Finite Element Study of Transcranial Direct Current Stimulation: customization of models and montages

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Finite Element Study of transcranial Direct Current Stimulation: customization of models and montages

Thesis
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Abstract
Transcranial Direct Current Stimulation (tDCS) is a non-invasive neuromodulation technique that applies low amplitude current via electrodes placed on the scalp. Rather than directly eliciting a neuronal response, tDCS is believed to modulate excitability – encouraging or suppressing activity in regions of the brain depending on the polarity of stimulation. The particular application of tDCS is often determined by the electrode configuration and intensity of stimulation. MRI-derived finite element models have been developed to analyze the effect of these parameters allowing novel electrode configurations to be tested in subject specific models. By creating a subject specific model of an obese subject, the effect of fat on tDCS was examined. The inclusion of fat into the model led to an increase in cortical electric field intensity. To further investigate the influence of fat the conductivity was varied from that of skull to that of skin. Cortical electric field intensity did not change monotonically with fat conductivity. It was postulated that this may be due to a shunting effect both when the shell of fat surrounding the skull is too resistive for penetration and when the fat is so conductive as to lead current around rather than through the head. The effect of electrode positioning was then examined in a new 2x1 Hybrid montage utilizing both HD electrodes and sponge pads. Systematically varying the location of both the anode and cathode led to changes in the electric field distribution. This is in contrast to the old heuristic convention of placing the “active” electrode over a region of interest and neglecting the influence of the “return” electrode. Lastly the radial directionality of electric field was examined in a 4x1 ring configuration. Previous models have predicted the spatial focality of the 4x1 ring configuration. Polarity specificity, the ability to selectively apply either anodal or cathodal stimulation, was demonstrated in a 4x1 montage over the motor strip. The customization of models for specific populations and montages provides new avenues for clinical practice.
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1. Introduction to transcranial Direct Current Modeling

**tDCS Background**

Transcranial Direct Current Stimulation (tDCS) is a non-invasive neuromodulation technique that applies low amplitude current via electrodes positioned on the scalp. The conventional tDCS electrode configuration utilizes large (5x7 cm) saline soaked sponge pads with 1 anode and 1 cathode (1x1). In recent years, High Definition (HD)-tDCS has been evaluated as a more focal electrode configuration utilizing small (12mm) electrodes with conductive gel in a 4x1 ring configuration [1], [2]. It is believed that by applying either anodal or cathodal stimulation endogenous brain activity can be encouraged or suppressed [3–5]. For example, subthreshold membrane potential oscillations that would not elicit an action potential could be brought to threshold with subtle membrane depolarization through extracellular stimulation. Past research into extracellular field effects have described peripheral nerve depolarization a function of the second difference in extracellular voltage potential \( \frac{d^2v}{dx^2} \), also known as the activating function. In the case of cortex field effects, electric field itself \( \frac{dv}{dx} \) could be assumed to be representative of modulation given that the induced electric field is uniform on the neuronal scale and neuronal modulation is correlated with uniform electric field magnitude [1], [6–8]. This, however, does not take into account the orientation between the neuron and the electric field. Recent modeling studies have used radial electric field as a representation of the directionality, and the polarity, of stimulation [9].

The clinical applications of tDCS span a wide range of conditions. Stimulation protocols have been explored as potential treatment options for deleterious conditions as well as for performance enhancement. Examples of clinical applications include depression, addiction, motor rehabilitation, neuropathic pain, memory enhancement, among others. The particular
electrode configuration, the montage, varies from application to application [10–16]. Different montages were initially explored empirically. Stimulating over a target region with a return placed elsewhere (contralateral shoulder or contralateral supraorbital) became a general heuristic rule.

**Modeling Background**

Recent clinical trials, however, have been guided by high resolution MRI-derived models [9], [17], [18]. These MRI-derived Finite Element (FE) models currently in use have evolved from previous models with simplified geometries. Initially, concentric sphere models were developed which could examine the role of various electrode configurations such as the 4x1 ring [19], [20]. Wagner et al. begin incorporating MRI-derived human geometry [21]. Gyri-precise modeling was then developed by Datta et al [1]. Finite Element models with anisotropic skull and white matter has been modeled as have models with sub-cortical structures [22–24].

