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Theoretical Investigation of Transcranial Alternating Current Stimulation using Realistic Head Model

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***Abstract*— Transcranial alternating current stimulation (tACs) is an important new technique that allows to modulate non-invasively high-order cortical processes. The underlying mechanisms of activation of this brain stimulation technique are still poorly understood. Herein, we use a finite difference time domain (FDTD) technique to investigate the penetration and focality of tACs in comparison to a time invariant (DC) stimulation. We show that stimulation using 10Hz generates cerebral fields that are larger (×2.5) and more focused than DC stimulation and that faster oscillating stimuli of 100Hz and 1000Hz, generate smaller and less focused cerebral fields than 10Hz. The outcomes of this study may help tACs users to design better protocols and interpret experimental results.**

I. INTRODUCTION

Non-invasive neuromodulation techniques based on a cranial application of weak currents have been gaining an increasing interest in the last few years [1-3]. The revitalization in the field has been driven by the escalating number of patients who suffer from incurable neuropsychiatric diseases and was boosted by the more recent success of a non-invasive stimulation technique based on magnetic excitation, transcranial magnetic stimulation (TMS).

Weak transcranial electrical stimulations typically utilize a battery-powered current generator device, capable of delivering a controlled electrical current of up to 2 mA. The stimulating currents are applied to the scalp via electrodes that are placed on the head. In comparison to TMS, weak transcranial electrical stimulation requires only a fraction of the power and hence it is simpler and cheaper to build. However, in contrast to TMS, the induced cerebral fields are not sufficient to evoke action potential response, instead, the activation mechanism, in the case of non-oscillating direct current stimulation (tDCs), is associated with an accumulative sub-threshold change in the neural rest potential which then impacts the mean spiking rate at that region [3].

The mechanism of action in the case of oscillating alternating current stimulation (tACs) is less understood. Antal *et al* [4] reported a lack of sustainable change in the excitability of the motor cortex (measured by TMS and EEG indexing), while Schroeder *et al* [5] reported an increase in low frequencies EEG signals post tACs protocols. There have been suggestions that the mechanisms by which tACs renders its neural response is associated with a modulation of the neurotransmission release [6], an interference with an on-going cortical activity [7], or a secondary effect via an excitation of the peripheral nerves [3].

Although the mechanism of action in the case of tACs is not clear, the cumulative outcomes of the weak transcranial electrical stimulation studies suggest that the capacity to modulate high-order cortical processes is strongly dependent on the waveform, duration and montage by which the currents are applied.

The need to better understand the underlying routes of action of weak transcranial electrical stimulations, led to a recent theoretical effort to simulate the cerebral distribution of the fields during the stimulation process (e.g. [8]-[11]). Different head models were used, ranging from simple infinite half-planes and perfect spheres to patient-specific accurate models based on magnetic resonance imaging (MRI). The focus of these studies was a constant DC stimulation.

The aim of this study is to explore via simulation the impact of the tACs frequency on the penetration and focality of the induced cerebral fields. This study uses a realistic human head model and, published in conjunction with a complementary laminar model [15], aims at improving our understanding of this important nascent field.

II. MATERIALS AND METHODS

A. Numerical human model

In the present study we used a numerical model from the Virtual Family project [12]. In particular, we used the model of 'Ella' (26-year-old female, 1.63m height, 58.7kg weight, 22kg/m^2 BMI) at a resolution of 1mm. In this head model, 40 tissues were distinguished during the segmentation. On the top surface of the head we placed two cylindrical electrodes of 5mm radius and 3mm height, both modeled as conductors with the conductivity of solid copper (5.8×10^7 S/m). The electrodes were placed at the C_Z and F_Z points of the 10/20 EEG international system, with the help of the image processing toolbox of MATLAB. The whole surface of the electrodes was in touch with the skin of the model.

We solved the problem of transcranial current stimulation with the above electrodes at four frequencies (0Hz, 10Hz, 100Hz, and 1000Hz), i.e. tDCS and tACS. For each of the frequencies studied the corresponding dielectric properties

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for the head tissues were taken from the work of Gabriel *et al* [14], except for the skin tissue, which was modeled as a weighted average of the skin and the subcutaneous adipose tissue (SAT), following the suggestion by Parazzini *et al* [8]. Another exception was at the frequency of 0Hz, where, for the dielectric properties of skin, skull, cerebrospinal fluid (CSF), grey matter and white matter, we used the values proposed by Datta *et al* [9]; to the rest of the tissues we assigned the dielectric properties of Gabriel *et al* [13] at the frequency of 10Hz.

