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STUDENT PAPER

Modeling accuracy of transcranial current stimulation: Static and quasi-static approximations errors

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Transcranial current stimulation is a non-invasive brain stimulation technique producing electric fields in the brain using scalp electrodes. Modeling of electric field induced by transcranial current stimulation has been studied with static, quasi-static (QS) approximations and the full set of Maxwell's equations (i.e., full-wave approach, FW) in order to quantify the error due to these approximations. This comparison helps to choose an appropriate approximation depending on stimulation modalities. The QS approximation shall be preferred to the static one, especially for the field analysis in high-permittivity tissues. The results also show that the error is less than 1% below 10 MHz when comparing the QS approximation to FW modelling.

Introduction

Transcranial current stimulation (tCS) is a rapidly growing non-invasive brain stimulation technique, which benefits from a low cost and compact technology, a high safety profile, and even the possibility to use it at home with appropriate devices. tCS can be delivered either using direct (tDCS) or alternative (tACS) current, with tACS being more versatile due to its capability to target frequency-specific brain oscillations. tCS has shown benefits in several neurological disorders and diseases, such as depression [Bennabi and Haffen, 2018], stroke [Boggio et al., 2008; Hummel et al., 2005], Parkinson's disease [Lee et al., 2019], and can even be used to enhance cortical function in healthy subjects [Nissim et al., 2019]. This regain in popularity began in the 2000s, with the evidence that tCS increases cortical excitability [Nitsche and Paulus, 2000]. Subsequent studies have shown evidence of a direct depolarization of the neuronal membrane induced by the exogenous electric field, proportional to the field magnitude itself [Bikson et

al., 2004; Radman et al., 2007; Radman et al., 2009; Rahman et al., 2013]. This has motivated numerous dosimetry studies, modeling the electric field at the cortical scale to predict and optimize stimulation effects [Datta et al., 2009; Grossman et al., 2017; Ruffini et al., 2013; Ruffini et al., 2014].

One common assumption in tCS dosimetry modeling studies is the quasi-static (QS) approximation, since stimulation uses almost exclusively either direct or alternative currents below the kilohertz range. In addition, the capacitive effect of tissues is often neglected (static approximation) considering tissues as perfect ohmic conductors [Plonsey and Heppner, 1967]. Considering the Kramers–Kronig relation, strong frequency dispersion, and high permittivities for biological materials [Gabriel et al., 1996], the validity of this assumption shall be quantitatively evaluated. The QS, however, should remain a valid assumption since the size of the head is negligible as compared to the wavelength in the kHz-range. The error introduced by the QS approximation using deep brain stimulation (an invasive brain stimulation modality based on the use of intracranial electrodes) has been quantified between 5 and 13% [Bossetti et al., 2008]. These results were obtained in the time domain on the electric potential. It is known that the approximation error can become significant when the input signals contain higher frequency harmonics. Quantifying this error could assist in choosing the most appropriate model formulation for accurate tCS simulation, especially using higher frequency interferential methods [Grossman et al., 2017].

Methods

tACS modeling was implemented in a geometry consisting of a slice from a segmented magnetic resonance imaging (MRI) scan of the icbm152 template [Fonov et al., 2011]. Segmentation was performed using the ROAST routine [Huang et al., 2018], which resulted in five segmented domains representing the head tissues: grey matter, white matter, cerebrospinal fluid, skull, and skin. The obtained transverse slice was vectorized (using the Inkscape vector graphics software), and imported into COMSOL Multiphysics for further analysis. This 2D formulation was invariant by translation along the z axis, that does not represent the real case of a human head which has finite dimension in the third axis and its geometry strongly varies from slice to slice. However, this model enables the quantification of the relative modeling error of different approximations, which is roughly a function of the ratio of the model dimensions in the x–y plane to the wavelength $\lambda(\varepsilon)$ [Plonsey and Heppner, 1967].

