The Importance of the Numerical Resolution of the Laplace Equation in the optimization of a Neuronal Stimulation Technique

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Abstract. For the past few years, the potential of transcranial direct current stimulation (tDCS) for the treatment of several pathologies has been investigated. Knowledge of the current density distribution is an important factor in optimizing such applications of tDCS. For this goal, we used the finite element method to solve the Laplace equation in a spherical head model in order to investigate the three dimensional distribution of the current density and the variation of its intensity with depth using different electrodes montages: the traditional one with two sponge electrodes and new electrode montages: with sponge and EEG electrodes and with EEG electrodes varying the numbers of electrodes. The simulation results confirm the effectiveness of the mixed system which may allow the use of tDCS and EEG recording concomitantly and may help to optimize this neuronal stimulation technique. The numerical results were used in a promising application of tDCS in epilepsy.

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INTRODUCTION

Transcranial direct current stimulation (tDCS) is a non invasive and painless technique that has been shown to modulate cortical excitability [9]. Its application is also very easy and economical because only two surface electrodes, which are connected to a current stimulator, are needed: one anode and one cathode. In the most commonly used configuration (the traditional montage) one rectangular electrode (25 cm^2) is placed over the region of interest, e.g. the motor cortex, and the other (also rectangular and with the same size) is placed away from the first one, e.g. above the contralateral eyebrow. A weak DC current (less than 2 mA) is injected between the two electrodes. These advantages combined with the fact that tDCS has been showing promising results as a therapy for several pathologies like stroke [3], Parkinson's disease [4], epilepsy [5] and depression [6], have interested several groups of investigators. An overview of recent tDCS experiments and methodological issues has been published recently [11]. However, the spatial distribution of the current density inside the head and the adequate electrode configuration for specific tDCS applications are not fully understood.

The traditional configuration used in tDCS has several drawbacks in terms of focality [10] and control of the impedances at the electrode-scalp interface [12]. In this work we compare different electrode montages using sponge and EEG electrodes: one with two sponge electrodes (M1), one mixed system with a sponge electrode and an EEG electrode (M2), another mixed system with four EEG electrodes (M3) and one montage with two EEG electrodes (M4). We used the finite element method to solve the Laplace Equation and compute the 3-D distribution of the current density in a spherical head model using the previous described electrode montages. The use of the proposed EEG electrodes help to easily identify the electrode positions in the 10-10 International System and also the electrode-skin impedances can be easily monitored. Importantly, these electrodes need less current compared to the larger electrodes, and allow the application of tDCS with the EEG recording concomitantly, which is of great importance in terms of safety, particularly because it allows the monitoring of the interictal activity in epilepsy patients.

The numerical results helped to choose an optimized electrode montage in terms of focality and stimulation in depth which was tested in an application of tDCS in refractory epilepsy with two epileptic patients.

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METHODS

The current density is a vector function, $\overrightarrow{J}(x,y,z)$, defined at every point in a conductive medium, whose direction is that of the current flow at the point under consideration and whose magnitude is given by the current divided by the area perpendicular to the flow, as this area tends to zero. The current density is obtained from the electric field, $\overrightarrow{E}$, by means of the relation $\overrightarrow{J} = \sigma \overrightarrow{J}$, where σ is the electric conductivity of the tissue. In turn, the electric field is determined by the spatial rate of change (gradient) of the electric potential, ϕ, i.e. $\overrightarrow{E} = -\nabla\phi$. Finally, the potential inside the conductive medium is obtained by solving the continuity equation, $\nabla \cdot (\sigma\nabla\phi) = 0$, subject to the appropriate boundary conditions.

Following the approach outlined above, we calculated the current density distribution in a spherical head model for various electrode configurations, using a finite element package (Comsol 3.3 with AC/DC module, http://www.comsol.com) to solve the continuity equation numerically. Electrodes were modeled as square sponges and EEG electrodes. The former is based on the electrodes supplied by Amrex-Zetron, Inc. (www.amrex-zetron.com) and consists of a metal mesh held over a 1 cm thick sponge by a rubber frame. The sponge is soaked in physiologic saline solution before being applied to the scalp. The latter is based on the electrodes supplied by Easycap - EEG Recording Caps and Related Products (www.easycap.de), and consists of an AgCl sintered ring EEG electrode (0.2 cm high) which is snapped into the adaptor (0.245 cm high) of the EEG cap. The area of the electrodes is 5×5 cm^2 for the sponge electrode and 1.1 cm^2 for the EEG electrodes. The sponge electrode was modeled as described in [8] and the EEG electrode was modeled similarly but with two circular cutouts from a 0.2 cm and 0.245 cm spherical shells placed over the scalp (see Fig. 1), representing the EEG electrode and the adaptor respectively. The gel surface in contact with the conductive electrode surface was set to be at a uniform electric potential. The electric conductivity value of the sponge electrode used was taken to be equal to that of the scalp, $\sigma = 0.332$ S/m and for the EEG electrode we used the value 10 S/m, which was measured experimentally.