The models used in this particular study are MRI-derived finite element models with subject specific anatomy. High resolution MRIs were segmented into different tissue/material masks of varying conductivities through a combination of automated and manual tools. Computer generated models of electrodes, gel, and/or sponge pads were incorporated into the segmentation. Volume meshes were generated, boundary conditions were applied, and the Laplace equation (\( \nabla \cdot (\sigma \nabla V) = 0 \)) was solved. The resulting cortical electric field was interpreted as a correlate for stimulation and modulation.
2. Obesity Modeling Parameters

**Background**
Disorders such as depression and chronic pain have been ameliorated by tDCS in a clinical setting [10], [25]. Studies have suggested the cravings associated with smoking and alcohol can be reduced [11], [12]. There is additional evidence that tDCS reduced the cravings for certain foods [26]. There is thus rationale for exploring tDCS in obese subjects.

However, a specific complication exists in treating obese subjects with tDCS. As a noninvasive technique, current delivery to the brain during tDCS is subject to the conductivities of all tissues that surround the brain. This includes the relatively low conductivity of fat. Electrical penetration into the brain – current flow through the skin, fat, and skull – may thus be an issue. Finite Element (FE) models are standard tools to predict brain current flow during electrical stimulation (“forward” model) but must be parameterized accurately.

Magnetic Resonance Imaging (MRI) derived FE models have been utilized in the past to predict the flow of current in the brain [1], [27]. An individualized patient specific model has also been created in the case of stroke [28]. The effect of fat in a normal head has been modeled [27], [29]. In particular, one of these papers found profound differences in the current density of skin and skull with the addition of fat [29]. The cortical current density was altered as well, but to a lesser extent (Relative Difference Measured: 5.0 %). This, however, was modeled in a normal head.

The efficacy of tDCS has been demonstrated in a range of individuals [4]–[8], but efficacy in an obese individual remains unknown. This modeling study is intended to serve as a preliminary analysis that will lead to an optimized FE model of obese heads undergoing tDCS. In the future this model can be applied to optimize tDCS electrode montage to deliver current to specific brain targets, such as those associated with appetite suppression.
Methods
Anatomical MRI scans were produced from a 3T Philips Achieva scanner for a thirty-five year old female with a Body Mass Index (BMI) of 53.5. The MRI scans were T1 weighted using an MP-RAGE (magnetization-prepared rapid acquisition with gradient echo) sequence, which produced high resolution scans with a spatial resolution of 1x1x1.2mm. From this data, tissues of interest were segmented. Large 5x7 cm sponge pads and electrodes were modeled and added to the segmentation, the segmentation was meshed, and the mesh was solved.

Segmentation and Mesh Generation
There were 7 tissues of interest to be segmented from the MRI scan: skin, fat, bone, cerebral spinal fluid (CSF), gray matter, white matter, and air. This was initially accomplished using an automated segmentation algorithm contained in Statistical Parametric Mapping (SPM8) software. Additional post-processing was applied via an in-house algorithm programmed in MATLAB (2010b, The MathWorks, MA) to correct for errors in continuity. Additional detail, however, remained to be segmented. The gyri and sulci needed to be resolved in greater detail, and fat was not included at all in the automated segmentation algorithms. Additional manual segmentation of the brain was necessary to complete the model. This was accomplished using ScanIP+FE (SIMPLEWARE LTD., UK). An initial segmentation of fat was generated through use of a thresholding flood fill algorithm. The segmentation data, which was originally sampled like the MRI scan at 1x1x1.2mm per voxel, was resampled to 1x1x1mm per voxel and smoothed. Additional close filters were applied to repair rough patches of fat at the base of head and neck.

Figure 1: The segmentation of homogenous skin and heterogeneous skin are contrasted in (a) and (b). A quarter of the model was cut away for visualization purposes only. Images of the fat, skull, CSF, and gray matter segmentation are shown in (c-f) respectively.
Large 5x7 sponge pads and electrodes were created in a computer aided design (CAD) program (SOLIDWORKS, DS SolidWorks, MA). A rectangle with a slight curve was sketched to approximate the cross-sectional dimensions of a sponge pad roughly 10mm thick. This slight curvature was drawn to reflect the curvature of the scalp at the placement site. Another curve was sketched in an orthogonal plane along which the cross-sectional profile of the pad was swept. This process was repeated for the corresponding electrode and was repeated for a second set of pads and electrodes with curvatures that matched the second placement site.