The dielectric dispersion of each tissue was defined according to the Cole–Cole model [Gabriel et al., 1996]. Note that the resulting low-frequency conductivities of this model can noticeably differ from the common values used for tCS dosimetry modeling [Ruffini et al., 2013; Wagner et al., 2004] and must be used with caution [Gabriel et al., 2009]. However, using this model enables i) accounting for the frequency dispersion, and ii) quantifying the relative error of static and QS approaches that depend mainly on the real part of the permittivity ε_r , according to the complex conductivity $\sigma_c(f) = \sigma(f) + j\omega\varepsilon_r(f)$, where σ is the real electric conductivity. The Laplace equation $\nabla \cdot (\sigma_c \nabla V) = 0$ (where V is the electric potential and σ_c the complex conductivity) was solved for the static (tissue properties given by the real conductivity σ only) and QS approximations. The following boundary conditions were assigned: electric insulation ($\mathbf{j} \cdot \mathbf{n} = 0$) on the skin surface, ground ($V = 0$ V) on one electrode and the terminal current on the other ($\int \mathbf{j} \cdot d\mathbf{S} = I_0 = 2$ mA). The full coupling between the electric and magnetic field is computed by solving the Maxwell–Ampère’s equation for the potentials as $\nabla \times (\mu^{-1} \nabla \times \mathbf{A}) + j\omega\sigma_c \mathbf{A} + \sigma_c \nabla V = 0$ supplied with the same boundary conditions on the potential in addition to the magnetic insulation ($\mathbf{A} \times \mathbf{n} = 0$) on all surfaces. This was the stabilized-low frequency full-wave formulation (FW), which is supplemented by a second equation obtained by taking the divergence of the Maxwell–Ampère’s law [Zhao and Fu, 2017]. Finally, the relative error was calculated according to $|E_{FW} - E_{QS}|/E_{QS}$ or $|E_{\text{other}} - E_{\text{stat}}|/E_{\text{stat}}$.

Results and Discussion

The results in Figure 2a present the ratio between the displacement current J_d and conduction current J_c , which are strongly tissue- and frequency-dependent. Moreover, for tissues such as grey and white matters, the high permittivity in the low-frequency range induces a higher contribution of the displacement current. The QS approach (using complex conductivity) minimizes the error, as shown by the comparison between the QS and static approaches in Figure 2b. Furthermore, the relative error between the static and QS approaches is more significant than between the FW and QS. This highlights the importance of proper knowledge regarding the tissue dielectric properties, and taking into account the capacitive effect through complex conductivity. Finally, the relative error of the QS approximation over the full set of Maxwell’s equations was underneath 1% up to the MHz range, which is not common in tCS. Although, the higher frequency harmonics of non-sinusoidal signals should also be considered. Finally, Figure 3 illustrates the electric field distributions for each approach at 10 Hz, 1 kHz, and 1 MHz. The distribution computed using the static approximation differs significantly from QS and FW. The electric field in vicinity to the electrodes decreased with the frequency, due to the conductivity increasing with frequency [Gabriel et al., 1996].

Conclusion

This study addressed the quantification of the error induced by the quasi-static and static approximations, in comparison to the generalized full-wave method. Our results show that the relative error remains significantly below 1% at low frequency (up to 10 KHz), which is the case for most tCS applications, and increase to 1% at 10.5 MHz. It also reveals that, depending on the dielectric properties of the model, the error is significantly larger between the static (i.e., medium properties are given only by the real conductivity) and quasi-static approximation.

