations and streamlines of current starting from multiple points around electrode 1, when the target brain area is around central posterior position (A: anterior, P: posterior, R: right, L: left). The two figures were obtained from different viewpoints. The streamlines show the directions of the currents and their colors represent the magnitude of the current density. Note that the density of streamlines does not reflect the current density distribution because they depend on the starting points. The current density at the target was 0.117 A m^{-2}.

optimization repeatedly to the same problem. We confirmed from the simulations that the results of the ten executions converged to the same parameters.

Figure 3 shows another example of the tDCS optimization problem, where the test target point was placed at a slightly deeper brain region (temporal lobe area) than that of the previous example. The optimal electrode locations were determined after 147 ES iterations. We also applied the optimization processes repeatedly and confirmed that the results of the ten independent executions converged to the identical angle parameters.

4. Discussions

In the present study, a new approach to determining optimal locations of stimulation electrodes in a tDCS system was introduced. With the proposed optimization processes using (2+6) ES and 3D FEM, we could readily find out optimal scalp electrode pairs which can stimulate target brain areas with minimal injection current.

In these preliminary simulation studies, an FEM model was constructed from a standard head MRI data set. The proposed approach can be readily applied to individual head model as the recent development of medical image analysis techniques made it easy for us to extract head

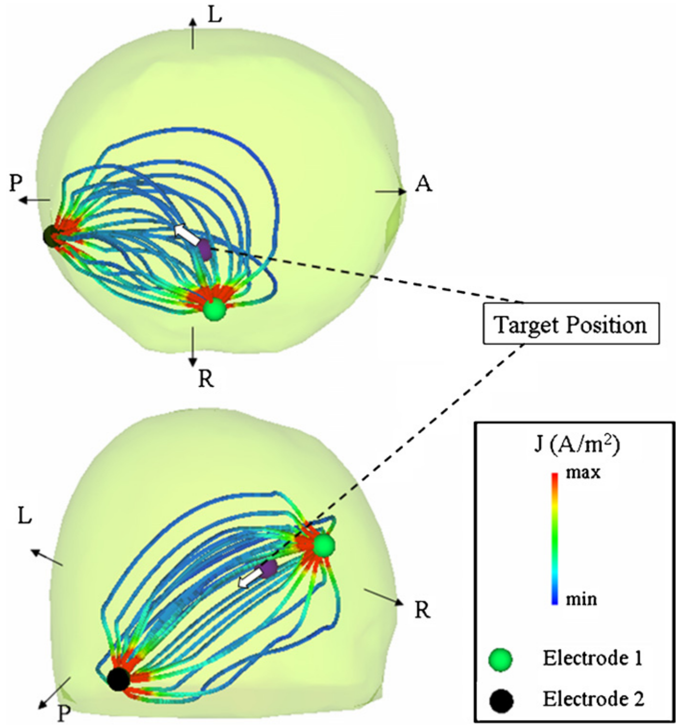

Figure 3. Optimized electrode locations and streamlines of current starting from multiple points around electrode 1, when the target brain area is located in a deeper position (A: anterior, P: posterior, R: right, L: left). The two figures were obtained from different viewpoints. The streamlines show the directions of the currents and their colors represent the magnitude of the current density. Note that the density of streamlines does not reflect the current density distribution because they depend on the starting points. The current density at the target was 0.077 A m^{-2}, which is smaller than that of the previous example because the target is further from the electrodes.

structures from individual MRI data sets and generate realistic FEM models. Therefore, once one decides a target brain area and the direction of the current based on neurophysiological consideration (the target direction should be determined carefully considering the direction of cortical pyramidal neurons) in the subject's anatomical brain images, such information can be readily reflected in the proposed optimization processes.

We applied (2+6) ES as the optimization algorithm, but any other optimization algorithms such as genetic algorithm (GA) and simulated annealing (SA) can be used. According to a recent study (Im *et al* 2004), ES is faster than GA and less likely to be trapped in local minima than SA. In this study, we focused on finding out optimal electrode locations which minimize the injection current, but we expect that the same optimization processes can be applied to other relevant problems in the future, e.g., finding out optimal multiple electrode locations which can activate multiple foci.

One important issue that has to be considered carefully before applying the proposed approach is the computational time. In our preliminary simulations, a single execution of the FEM solver took about 10.2 s in a Pentium IV 3.2 GHz personal computer environment and the first optimization results in figure 2 were obtained in about 2 h. Therefore, if one wants to obtain just rough electrode locations in practical applications, it is desirable to reduce the model order (the number of tetrahedral elements). In our future studies, we will perform several simulations with different head models, which can be used to determine the model orders adequate for the specific applications.

As in the other neuroelectromagnetic problems, the tDCS field analysis is hard to verify with *in vivo* experiments because we can hardly measure the electrical current flow directly in the human brain. Therefore, future studies should be focused on the experimental validation of the tDCS field analysis. It is expected that human skull phantom experiments and simultaneous tDCS-EEG recordings might be the alternative options.

In the present simulation study, we did not separate the CSF and brain regions, which have slightly different conductivity values (the effective conductivity values of CSF and brain are 1.79 and 0.22 S m^{-1}, respectively), in order to reduce model complexity due to the folded cortical structures. Although the brain region has been often neglected in similar current conduction problems such as EEG forward calculations (Liu *et al* 2002), it can be taken into account in our future studies to enhance the computational accuracy.

As mentioned in the introduction, TMS (including repetitive-TMS) has several advantages over tDCS in that TMS is less invasive and its induced current distribution is more focalized than tDCS. Nevertheless, tDCS is still useful for patients who need long-term or frequent therapy. The patients cannot move their heads at all during the TMS therapy, while they can freely move their heads during the tDCS therapy. The tDCS system can be used even in mobile environments if it is installed in a wheelchair or a bed. We expect that our work will make a contribution to the popularization of tDCS because the approach proposed in the present study can be a potential solution for the targeting and safety issues of the tDCS system.

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