The pad and electrode pairs were then imported into ScanCAD (SIMPLEWARE LTD., UK) alongside the segmentation model as a Standard Tessellation Language (STL) file. The pads were then placed according to a possible montage, F8 active with the return over the contralateral supraorbital [13]. Once these CAD models were in place, the models were converted to segmentation masks and exported back into ScanIP+FE for meshing.

An adaptive tetrahedral meshing algorithm within ScanIP+FE was used to mesh the models. The initial model with fat segmented had 7 tissue masks in addition to the electrodes and pads. This model meshed at approximately 11 million quadratic elements with about 15 million degrees of freedom. The second model with the fat mask merged into the skin managed to mesh at approximately 6 million quadratic elements and about 8 million degrees of freedom.

**Finite Element Model**

A FE model based on electrostatic volume conductor physics was created in COMSOL Multiphysics 3.5a (COMSOL, Inc., MA). Each mesh was imported into this FE solver and isotropic conductivities (in S/m) were assigned as follows: skin: 0.465, fat: 0.025, skull: 0.01, csf: 1.65, gray matter: 0.276, white matter: 0.126, air: 1e-15, sponge pad: 1.4, gel: 0.3, electrode:
5.99e7 [9]–[12], [14]. Additional models were run using a range of conductivities for fat. These values (in S/m) are 0.0125, 0.07, 0.125, and 0.250.

Boundary conditions were applied as electrically insulated to all exterior boundaries and continuous to all interior boundaries. The exterior boundaries of the electrodes were altered to be 1A/m2 of inward current injection for the active electrode and ground (V=0) for the return electrode. For the active electrode, 1A/m2 corresponded to an inward current injection of about 4.38mA in the homogeneous skin model and 4.43mA in the heterogeneous skin (skin and fat) model. The model was then solved to a relative tolerance of 1e-6.

After solving, boundary plots of the cortical surface (gray matter) were plotted with a false color map and scaled to a visible range. This scale was then normalized to be per 1 mA of current injection. Additional lighting was used in some images to better visualize brain morphology and the spatial distribution of electric field.

**Results**

Fat represented a large proportion of what would normally be modeled as skin. As seen in Fig. 1 (a-c), the addition of fat thins the skin greatly – to just a few millimeters in some areas such as the forehead. The other tissue masks were segmented in the same manner as a non-obese head. This generated fairly typical looking skull, CSF, gray matter, and white matter tissue mask as seen in Fig. 1 (d-f).
The results of the homogeneous skin condition were contrasted to the heterogeneous skin condition. In Fig. 2 (A.1-A.3), peak electric field is plotted on the same scale. An apparent difference can be seen between the two conditions. The inclusion of fat leads to greater electric field peaks than in the model without fat. The scale for Fig. 2 (A.3) is adjusted to show electric field peaks in the homogenous condition. While the locations of the peaks are similar, the magnitudes differ greatly. The maximum peaks plotted in the heterogeneous condition are at 0.36 V/m per mA, while the maximum peaks in the homogeneous condition are only 0.23 V/m per mA. This is an increase of close to 60%. Significant shifts in spatial targeting are also apparent, including electric field peaks in the medial orbitofrontal cortex (OFC).

In Fig. 2 (B.1-B.8) the results of the cortical electric field due to varying fat conductivity are displayed from left to right in order of increasing fat conductivity. A surprising trend is seen in which cortical electric field intensity increases from (B.1) to (B.3) before the intensity again diminishes in (B.8). Coincidentally, the most commonly used value for fat conductivity (0.025
S/m; as reported in literature) may be near the optimal range for current penetration in this particular model. A possible explanation for this inflection in cortical electric field intensity may be a shunting effect through the skin. At the low extreme in (B.1), the shell of fat that surrounds the skull is too resistive for much current to penetrate into the brain. As the conductivity is increased there is an “optimum” at which current can pass into the brain. But if the conductivity is increased further as in (B.8), current again shunts around the skull.