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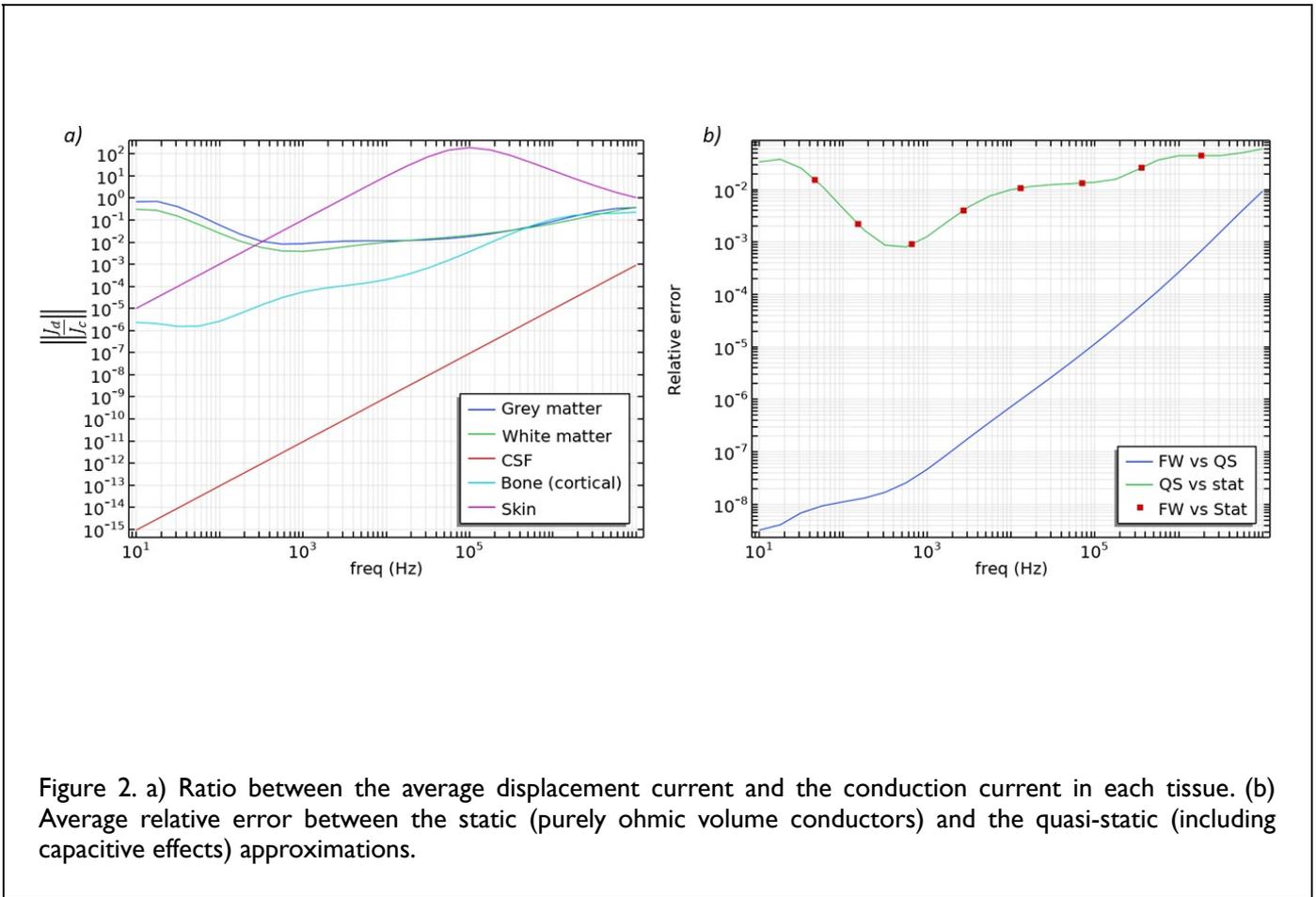
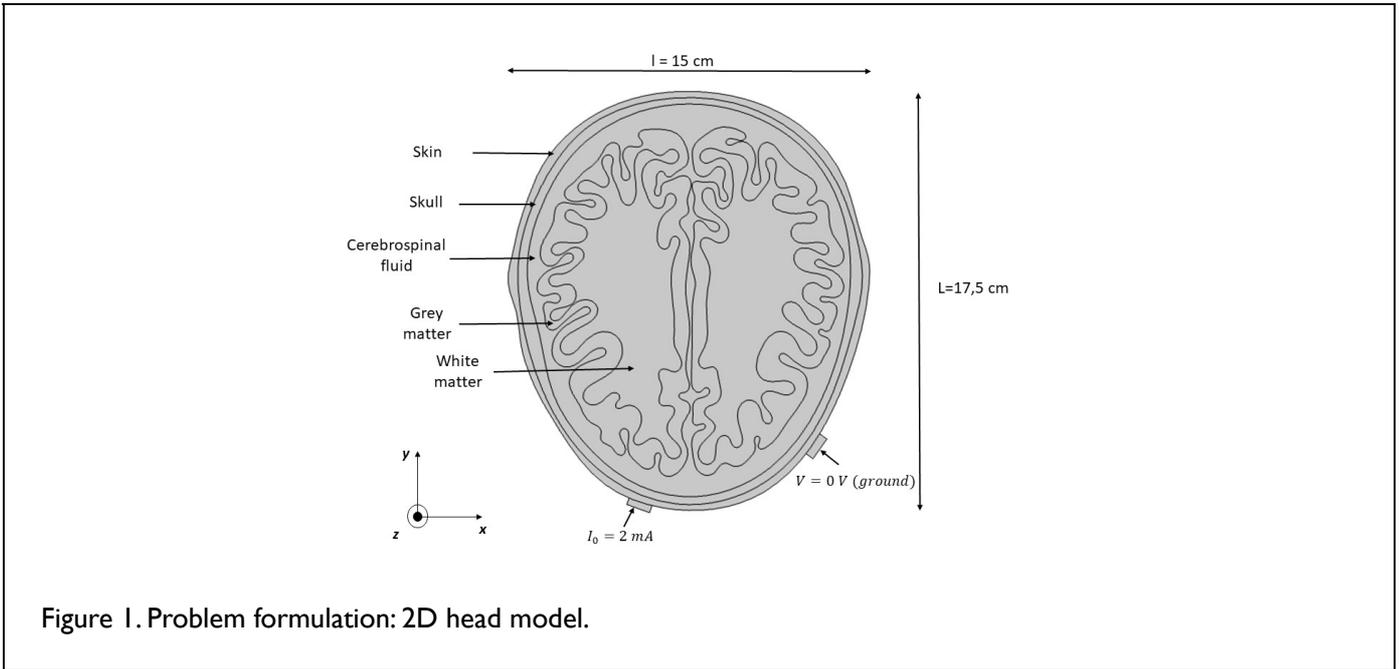
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Figures