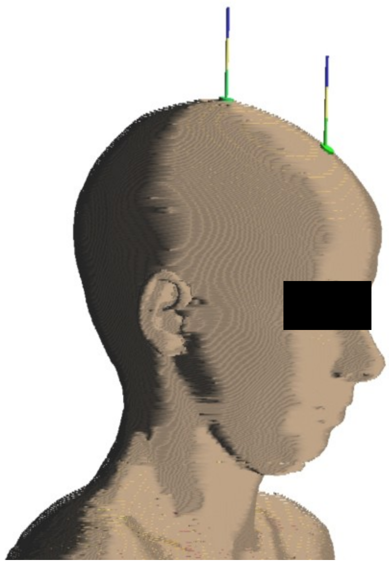

Figure 1: The electrode configuration investigated with the numerical female model of Ella.

B. Numerical technique

In order to solve the electromagnetic problem we used the low frequency solver of the commercially available software package SEMCAD-X (Schmid and Partner Engineering AG, Zurich, Switzerland). We employed the mode of 'stationary currents' available in the software, i.e., we assumed for all frequencies that conductive currents dominate with respect to displacement currents ($\sigma \gg \omega\varepsilon$) [14]. In this quasi-static approximation, it is necessary to solve the Laplace equation and determine the electric potential (φ) distribution inside the human head model:

$$\nabla(\sigma \nabla\varphi) = 0 \qquad (1)$$

where σ is the electric conductivity of each tissue (S/m). For all simulations the current was set at 1mA. The anodal electrode was placed at point C_Z and the cathodal one at F_Z. At the outer boundaries of the computational domain we assumed the Dirichlet boundary of grounding ($\varphi = 0$ V), whereas the lower boundary was set to a homogeneous Neumann condition (insulation) to separate the head from the rest of the body. An integrity check was performed in the model by comparing the values of the anodal and cathodal currents, which should be the same; in our simulations the difference between the two currents was less than 10%. In order to study the effect of polarity during tDCS, we performed simulations by interchanging the anode and the cathode at the electrode positions.

III. RESULTS

A. Fields Penetration

Figure 2 shows the distribution of the normalized value of electric field on the sagittal plane that goes through the electrodes. It indicates strong fields at the upper layers underneath the electrodes and lower fields that propagate to deeper brain structures.

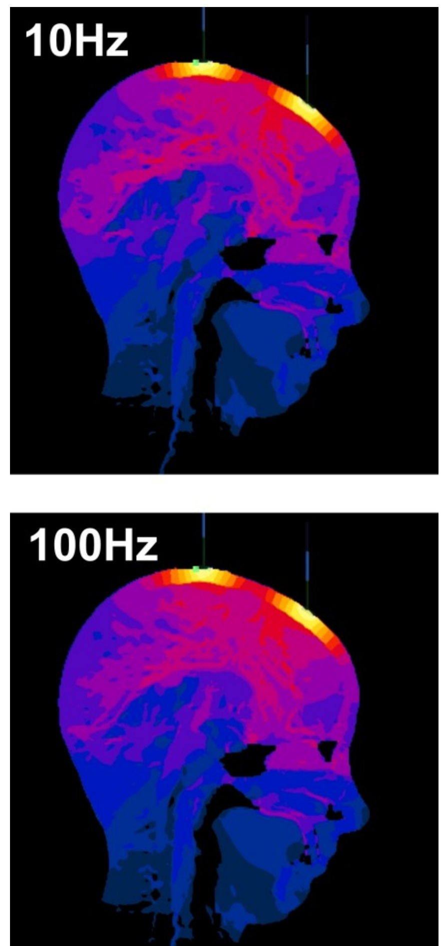

Figure 2: Distribution of the normalized value of electric field (dB) on the sagittal plane that goes through the electrodes.