This concept of current shunting through soft tissue can be used to explain the results in parts (A.1-A.3). The increased current penetration in the heterogeneous model could be explained by a
reduction in skin volume. In the homogeneous model, more current may shunt through the scalp instead of penetrating the more resistive skull. The soft tissue commonly modeled with the conductivity of skin is essentially wedged between the surrounding air and the skull – both of which are extremely low in conductivity. Skin is relatively conductive compared to skull, air, and fat. It is modeled with a value of 0.465 S/m in contrast to 0.01, 1e-15, and 0.025 S/m for skull, air, and fat respectively. Replacing much of the skin for fat may lead to a dramatic reduction in the conduction through the skin. Indeed, this concept of a “preferential pathway” through skin was postulated by Shahid [29] after a similar effect was observed in a normal head model with fat. The effect, however, appears to be magnified in an obese model in which the inclusion of fat leads to an increase of nearly 60% in peak electric field. From these results, fat should not be neglected and should be precisely parameterized in an accurate model of an obese head.

**Conclusion**

This modeling study provides the first indication of current flow through the head of an obese subject during tDCS and considers general modeling methodology for such cases. As with any modeling effort, addition details (e.g. muscle mask, DTI) can be further considered, but our results indicate that precise consideration of fat anatomy and properties is essential for accurate predictions.
3. Prefrontal Cortex Stimulation via Combined HD and Conventional Electrodes

Background
Transcranial Direct Current Stimulation (tDCS) has been and is being used in a broad spectrum of experiments, ranging from basic cognitive research to clinical trials. tDCS involves passage of low-intensity current through electrodes on the scalp to produce weak electric fields in the brain that lead to neuromodulation and plasticity. The flexibility of tDCS and its customization to a broad range of applications stems from the ability to shape the flow of brain current by selecting the electrode montage. One region in particular, the prefrontal cortex (PFC), has been implicated in affecting a range of normal brain processes and pathology, and so has been nominally targeted in tDCS research. Clinical studies have suggested conditions such as depression, alcohol cravings, and working memory can benefit from tDCS of the PFC [10], [12], [15].

The selection of a tDCS montage for PFC stimulation has typically followed a basic “rule-of-thumb” approach using conventional tDCS sponge electrodes (~25 cm² pads) where the active electrode (anode or cathode) is placed over PFC and the return (cathode or anode) over of another brain region. Yet, several modeling studies have shown that brain current flow during tDCS may be complex and idiosyncratic. Depending on electrode montage, peak brain current flow may in fact be between rather than under the electrodes [1], [19], [21], [27]. The position of the return electrode will influence overall current flow including under the active electrode [30]. Therefore, the influence of the return electrode cannot be selectively ignored.

Attempts to focalize tDCS include decreasing the size of the active sponge electrode, while increasing the size of the return electrode [31], [32]. The use of smaller High-Definition (HD)
electrodes [33] has allowed stimulation with multiple electrodes in optimized configurations [34] including the 4x1-Ring [1]. Here, we simulate PFC current flow generated using a combination of two frontal HD electrodes and one conventional return electrode. Our goal was to understand and optimize bilaterally symmetric PFC and general brain current flow using this relatively simple to implement “2x1-Hybrid” configuration.

**Methods**

Finite Element (FE) models were generated and solved for a variety of montages based on the International 10-10 system for electroencephalogram (EEG) electrode placement. In each of these montages, two HD electrodes are active and a single 5x5 pad is the return. Specifically, the montages modeled were: F3 and F4 active, neck return; F3 and F4 active, Pz return; F3 and F4 active, Oz/POz return; AF3 and AF4 active, Oz/POz return; Fp1 and Fp2 active, Oz/POz return; Fp1 and Fp2 active, Cz return. These models were derived from the same T1 magnetic resonance imaging (MRI) scan with a spatial resolution of 1x1x1 mm of a healthy adult male. These models followed electrostatic volume conductor assumptions. With the exception of electrode positioning, a common method was shared by the models that can be delineated into two primary phases: the model construction and the model solution.

**Segmentation and Meshing**

The geometry of the volume conductors were originally derived from a high resolution 3T MRI scan (1mm3). Initially, the MRI scan was automatically segmented using Statistical Parametric Mapping (SPM8) software. This automated segmentation software was used to delineate six different tissues within the MRI scan: skin, bone, cerebral spinal fluid (CSF), gray matter, white matter, and air. The automatic segmentation; however, was not perfect. Spatial aliasing problems existed in thin layers such as the CSF and in parts of the gray matter. This required manual
correction to patch small holes as well as to resolve additional detail in anatomical features such as the cortical surface. Additional smoothing was applied to the surface of skin via Gaussian filters. This manual correction, filtering, and subsequent volume meshing was performed using ScanIP+Fe (SIMPLEWARE LTD., UK).