The 3D spherical head model of Rush and Driscoll [13] was used and adapted in order to implement a four-layer spherical head model. The latter contains four homogeneous and isotropic layers representing the scalp, skull, cerebrospinal (CSF) and brain. The radii and the electric conductivity values were the following $r_{\text{brain}} = 7.9$ cm, $r_{\text{CSF}} = 8.1$ cm, $r_{\text{skull}} = 8.6$ cm and $r_{\text{scalp}} = 9.2$ cm; $\sigma_{\text{brain}} = 0.332$ S/m, $\sigma_{\text{CSF}} = 1.79$ S/m, $\sigma_{\text{skull}} = 0.0083$ S/m and $\sigma_{\text{scalp}} = 0.332$ S/m (see references at [2]). The spherical head model was centered on the origin of an orthonormal reference frame where the x-axes passes through the left and right pre-auricular points and the y-axes passes through the nasion. The positions of the $10-10$ International System electrodes are represented in the model by circles (see Fig. 1 - right).

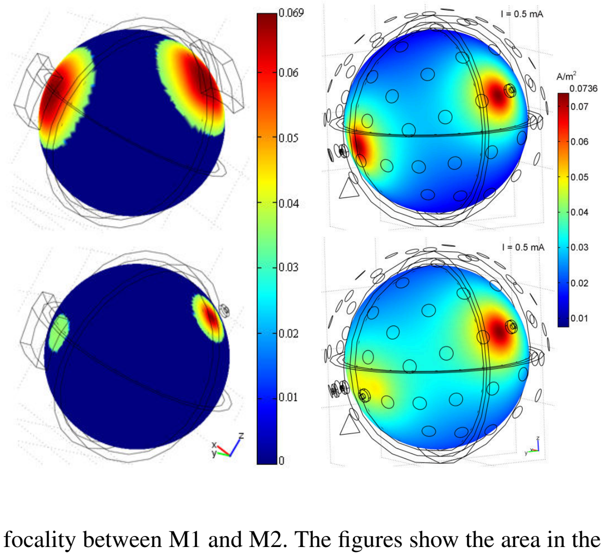

FIGURE 1. Comparison of the focality between M1 and M2. The figures show the area in the brain where the magnitude of the current density is higher than 50% of the maximum of the current density in the brain, 0.069 A/m^2 (left). The distribution of the magnitude of the current density at the surface of the brain for M3 and M4. For both configurations an injected current of 0.5 mA is needed to achieve a current density of 0.073 A/m^2 in the brain under the center of the cathode (right).

The coordinates of the $10-10$ International System electrodes were obtained using the program Source V (www.neuroscan.com/source.cfm) and projected onto the spherical model. The stimulation electrodes used in the

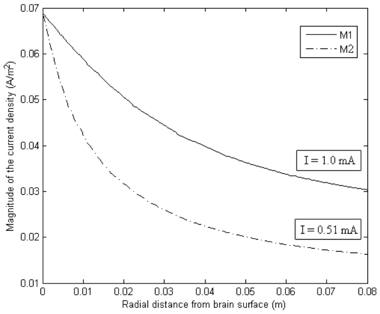

FIGURE 2. Comparison of the magnitude of the current density distribution in the brain along a straight radial line in the direction of the center of the electrode placed over the motor cortex.

four models were placed as follows: one with two sponge electrodes placed at FPz and CP5 (M1), one mixed system with a sponge electrode at FPz and an EEG electrode at CP5 (M2), another mixed system with four EEG electrodes placed at FP1, FPz, FP2 and CP5 (M3) and one montage with two EEG electrodes at Fpz and CP5 (M4). The potential difference between the anode(s) and the cathode was adjusted so that at a point P on the brain surface and located radially under the cathode (CP5), the magnitude of the current density was the same for all the models. We estimated that this is the current density in the brain using a standard configuration of 1 mA into 25 cm^2 electrodes, which has been shown to modify the excitability in the human brain [9]. By doing this, we ensured the same current density for each model at point P.

