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Theoretical Investigation of Transcranial Alternating Current Stimulation using Laminar Model

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Abstract— **Transcranial alternating current stimulation (tACs) has been gaining an increased interest in the last few years due to its capacity to modulate non-invasively high-order cortical processes, such as decision-making, language and sensory perception. Nevertheless, the underlying mechanisms of activation of this brain stimulation technique are still poorly understood. Herein, we use a finite element modelling (FEM) technique to investigate the penetration and focality of tACs in comparison to a time invariant (DC) stimulation. We show that AC stimulations generate cerebral fields that are an order of magnitude larger in the radial direction, approximately 5 times larger in the tangential direction and more focused than DC stimulations. We argue that the basis for this effect is the reduced scalp's conductivity, which minimizes the surface shunting of the stimulating currents. The outcomes of this study may help tACs users to design better protocols and interpret experimental results.**

I. Introduction

Cranial electrical stimulation techniques have been used for many years and regained significant appeal for their capacity to safely modulate brain activity for the treatment of neuropsychiatric diseases such as anxiety, depression, stress, and insomnia. It is recognized by the American food and drug administration (FDA) for treating anxiety, depression (serotonin effect) and pain (endorphin effect). Typically, the stimulation protocol uses a battery-powered current generator device that is capable of delivering a constant electrical current of up to 2 mA. The device is attached to two electrodes that are soaked in saline (or water) and placed inside sponges that are then held in place by a non-conducting rubber montage affixed around the head.

The cerebral fields that are induced by the external electrical stimulation are determined not just by the amplitude of the injected currents but also by the dimension and location of the electrodes. Datta et al. [1] found that the typical distant-bipolar stimulation produces radial fields under each electrode and tangential fields between them. Sadleir et al. [2] confirm the distant-bipolar findings stating that the high current densities were found directly below each electrode. Yet due to a more realistic head model they were able to report that values of the same order of magnitude were found as well in other regions of the brain, which may become behaviourally active due to such currents. Later Dmochowski et al. [3] have corroborated the findings of Datta et al. [1] and Sadleir et al. [2]. Through a novel method of optimal electrode placing, Dmochowski et al. [3] conclude that if a bipolar configuration must be used then: when maximal intensity tangential current is sought the electrodes should be placed at a "considerable" distance from the target, one on each side, along the direction of desired current; when maximally intensive radial fields are needed then one electrode should be placed directly above the target and a return electrode far away. There is a fundamental trade-off between achievable intensity at target and focality, Dmochowski et al. [3]. The smaller the distance between the electrodes the higher the focality but this also leads to higher surface currents, thus for a set electrode current, a lower current reaches the brain. Holdefer et al. [4] studied a single coronal MRI section through the head and suggested that the closer the electrodes are more current is shunted through the scalp and less current reaches the brain.

Nitsche et al. [5] suggested that in order to increase focality of stimulation the anode (stimulating) electrode size should be decreased and the cathode (reference) electrode size should be increased. This is in line with the results produced by Datta et al. [1] "ring" and "belt" electrode configurations. Miranda et al. [6] disprove the long held assumption that the current distribution inside the brain was determined by the ratio of injected current to the electrode area (I/A), also called injected current density. It was shown that there are only two regions at which the injected current, I, and electrode area, A, are the main parameters determining the current density distribution: very close to the electrode (ignoring edge effects) and very far away (where the electrode is seen as a point source). Parazzini et al. [7], who investigated a transcranial electrical stimulation of the primary motor cortex using a realistic human head model with 40 different tissues, found that the standard tDCs electrode area results in cerebral fields that are not limited to the cortex, though the high percentile fields were mainly localized below or close to the anode. In principle, the area of the electrodes correlates with the spatial distribution of the cerebral fields while the current correlates with the fields' amplitudes.

Alternating current (AC) cranial stimulations, in contrast to direct current (DC), use oscillating currents with pulses of rectangular currents (intensity rapidly increased to a certain amplitude, held at the peak without change, and then interrupted by zero current) or sinusoidal waves (intensity constantly varies as a function of time). The impact of the

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osculating currents on the cerebral stimulating fields is at present poorly understood. The aim of this paper is to explore via simulation how these factors may impact the penetration and spatial distribution of the induced cerebral fields.