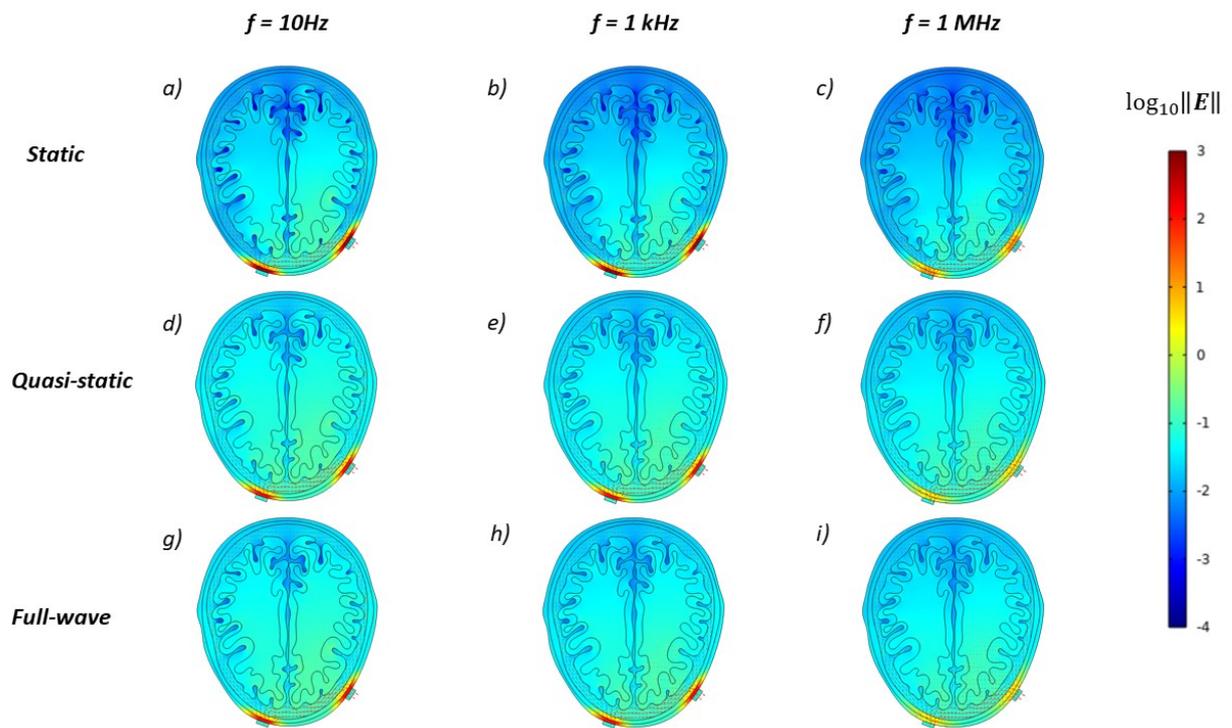


Figure 3. Electric field distributions induced by the current stimulation for (a–c) static approximation, (d–f) quasi-static approximation, and (g–i) full-wave approach. Considered stimulation frequencies: (a, d, g) 10 Hz (typical tACS frequency range), (b, e, h) 1 kHz (prospective tACS range), and (c, f, i) 1 MHz (estimation of higher-order harmonics).