In order enable a better comparison between the different frequencies, we plotted the cumulative distribution function (CDF) of the electric fields, Figure 3. At the grey matter, the maximal amplitudes of the induced fields were 0.5V/m (0.25V/m 90% CDF), 1.2V/m (0.57V/m 90% CDF), 1V/m (0.4V/m 90% CDF) during DC, 10Hz, 100Hz and 1000Hz, respectively. The cortical fields induced by 10Hz stimulation were approximately 2.5 times larger than fields that were induced by DC stimulation. Both 100Hz and 1000Hz resulted in slightly lower cortical fields than 10Hz. At the white matter, all fields were ~20% smaller, however the relative differences between the frequencies were preserved. At deeper layers such as the cerebellum, the magnitude of the electric fields is ~80% smaller than on the grey matter. Here, there was almost no difference between the DC and 10Hz stimulation, both slightly higher than 100Hz and 1000Hz.

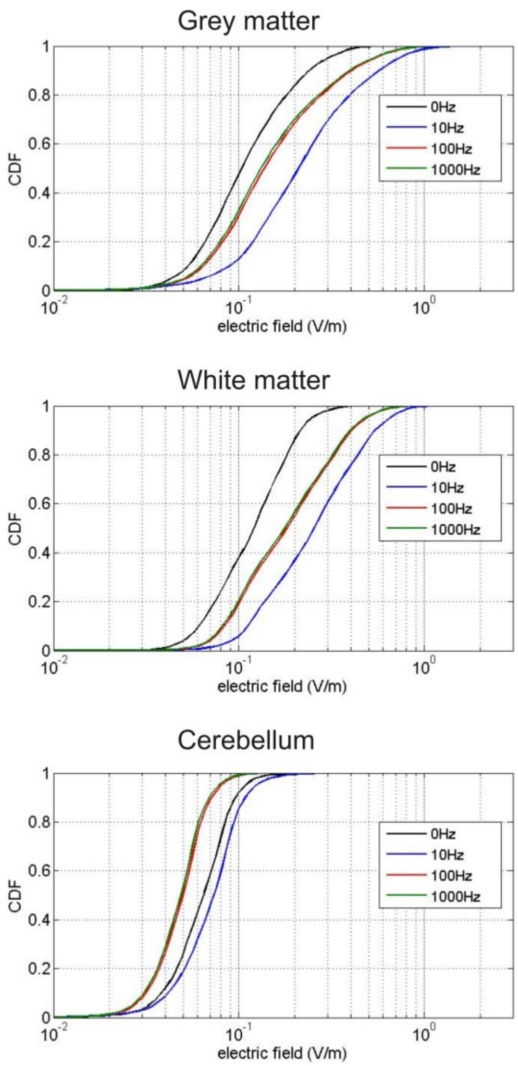

Figure 3. Cumulative distribution functions for grey and white matter and all three frequencies investigated for tACS.

B. *Fields Focality*

Figure 4 shows the normalized distributions of the peak electric field on the grey matter layer. The distributions are normalized to the maximum value of each and are in logarithmic scale, because the dynamic range of the induced field is very wide. It is clear from the results that at the higher frequencies of 100 and 1000Hz there are no significant differences in the field distribution.

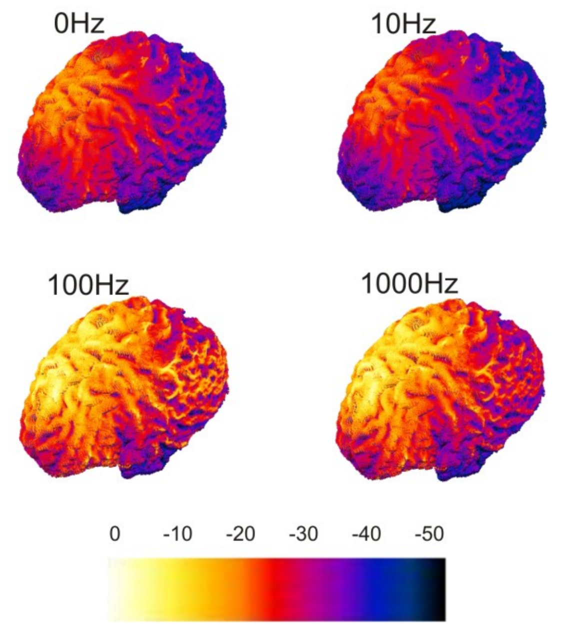

Figure 4. Normalized distribution of the peak induced electric field on the grey matter of the brain at various frequencies.

