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Bio-heat Transfer Model of Transcranial DC Stimulation: Comparison of Conventional Pad versus Ring Electrode

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Abstract— **Transcranial Direct Current Stimulation (tDCS) is a non-invasive procedure where a weak electrical current is applied across the scalp to modulate brain function. The proliferation of this therapy has been accompanied by isolated reports regarding concern about their safety namely skin irritation. The potential cause of skin irritation has sometimes been attributed to increased scalp temperature during stimulation. We have developed novel technology for tDCS that improves spatial focality at the cost of increased stimulation electrode current density; high density tDCS (HD-tDCS). The goal of this paper was to provide information on the thermal effects of tDCS using a MRI-derived finite element human head model. The tissue temperature increases of tDCS using conventional rectangular-pad (7 X 5 cm^2) and HD-tDCS using the ring (4 X 1) electrode configurations were compared using a bio-heat model. Our results indicate that clinical tDCS do not increase tissue temperature and 4 X 1 ring configurations leads to a negligible increase in scalp temperature.**

I. Introduction

Conventional transcranial direct current stimulation (tDCS) involves weak direct currents (1–2 mA) applied to the scalp via sponge-based rectangular pads (nominally 25-35 cm^2) for 10-20 min [1],[2]. tDCS is generally considered safe and comfortable (due to the low stimulation current intensities), however isolated reports on skin irritation/damage have suggested local scalp temperature increases as one probable cause [3],[4]. It is known that electrical stimulation of tissues may lead to temperature increases as a result of both joule heat and metabolic responses to stimulation [5]. The aim of the present paper was to develop a bio-heat transfer model of tDCS and thereby investigate whether tDCS currents would lead to tissue temperature increases.

The spatial focality (targeting) of tDCS has been proposed to increase using a "ring" electrode configuration with electrodes < 11 mm in diameter: 4 X 1 ring [6],[7]. Such stimulation electrodes owing to their proximity and reduced area are referred to as high density (HD) electrodes. The magnitude and spatial distribution of induced temperature changes for conventional big pad tDCS and the 4 X 1 ring (HD-tDCS) configurations are calculated and compared in this paper.

Brain function is sensitive to changes in temperature. An increase in temperature by ~ 1 °C can affect properties of single neurons and neuronal network function [5]. Increases to ~ 40°C for an extended period of time may lead to tissue damage. In addition, the local tissue temperature changes are typically affected by various factors, namely: 1) heat conduction to surrounding tissue; 2) advection (blood flow); 3) scalp heat loss to ambient due to convection, and 4) other thermoregulatory responses (sweating, pilo erection) [8].

Experimental observations have indicated no temperature rise underneath the pads for conventional tDCS [1]. However, tDCS tissue temperature increases has not been previously determined in the scalp and the brain. A realistic MRI derived bio-heat transfer model can guide development of rational therapy and establishing safety standards. By solving the coupled Laplace equation of electrical field and the Pennes bio-heat transfer equation, we simulate how tDCS affects temperature field distribution in the tissue.

II. Methods

A. MRI derived human head model

The volume conductor 3D model used in this study was developed previously by our group to calculate tDCS induced electric fields [7]. High resolution (1 mm^3) raw 3T MRI scans were contrast enhanced and noise filtered. The head was segmented into compartments representing the brain tissue, cerebrospinal fluid, skull, muscle, fatty tissue, eyes, blood vessels and the scalp respectively (SIMPLEWARE Ltd., Exeter, UK). The solutions for the coupled temperature and electric field for the whole head model at 1 mm^3 resolution would demand significant memory resources. The 3D model was thus resampled to 2 mm^3 which provided an optimal compromise between computation time and model size (accurate representation of tissues of interest). The stimulation rectangular pads, discs, gels were imported as CAD models and positioned within image data [7]. The volumetric mesh was generated (minimum quality factor > 0.5) from the segmented data and eventually exported to COMSOL Multiphysics 3.5 (COMSOL Inc., MA, USA). The resulting finite element model comprised >5,000,000 tetrahedral elements (>20,000,000 degrees of freedom).

Tissue properties were assigned representative average

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values obtained from literature and are listed in Table 1. The muscle, fatty tissue, eyes and blood vessel compartments were assigned the same tissue properties as that of scalp [7]. For the CSF compartment, blood flow rate and metabolic heat is considered zero [8],[9].