In all models the unique cathode was placed in the CP5 position. This choice was motivated by the case of a patient with an epileptogenic focus localized under this electrode position. Since cathodal stimulation has been shown to decrease cortical excitability, this setup may be relevant for this patient. For montage M1, M2 and M4, the unique anode was placed at FPz. For montage M3, the three anodes were placed in the frontal cortex where there was no epileptogenic activity (FP1, FPz and FP2). This montage is similar to the one traditionally used to stimulate the motor cortex [9].

In this study we start by comparing M1 and M2 and then M3 and M4 in terms of focality and distribution of the current density in depth along a straight radial line (S) that passes through the center of the cathode. The focality of the models was quantified through the calculation of the area (A50) of the brain where the current density was within 50% of its maximum power at the brain surfaced [1]. The numerical results obtained were tested in an application of tDCS in epilepsy with two epileptic patients.

RESULTS

Comparison of four different electrode montages

The first electrode montage considered used two sponge electrodes (M1). Fig. 2 allows for the comparison of the variation of the current density in depth along a straight radial line in the brain (S) in the direction of the center of the electrode placed over the motor cortex. The results show that, at the surface of the brain, the M2 configuration achieves the same current density with half of the current injected in the M1 montage. Additionally, the current density induced in M2 decreases more rapidly in depth as compared to the M1 montage.

At a point 2 cm away from the inner surface of the skull, for instance, the magnitude of the current density in M1 is 73% of the maximum value obtained in the brain whereas for the M2 this value is only 46%. The comparison of the focality obtained using M1 and M2 electrode configuration was done by calculating the area of the brain where the magnitude of the current density was 50% of the maximum value of the current density, 0.069A/m^2 in each models. The results can be seen in Fig. 2, for this case. In terms of focality the A50 value obtained for M1 was 173 cm^2 and for M2 was 26 cm^2.

The variation of the current density with depth along a radial line in the brain (S) passing through the center of the cathode, for configurations M3 and M4 was studied and the results show similar results for both montages. For example, the same current density at the surface of the brain, 0.073 A/m^2, is attained with less injected current for the M3 and M4 configurations with 0.5 mA. At a point 1 cm below the inner surface of the skull, the magnitude of the current density for M3 and M4 is approximately 64.5% of the maximum value obtained in the brain. In terms of focality the A50 areas were 19.9 cm^2 for M3 and 10.0 cm^2 for M4.

Clinical Application

The numerical results allowed the choice of the best electrode montage that was tested in an application of tDCS in epilepsy with two epileptic patients. The results show that this electrode montage was able to reduce the epileptic activity of two refractory epileptic patients.

DISCUSSION

This study quantifies the current density in the spherical brain model using different electrodes montages. This type of model was proved to give a good accuracy for the target region studied in this work [7]. The comparison of the four electrode montages presented in this work suggest that the electrode montage M3 can help increase focality in the cortex and allow for effective stimulation in regions that are located at the surface of the brain. This mixed electrode montage has also the advantage that it may allow the acquisition of the EEG signals during the application of tDCS.

The results show that, the use of M3 may improve the interpretation of the functional effects of stimulation because it will restrict its effects to more clearly defined cortical areas. Moreover, it may avoid unwanted reversed effects of tDCS under the anode, which is of special importance in clinical settings, such as for the treatment of epilepsy [10]. This kind of modelling may be useful to help identify the epilepsy focus. We can obtained, numerically, the exactly position for the cathode and use that information to stimulate the target region, experimentally, and confirm the precise localization of the epilepsy focus. This shows that tDCS is a promising tool for the treatment of this pathology.

The clinical results are encouraging, but it has yet to be demonstrated that it is possible to implement a tDCS system that is safe and well tolerated and has enough cortical polarization power to modulate epileptic activity. If this proves to be the case, then a bright future for the use of the technique in epilepsy is at hand. The proposed system opens interesting perspectives for the application of tDCS in the modulation of the epileptogenic foci in humans, allowing a detailed characterization of the EEG activity during the stimulation procedure and resulting in increased sensitivity and safety.

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