II. METHODOLOGY

A simplified five layer planar model of the head was developed and simulated in ANSYS Maxwell 3D (Ansoft Inc.). A schematic of the model is shown in Fig 1. The overall dimension of the model was 30cm × 30cm× 7.65cm. The thicknesses of the layers were chosen according to Datta et al. [1] with one difference that the brain was divided into grey and white matter. The depth of the grey matter was the average reported by Fischl and Dale [8]. The model was built with different conductivities for each frequency as reported in Table 1. The conductivities values that were assigned are taken from the Information Technologies in Society (IT'IS) database of tissue properties [9], which are based on Gabriel et al. in-vitro experimental characterization [10].

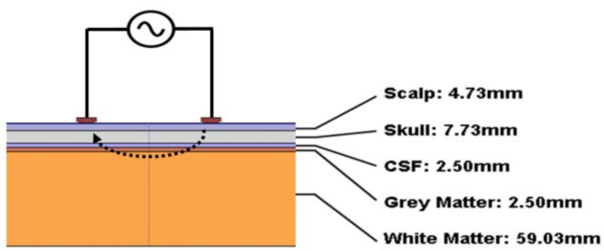

Fig. 1. Schematic illustration of the laminar head model.

Similar to Parazzini et al. [7] we used a weighted average of the skin and the subcutaneous adipose tissue for the scalp. The model did not take into account the contribution of the permittivity component of the tissues. The electrodes were modelled as solid copper disks with a diameter of 10mm and a thickness of 3mm. The electrodes were positioned at a distance of 60mm (similar to the 10/10 system), and a current of 1mA was generated between the electrodes. The mesh was generated using Ansoft's classic mesh generator and the fields were solved using the Maxwell 3D transient solver.

TABLE I. TISSUE CONDUCTIVITIES (S/M)

Layer	**Frequency**		
	0Hz	***10Hz***	***100Hz***
Scalp	4.65E-01	2.00E-04	2.00E-04
Skull	1.00E-02	2.00E-02	2.01E-02
CSF	1.65E+00	2.00E+00	2.00E+00
Grey matter	2.00E-01	2.75E-02	8.90E-02
White matter	2.00E-01	2.77E-02	5.81E-02

III. RESULTS

A. Fields Focality

Fig. 2 compares the distributions of the cerebral fields at the central cross section between the electrodes (i.e. a plane perpendicular to and in the centre of the inter-electrode direction). The plots reveal a clear difference in the localization of the fields at the brain. The full wave half maximum (FWHM) at the grey matter is 100mm in the case of DC and 60mm in the case of 10Hz, i.e. almost half the size. There is no significant difference in the field distribution between 10Hz and 100Hz (results are not shown). The main difference between tACs and tDCs occurs 50mm distance from the centre, where tDCs results in residual fields that can be 20% larger than tACs.

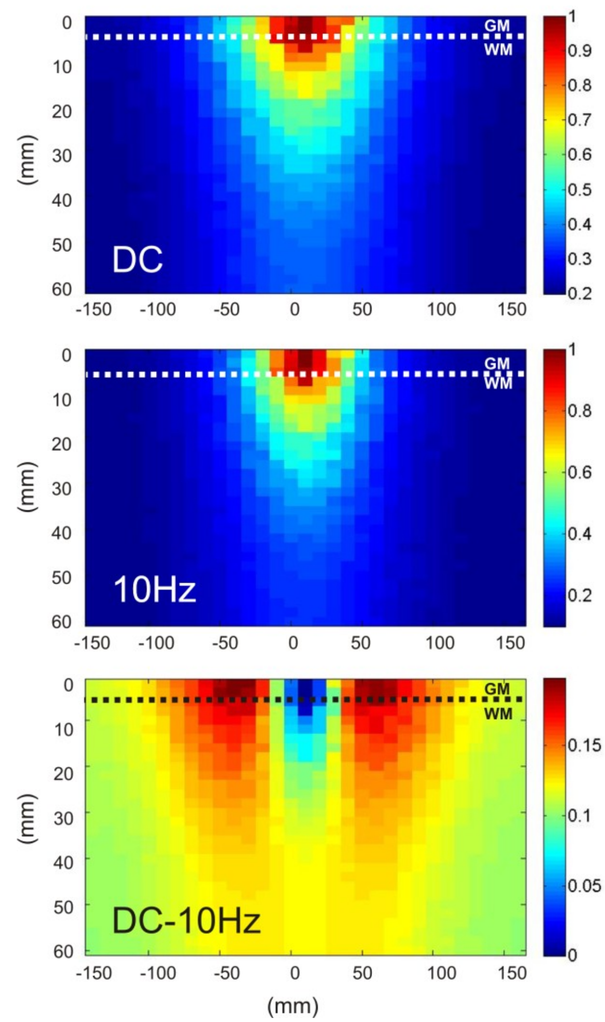

Fig. 2. Distribution of the tangential electric fields at the central cross section between the electrodes. Fields are normalized to the maximal value in the corresponding plane. Depth zero corresponds to the upper boundary of the grey matter layer.