TABLE I
TISSUE ELECTRICAL AND THERMAL PROPERTIES

Tissue	Electrical Conductivity (σ) (S/m)	Thermal Conductivity (κ) (W/m.°C)	Blood flow Rate (ω_b) (1/s)	Metabolic heat (Q_{met}) (W/m^3)
Scalp	0.465	0.39	0.00143	363
Skull	0.01	1.15	0.000143	70
CSF	1.65	0.61	0	0
Brain	0.2	0.57	0.08	10437

B. Heat Transfer Model

An approximate temperature distribution throughout a perfused tissue can be found by solving the bio-heat transfer equation suggested by Pennes [9]:

$$\rho C \frac{\partial T}{\partial t} = \nabla(k\nabla T) - \rho_b \omega_b C_b (T - T_b) + Q_{met} \quad (1)$$

where ρ: density; C: specific heat; κ: thermal conductivity; ω_b: blood perfusion rate; T: temperature; T_b: arterial blood temperature; Q_{met}: metabolic heat.

During electrical stimulation, additional joule heating arises when energy is dissipated by an electric current flowing through a conductor. The Laplace equation $\nabla.(\sigma\nabla V) = 0$ (V: potential; σ: conductivity) was solved to determine the electrical potential. The Pennes equation (1) above is modified to incorporate joule heating during electrical stimulation [5]. In addition, we considered steady state temperature increases consistent with prolonged tDCS protocols (>10 minutes) [1],[2]; resulting in the following bio-heat equation:

$$\nabla(-k\nabla T) = -\rho_b \omega_b C_b (T - T_b) + Q_{met} + \sigma|\nabla V|^2 \quad (2)$$

C. Electrode Configurations

We modeled two electrode configurations:

(1) 'Rectangular-pad' (Figure 1A): Two pads (7 X 5 cm^2) were placed at sites typically used for tDCS of the primary motor cortex [7]. Rectangular sponges are typically soaked in saline for conventional tDCS application and the abutting electrode is energized. The sponge was therefore assigned the electrical and thermal properties of saline (σ : 1.4 S/m; κ = 0.3 W/m.°C).

(2) '4 X 1 ring' (Figure 1B): Four cathode disc electrodes were arranged in a circular fashion around an anode center electrode [7]. The anode electrode is placed over the motor cortex (coinciding with the center of the anode pad used for rectangular-pad stimulation) and surrounded by four cathode electrodes (each at a disc center to disc center distance of 4 cm from the anode electrode). All disc electrodes had an 11 mm diameter. Current was conducted through disc electrodes into the scalp using a customized tDCS gel (CCNY-4) with electrical conductivity: 0.3 S/m and thermal conductivity: 0.6 W/m.°C.

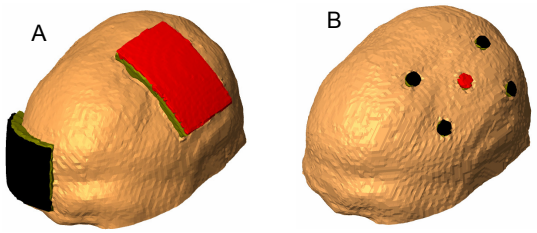

Fig. 1. Finite element model of conventional 7 X 5 cm^2 rectangular-pad and 4 X 1 ring configurations (Red: Anode electrode, Black: cathode electrode; Olive Green: Sponge/Gel)

All electrodes were modeled as conductors with σ = 5.8 x 10^7 S/m and κ = 31 W/m.°C .

D. Initial and Boundary Conditions

The electrical boundary conditions used were (1) inward current flow = J_n (normal current density) applied to the exposed surface of the anode electrode, (2) ground applied to the exposed surface of the cathode electrode(s) and (3) all other external surfaces treated as insulated. Current densities corresponding to 1 mA total current for the rectangular pad configuration and 2 mA for the 4 X 1 ring configuration were respectively applied [7]. 2 mA of current applied through the 4 X 1 ring configuration results in similar electric field magnitude peak directly *underneath* the pads for the conventional rectangular-pad stimulation [7].