Prior to meshing; however, the stimulation electrodes, pads, and gel had to be modeled, imported into the segmentation model, and positioned upon the head. This was accomplished using a variety of tools, starting with the computer aided design (CAD) program, Solidworks (DS SolidWorks, MA). The sponge pad was created by sketching a curved rectangular profile in one plane and sweeping this sketch in an orthogonal plane. This was done to create a pad that resembles a conventional 5x5 cm pad with a thickness of about 1 cm. This process was then repeated to create a 5x5 cm electrode that would fit directly on top of the aforementioned pad. The HD electrodes and gels were created in similar fashion by extruding a 4 mm radius circle to form a 4 mm thick disk. These CAD models were then exported as a Standard Tessellation Language (STL) file and imported into ScanCAD (SIMPLEWARE LTD., UK) along with the segmentation model. Here, the CAD models were placed upon the head and converted into a segmentation mask.

The following “2x1-Hybrid” montages were evaluated:

- Montage A: HD electrodes at F3 and F4, return pad centered on the neck.
- Montage B: HD electrodes at F3 and F4, return pad centered at Pz
- Montage C: HD electrodes at F3 and F4, return pad centered between Oz/POz.
- Montage D: HD electrodes at AF3 and AF4, return pad centered between Oz/POz.
- Montage E: HD electrodes at Fp1 and Fp2, return pad centered between Oz/POz.
- Montage F: HD electrodes at Fp1 and Fp2, return pad centered at Cz.
The HD electrodes always have the same polarity, and the return the opposite polarity. Because of the linearity of the solution, our results can be applied for either the HD-anode/pad-cathode case or HD-cathode/pad-anode case (only electric field magnitude, not current direction, is represented). The use of “active” to describe the HD electrodes and “return” to describe the pad is thus arbitrary with current flow across the whole brain. Similarly, because of linearity the results can be extrapolated to any current intensity of DC/low-frequency AC waveform.

The completed segmentation model – head, pads, and electrodes – was then meshed in ScanIP+Fe using the adaptive tetrahedral meshing algorithm. This produced meshes with approximately 9 million quadratic elements, which correspond with about 12 million degrees of freedom.

**Finite Element Model**

The meshes were then imported into an FE solver (COMSOL Multiphysics 3.5a, COMSOL Inc., MA). Within the FE solver, isotropic conductivities were assigned to each subdomain – to each tissue, pad, and electrode within the mesh. These conductivities (in S/m) were assigned as follows: skin: 0.465, skull: 0.01, CSF: 1.65, gray matter: 0.276, white matter: 0.126, air: 1e-15, sponge pad: 1.4, gel: 0.3, electrode: 5.99e7. [4], [13]

Boundary conditions were then applied to the model. The surfaces of the model that were exposed to the surrounding air were assumed to be insulated. This included the surfaces at the base of the neck and shoulders where the model was truncated. Exceptions to this were the exposed surfaces of the electrodes. The surfaces of the HD electrodes were assigned to have an inward current of 1A/m2 each. For 2 HD electrodes, this corresponds to 2 A/m2. Taking into account the area of the exposed surfaces, this corresponds to a total current injection of about 4.9e -4 A. The return electrode was also assign a separate boundary condition; it was assigned
the condition of ground, i.e. $V=0$. All other boundaries, namely all the internal boundaries, were set as continuous.