In all cases the area with the highest values is below the cathodal electrode at F_Z. The distribution of the cortical fields is more focused in the case of 10Hz stimulation in comparison to tDCs. However, at higher tACs frequencies (100Hz and 1000Hz) the excitation became more disperse.

IV. Discussion

This paper investigated the cerebral fields induced by oscillating weak transcranial current stimulation. A summary of the results is given in Figure 5, where the descriptive statistical values of the electric field distributions for three brain tissues are shown. We showed that stimulation using 10Hz generates cerebral fields that are larger (×2.5) and more focused than DC stimulation. Faster oscillating stimuli of 100Hz and 1000Hz generate smaller and less focused cerebral fields than 10Hz. The flow of the stimulating currents is affected by the boundaries conditions at the tissue interfaces, which ensure 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 electrical permittivity and conductivity of the tissues. In this paper we used the values that are based on Gabriel et al. characterization work [13], with an amendment to the scalp values as was proposed by Parazzini *et al* [8] and at 0Hz as was proposed by Datta *et al* [9].

This study uses particular size and location of the electrodes however the physical principles that underlie the model are well in agreement with other models. The findings presented here agree with a study we have done on a simplified laminar model [15], which proposed that the basis

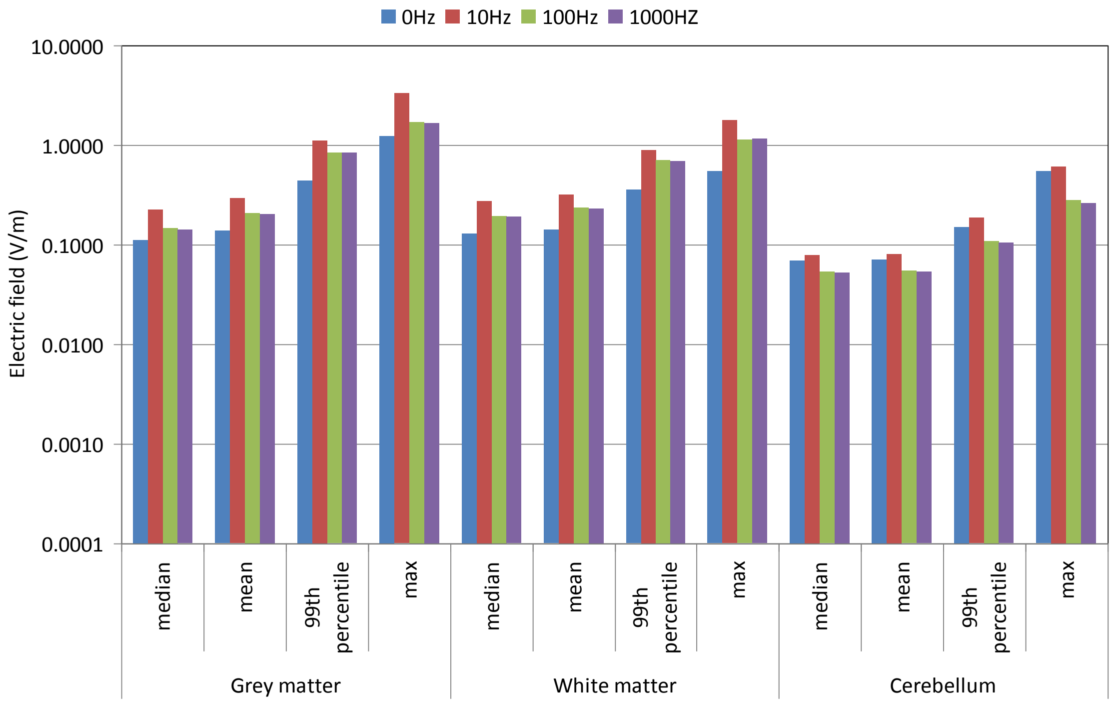

Figure 5. Descriptive statistical values for the distribution of the induced electric field in different brain tissues.

for this differences is the reduced conductivity of the scalp at AC which minimizes the current shunting before propagating to deeper layers. The simulation is done in FDTD which will allow future investigations of the time varying nature of the cerebral field distribution. We hope that the outcomes of this study may help designing stimulating protocols and interpret experimental results.

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