B. Fields Penetration

Fig. 3. compares the maximal magnitudes of the electric fields and current densities at different planes across the

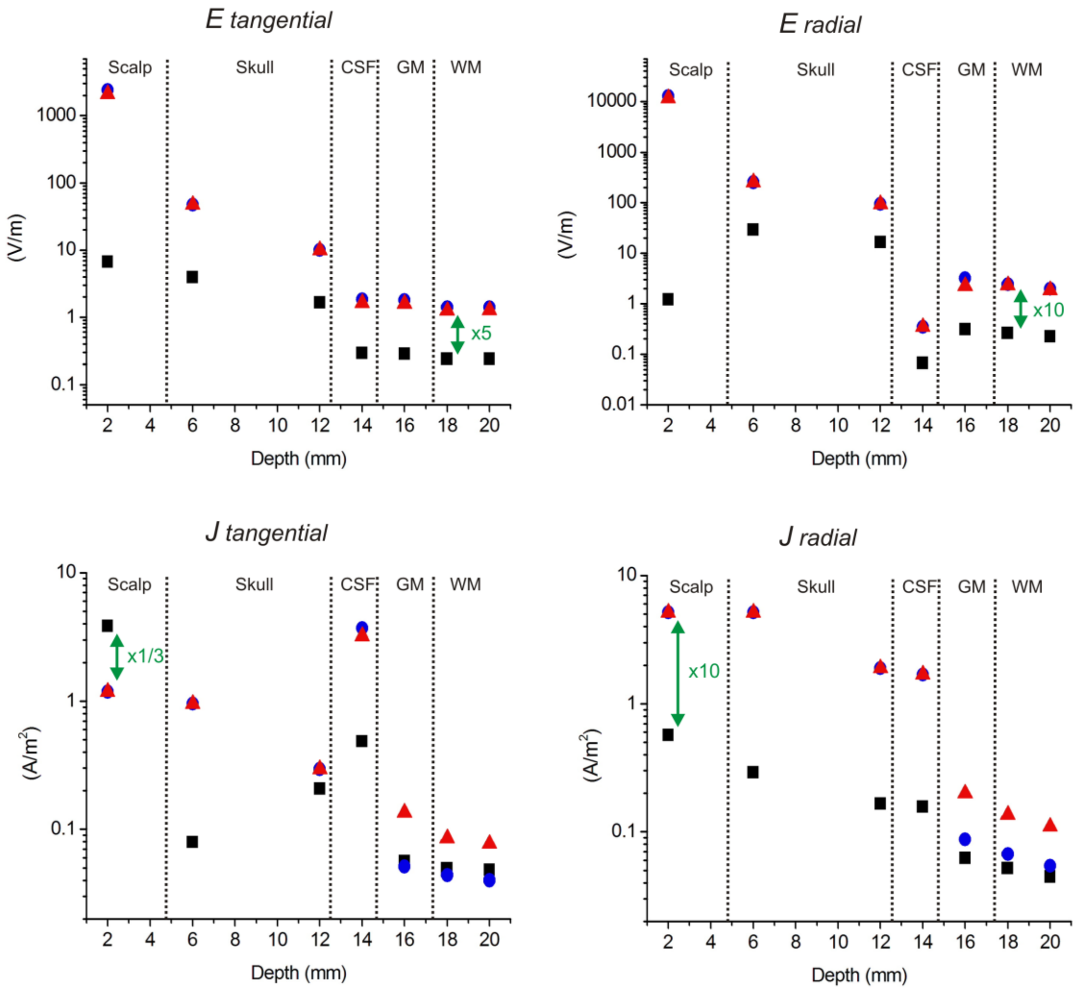

Fig. 3. Maximal electric field (E) and current density (J) as a function of penetration depth. black square- DC, blue circle- 10Hz, red triangle- 100Hz. GM-grey matter, WM- white matter.

tissue depth. In the grey matter layer, the maximal magnitude of the radial component of the electric field is 0.3V/m, 3.1V/m and 2.2V/m in the case of 0Hz, 10Hz and 100Hz, respectively, and the maximal magnitude of the tangential component of the field is 0.3V/m, 1.8V/m and 1.6V/m in the case of 0Hz, 10Hz and 100Hz, respectively. Thus, in comparison to tDCs, low frequency tACs generates fields in the layer of the grey matter that are 10 times larger in the radial direction and 6 times larger in the tangential direction.