The FE models were solved to a relative tolerance of $1e^{-6}$. The results were plotted as false color images of the electric field of the cortical surface. Like previous tDCS modeling studies [4]–[7], it is believed that membrane depolarization can be elicited when electric field peaks coincide with axon terminals or bends [14]. Based on this assumption, electric field intensity was the chosen metric for stimulation.
Figure 3: Electric field generated during tDCS using six hybrid (HD plus conventional electrode) montages. From left to right, the columns contain images of the electrode placement on the skin, peak electric field viewed from the right hemisphere, peak electric field viewed over the dorsal surface, peak electric field viewed from the left hemisphere with additional lighting to display morphology, and a coronal slice under F3-F4. The dashed red box represents montages in which the active electrode is fixed. The dashed blue box represents the montages in which the return electrode is fixed. Montages A, B, C, D, E, and F represent the following: F3 and F4 active, neck return; F3 and F4 active, Pz return; F3 and F4 active, Oz/POz return; AF3 and AF4 active, Oz/POz return; Fp1 and Fp2 active, Oz/POz return; Fp1 and Fp2 active, Cz return.
**Results**
The location of electric field peaks varied significantly with both the position of the active and passive electrodes. By manipulating both sets of electrodes, current flow can be directed to or away from certain regions. In Fig. 3, six different hybrid montages are presented. The position of the bipolar HD electrodes at F3 and F4 are held fixed in Montages A, B, and C as the position of the return electrode is varied (dashed red line). In Montages C, D, and E the position of the return electrode is held constant at Oz/POz (dashed blue line) as the position of the bipolar HD electrodes is varied. It can be seen in each set that current flow is modulated by the combined position of the HD electrodes and return electrode.

In each montage, the resulting complex pattern of cortical current flow can be understood by consulting the figures. But several features are notable. For example, the current slips underneath the brain in Montage A as current flows towards the neck pad. Moving the return pad higher, more superior, as in Montages B and C, leads to electric field peaks on the dorsal side of the cortex.

Position of the return on Oz/POz produced significant current across occipital cortex for all HD electrode positions tested, but the overall current flow across the brain, including PFC, is different. In Montage C, peak electric field does not appear directly under the active electrodes, but rather appears between the active and the return. However, this skewing effect appears to be reduced as the active electrodes are moved further inferior and consequently further away from each other as seen in Montages D and E. In fact, in Montage E peak electric field is not between the active and return, but is rather underneath the electrodes. Electric field intensity is substantially higher in this montage and had to be plotted to a different scale. Plots of Montage E at the same scale as the other montages are included in Fig. 3 as well.
An important point, which illustrates the limits of rule-of-thumb montage design, is that effective montages can be designed in which electrode placement is not necessarily directly over the area of interest. In Montage F, the active electrodes are placed inferior of the dorsal lateral prefrontal cortex (DLPC) atop Fp1 and Fp2, while the return is placed nearby, posterior of the DLPC atop the vertex position Cz. It can be seen that peak electric field is neither directly under Fp1 and Fp2 or Cz; rather, peak electric field occurs between the electrodes reaching more of the DLPC and less of the orbitofrontal cortex.

**Conclusion**

There is no “magic bullet” for specific modulation of only PFC; rather each montage results in specific patterns of current flow across PFC and other cortical regions. Ultimately, the most suitable montage will depend on the clinical study objectives. Having a greater variety of possible montages will allow for greater flexibility in tailoring the stimulation prescription to match these clinical needs. The 2x1-Hybrid montages evaluated here present additional alternatives for “rational” tDCS design that is still relatively straightforward to implement. Specifically: 1) Two “HD” electrodes [11] can be positioned on the forehead using a conventional EEG cap or even, for below hairline positions, adhesives; 2) The return electrode is a conventional pad positioned using a cap or straps; 3) Electrodes can be energized using a conventional 1x1 tDCS stimulator with a passive split to the HD pair, assuming reasonable impedance matching, or active control (2x1).

More complex platforms make use of additional HD electrodes for targeted [1] and automatically optimized [34] configurations. The 2x1 Hybrid represents a middle ground between traditional pads, with poor targeting, and multichannel electrode arrays, which require specialized software and hardware.
As common for modeling studies, the representation of “neuromodulation intensity” is assumed to reflect local electric field (Quasi-Uniform assumption), though consideration of directionality or explicit neuron modeling may provide additional insight. Interestingly, the use of two “lint” supra-orbital active electrodes (with an extracephalic return) dates to early clinical studies of electrosleep and cranial electrostimulation [35–37].
4. Radial Directionality of 4x1 Stimulation

Background
Past models have demonstrated the spatial focality of the 4x1 ring configuration [1], [18], [20]. However, another factor to consider is the polarity of stimulation, anodal or cathodal. Past electrophysiology research has indicated electric field to be representative of neuromodulation provided that the electric field is uniform on the neuronal scale and electric field intensity is correlated with neuron modulation [1], [6–8]. The orientation of neurons with respect to the electric field is been neglected in a simple magnitude plot of field intensity. Polarity specific effects such as anodal versus cathodal stimulation are not considered despite research indicating differing clinical results [15], [38], [39]. By considering radial electric field, the component of electric field perpendicular to the cortex, polarity specific focality can be modeled in unconventional montages. Polarity specific stimulation can then be used as another factor in montage selection.