The heat loss at the scalp surface may be due to both sweat evaporation and convection to ambient [8],[10]. However, heat loss due to sweat evaporation was not considered in this study. Thus for the thermal boundary conditions, all external boundaries of the head model were assigned the following heat flux:

$$q = h(T_{amb} - T) \quad (3)$$

T_{amb} : External temperature = 24 °C
h: Heat Transfer coefficient = 4 W/m^2.°C

The following parameters were applied: ρ_b = 1050 Kg/m^3; c_b = 3600 J/(Kg.°C); T_b = 36.7 ºC [5]. The electrodes, sponge, gel were assigned the initial temperature of T_{amb}. All the tissue compartments were assigned the initial temperature (T_0) of 37 °C.

Estimates of steady state tissue temperatures were first obtained by evaluating the model under prescribed conditions with zero applied current density (no stimulation). The solution of the model with injected current density was then compared with the 'no stimulation' condition to evaluate *relative* tissue temperature increases. In addition for the 4 X 1 ring, we calculated tissue temperature increases for

injected current density (J_n = 142.9 A/m^2); corresponding to 13.58 mA total current. The aforementioned current density value has been reported to the minimum electrode current density at which brain lesions are observed in a tDCS rat model for stimulation duration greater than 10 minutes [11].

III. Results

For the 7 X 5 cm^2 rectangular-pad (1 mA) and the 4 X 1 ring (2 mA and 13.58 mA) configuration models (Fig. 1A and 1B), we calculated the induced tissue temperature distribution due to stimulation (Fig. 2).