The improved penetration efficiency of the stimulation is fundamentally due to a difference in the tissues conductivities. The conductivity of the scalp at low frequency AC is four orders of magnitude smaller than the conductivity in DC, σ_{scalp}(AC) << σ_{scalp}(DC). As a consequence, at low frequency AC stimulation, a smaller amount of current flows between the electrodes ($J_t = \sigma \cdot E_t$) at the scalp, see Fig. 4. The low conductivity of the scalp at tACs reduces as well the drop in conductivity at the scalp-skull interface by approximately three orders of magnitude, which minimizes the accumulation of surface charge between the layers ($\nabla J_r = \rho$ where ρ is free charge density). The resulted radial currents at the skull layer are almost two orders of magnitude larger in the case of tACs in comparison to tDCs. The conductivity of the CSF layer at low frequency AC is almost similar to DC (× 1.2). This highly conductive layer introduces little losses in both tACs and tDCs to the radial currents from the skull before reaching the grey matter.

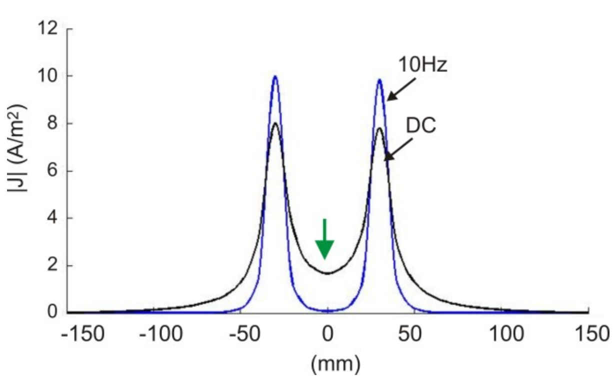

Fig. 4. Current density at the scalp layer. Green arrow highlights the current difference at the inter-electrode space.

The conductivity of the grey matter is just × 1.3 larger at 10Hz in comparison to DC, resulting in overall larger cerebral electric fields. In comparison to 10Hz, the grey matter has × 3.25 larger conductivity at 100Hz, resulting in a correspondent reduction in the magnitude of the fields at that frequency.

IV. Discussion

Different head models have been used in the past, ranging from simple infinite half-planes and perfect spheres (e.g. Datta et al. [1]) to patient-specific accurate models based on magnetic resonance imaging (MRI) with idiosyncratic details of the individual's brain anatomy. Similarly, the conductivity details vary from a homogenous model with tabulated values for the major tissues (e.g. Dmochowski et al. [3]), to detailed values of 40 different tissues (e.g. Parazzini et al. [7]). Each type of modelling approach has its pros and cons. Models with simplified geometries provide vital insight on generic trends and principles, though cannot be translated directly for guiding clinical applications. In contrast, realistic head models have been instrumental for clinical diagnostic and therapy, however they are computational expensive and often do not easily reveal the underlying biophysical principles.

The head model used in this study had isotropic conductivities. However, the tissues making the human head are anisotropic. For example white matter fibre tracts are anisotropically resistive, with resistivity values varying as much as one order of magnitude with direction of current flow [4]. Holdefer et al. [4] show that the results obtained with or without anisotropies in conductivity vary considerably in topography and magnitude. Thus, future work should address the impact of anisotropic conductivity on the cerebral fields.

The reactance of the brain tissues at 10Hz are five orders of magnitudes smaller than the resistance (the reactance increases approximately linearly with frequency). The same order of magnitude difference between the reactance and the resistance exists at the scalp and skull layers. At the CSF the difference is only two orders of magnitudes however it does not increase much up to kilohertz frequency range. Thus, it is a valid simplification to neglect the reactance component of the tissues when analysing the magnitude of the induced tACs fields.

The internal path of the current flow is affected by the boundaries conditions at the tissue interfaces, which enforce a continuity of the normal component of the current density and the tangential component of the electric fields. The boundary conditions are sensitive to the particular values of the tissue conductivity. In this study, we used values based on the characterization of Gabriel et al. [10]. The basic findings presented in this study are in agreement with a study we have done in a realistic head model [11].

V. Conclusions

This paper investigated the penetration and focallity of cerebral fields induced by transcanial alternating (AC) current stimulation. It was shown that, in comparison to time invariant stimulation (i.e. DC), AC stimulation generates cerebral fields that are upto 10 larger and 20 percent more focused. It was argued that the basis for this effect is the reduced conductivity of the scalp which minimizes the current shunting before propagating to deeper layers. The outcomes of this study may help designing tACs protocols and interpret experimental results.

Acknowledgment

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