Methods
A finite element model simulating 4x1 stimulation over C3 was generated based on previously described protocols [40], [41]. A 3-D 1mm isotropic T1 MRI of an adult male was segmented into 20 different head regions using a combination of automated and manual techniques. These 20 regions were then assigned one of seven possible conductivities: skin, fat, skull, cerebral spinal fluid, gray matter, white matter, or air. Electrodes with conductive gel were modeled to resemble a 4x1 montage with a radius of approximately 75mm. This corresponded to a center electrode at the 10-20 system position C3 and surround electrodes at F3, Cz, P3, and T3. An inward current density of 1A/m² was applied to the surface of the center electrode; ground was applied to the surrounds. Cortical electric field magnitude and radial electric field was then calculated and scaled for 2mA and -2mA of stimulation.
Figure 4: Results from finite element brain model. Values are given for a 4x1 HD-tDCS montage with a 75mm radius and intensity of 2mA. Peak cortical electric field is circumscribed within the ring electrodes. Anodal and cathodal stimulation have symmetric spatial distributions with reversed polarity. Further analysis shows stimulation due to both the center (active) and outer (return) electrodes; however, there are 70% more outer-polarity nodes at low intensity while peak intensity is nearly 100% center-polarity”. a) Electric field magnitude (top row); b) Position of HD electrodes (center row, left); c) Radial electric field: anodal (center row, middle) and cathodal (center row, right); d) Radial polarity magnitude: distribution comparison (bottom row, left), active component (bottom row, middle), return component (bottom row, right).
**Results and Conclusion**

Consistent with previous models, peak cortical electric field was restricted to cortical regions circumscribed by the ring electrodes, with local clustering based on brain idiosyncratic anatomy (Fig. 4, Top Row) [1]. The polarity of stimulation was nominally set by the center electrode, and modeling of current flow in and out of the cortex confirms that, qualitatively, anode center stimulation produces dominantly inward current inside the ring, while cathode center stimulation produces dominantly outward current inside the ring (Fig. 4, Center Row). However, the biophysics dictates that current flowing into the cortex needs to flow out. Because anode-center and cathode-center produce symmetrical current flow, we simply consider current flow “associated with” the center electrode (radial inward for anode-center and radial outward for cathode-center) verse current flow associated with the outer electrode (Fig. 4, Bottom Row). Using this representation, qualitatively, inside the ring current is consistent with the center polarity while, especially on the outer-banks of gyri walls and under the ring electrodes, some current flow consistent with the outer polarity is evident. Comparing the percentage of center-electrode polarity nodes to outer-electrode polarity nodes over the range of radial electric field values reveals a quantitative difference in the intensity distribution. There are more outer polarity nodes with a low electric field, whereas center polarity nodes are about 30% more abundant at peak electric field. Thus, while the total inward/outward cortical current must be balanced, using the 4x1 HD-tDCS montage produces a concentration of center-dominated current flow under this electrode, and distributes the outer-electrode current flow. This FEM model suggests there is a degree of both spatial and polarity focality afforded by the HD-tDCS 4x1 montage.
**Conclusion**

MRI-derived finite element models of tDCS have become a tool capable of guiding clinical practice. Models can be customized to individual subjects as well as specific novel montages. In the case of abnormal populations such as obesity, customization can lead to more precise models. Still, further customization of models requires accurate parameterization. Variations in tissue conductivities can affect cortical electric field magnitude in a non-monotonic manner. The modeling of custom montages allows researchers to explore new configurations that do not necessarily follow simple heuristic rules. This reinforces the need to model novel configurations prior to clinical application to ensure the desired focality and intensity. Radial directionality, the polarity of stimulation, is another parameter to consider. While the total current entering and leaving an enclosed volume such as the brain must be neutral due to the conservation of current (a condition enforce by the model), the density of that inward and outward current flow can vary. In the case of the 4x1 ring configuration, the polarity of simulation can be concentrated allowing for focal, unidirectional stimulation of a target. Careful consideration of these parameters in model and montage design can allow for new clinical populations and methods.
References