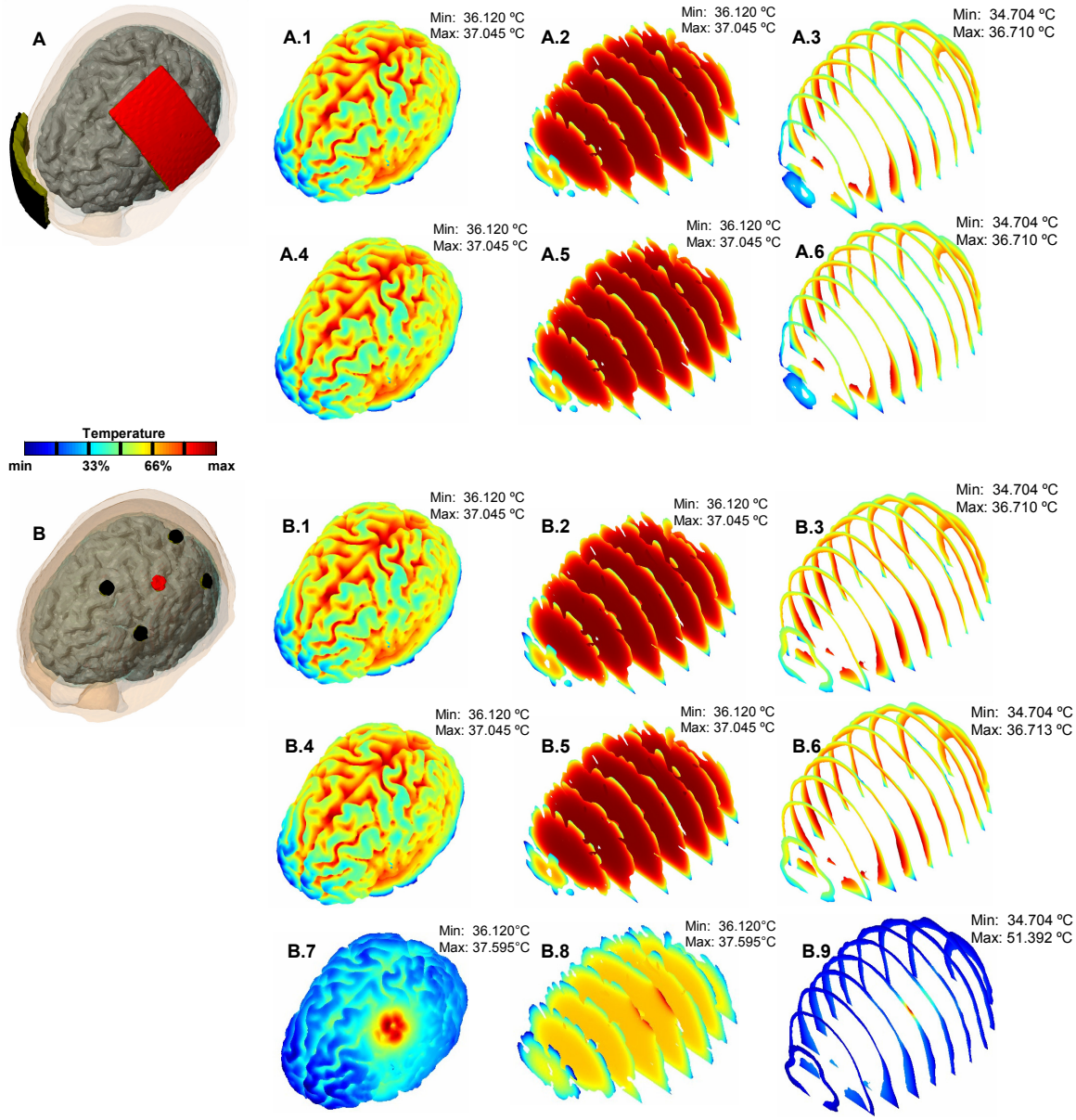

Fig. 2. Bio-heat transfer model of transcranial direct current stimulation using conventional rectangular-pad (A) and 4 X 1 ring electrode configuration (B). For each configuration, we calculated the induced cortical temperature distribution due to stimulation. The first row of each of the two configurations (A.1 - A.3, B.1 - B.3) models the '*no-stimulation*' condition (see Methods). The second row (A.4 - A.6) of the rectangular pad configuration models the 1 mA '*stimulation*' condition. The second and third rows (B.4 - B.6, B.7 - B.9) model the 2 mA and 13.58 mA '*stimulation*' conditions respectively (see Methods). The first two columns plot the calculated induced cortical surface and cross- sectional temperature distributions. The last column plots the induced temperature distribution on a series of cross-sectional slices.

Conventional rectangular-pad stimulation results in no tissue temperature rise in the scalp or brain (Fig. 2A.4 and 2A.6).

The 4 X 1 ring (2 mA) configuration does not lead to a brain tissue temperature rise (Fig. 2B.4); similar to conventional pad stimulation. However, negligible tissue temperature rises are observed at the scalp (<0.01 °C).

For the 4 X 1 ring (13.58 mA), significant scalp temperature rise of 14.68 °C (Fig. 2B.9) and a nominal brain temperature rise of 0.55 °C is predicted (Fig. 2B.7).

IV. DISCUSSION

Our results indicate conventional tDCS protocols do not cause tissue temperature increases in the scalp. It should however be noted that magnitude of temperature increases critically depend on the electrical/thermal properties and stimulation settings considered. Our results are consistent with studies where scalp temperature increases underneath the pads [1] and HD electrodes [12] were measured using a thermocouple.

The observed temperature increases in brain (which is a function of current flowing into the tissue) is electrode montage specific. This is indeed expected, given the particular electrode montage determines the magnitude and the spatial extent of current entering the brain [6],[13]-[15]. In the Liebetanz et al. study, electrode current density of 142.9 A/m^2 was found to be the threshold for rat brain lesions (presumably due to burning) using a montage that maximizes current fraction entering the brain [11],[16]. However, using the same surface electrode current density, we predict minimal brain temperature rises for the 4 X 1 ring; reflecting significant current shunting across the scalp. Moreover, a potentially hazardous temperature rise was predicted at the scalp. This highlights the importance of independently considering safety limits for scalp and brain, for each electrode montage [6],[7].

Our model does not consider micro architecture of the skin which potentially might play a role in explaining cases of skin irritation observed in conventional tDCS. There have been studies where the skin has been considered to be composed of an outer unperfused layer (ω_b = 0), sweat glands, hair follicles, and a fat layer. An extension of this study would be to incorporate detailed architecture of the skin.

Additionally, in the present study we did not consider temperature induced changes due to variation in the electrical and thermal properties. A parametric study for evaluating temperature increases under all possible conditions is important and should be addressed in future.

The results of this present study provide an initial basis for determining tDCS induced temperature rises.

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