Simultaneous High-Definition Transcranial Direct Current Stimulation of the Motor Cortex and Motor Imagery

Bryan S. Baxter, *EMBS Student Member,* Bradley Edelman, *IEEE Student Member*, Xiaotong Zhang, *IEEE Member*, Abhrajeet Roy, and Bin He, *IEEE Fellow*

***Abstract*— Transcranial direct current stimulation (tDCS) has been used to affect the excitability of neurons within the cerebral cortex. Improvements in motor learning have been found in multiple studies when tDCS was applied to the motor cortex during or before task learning is performed. The application of tDCS to motor imagery, a cognitive task showing activation in similar areas to motor execution, has resulted in differing effects based on the amplitude and duration of stimulation. We utilize high definition tDCS, a more spatially localized version of tDCS, to investigate the effect of anodal stimulation on human motor imagery performance. In parallel, we model this stimulation using a finite element model to calculate stimulation area and electrical field amplitude within the brain in the motor cortex and non-stimulated frontal and parietal regions. Overall, we found a delayed increase in resting baseline power 30 minutes post stimulation in both the right and left sensorimotor cortices which resulted in an increase in event-related desynchronization.**

I. Introduction

Transcranial Direct Current Stimulation (tDCS) is a non-invasive neuromodulation technique during which a low level of current is applied to the scalp and passes through the brain [1]. It can be used to safely modulate neural activity [2]. Based on in vitro and computational modeling studies, tDCS either depolarizes or hyperpolarizes the membrane of neurons based on the polarity of the electrode but does not directly induce action potentials [2]. Traditionally, it has been found that cortex beneath the anode has an increased motor evoked potential (MEP) amplitude, while that under the cathode has a decreased MEP amplitude. This modulation, as evaluated by motor evoked potentials, has been found to last up to an hour following stimulation [3] and learning improvements for cognitive tasks can remain up to 6 months post stimulation.

Motor Imagery (MI) is a cognitive task consisting of imagining a motor movement but not performing the execution of the movement. The performance of this task generates an event related desynchronization (ERD) of oscillatory activity in the hemisphere contralateral to the imaginary movement. Event related synchronization (ERS) can occur in the ipsilateral hemisphere during this imagination [4]. The ERD and ERS generally occur in the mu (8-13Hz) and/or Beta (15-30Hz) bands in electrodes over the sensorimotor cortex.

Previous work combining MI and tDCS has found stronger ERD in response to MI in healthy [5] subjects. Anodal tDCS has been reported to increase event-related desynchronization over the stimulated motor cortex during BCI performance following 15 minutes of 1mA sponge electrode stimulation, but this did not result in an increase in performance [6]. Lapenta and colleagues [7], using a similar setup but with 2mA for 20 minutes, found the opposite effect; that anodal stimulation decreases the ERD in the same hemisphere as stimulation for both motor imagery and during motor observation. These conflicting data suggest the effect of tDCS on motor imagery needs to be investigated further. In these works tDCS was applied with $35cm^2$ sponge electrodes localized on opposing hemispheres of the brain, primarily over motor cortex and the contralateral orbit, to stimulate large areas of the cortex. This yields an effect that is difficult to localize and results in widespread distribution of injected current throughout the brain. Current flow induced in the brain by tDCS is complex due to tissue types composing the head and tissue geometry, including cortical sulci and gyri and neuron orientation within these macrostructures [8].

With recently developed high-definition tDCS (HD-tDCS) [9] systems, online recording of the EEG during stimulation is possible [10]. This setup also allows a more localized stimulation area, reducing current flow to a limited area within the brain. As tDCS stimulation can increase or decrease performance in a tasks [11] it is important to understand and limit current flow to areas of interest. Here, we utilize high-definition tDCS (HD-tDCS) in order to address this question with the task of motor imagery. In addition, we utilize Finite Element Modeling (FEM) of the tDCS stimulation to evaluate the spread of stimulation and the amplitude of current on the target of stimulation, the primary motor cortex. This work allows us to better understand the effects of focal tDCS stimulation and the ability to target specific motor planning and output networks of the cortex.

II. Methods

A. Experimental Setup

Subjects: 5 subjects (2 female) were recruited to participate in these experiments (Ages: 22-29 years). All procedures and protocols were approved by the University of Minnesota Institutional Review Board.

Hardware Setup: A 64 channel Biosemi EEG cap with active electrodes and ActiveTwo amplifier were used to

This work was supported in part by NSF CBET-1264782, NSF DGE-1069104, NIH R01EB006433, NIH T32EB008389, and ONR N000141110690.

B.S. Baxter, B. Edelman, X. Zhang, and A. Roy are with the Department of Biomedical Engineering, University of Minnesota, Minneapolis, MN 55455, USA.

*B. He is with the Department of Biomedical Engineering and Institute for Engineering in Medicine, University of Minnesota, Minneapolis, MN 55455, USA (e-mail: binhe@umn.edu).

record the EEG signal at 2048 Hz (BioSemi B.V., Amsterdam, Netherlands). A tDCS device with high-definition tDCS adapter (Soterix Medical, Inc. NY, USA) was used to deliver 1.5 mA of current to the center electrode with four return electrodes. Stimulation was performed with the polarity of the center electrode was anodal, with surround as cathodal. The cap was adapted to fit HD-tDCS electrodes adjacent to EEG electrodes. Center electrode was placed directly adjacent to C3 between C3/CP3. Surround electrodes were placed adjacent to CP3 between CP3/P3, C1 between C1/FC1; C5 between C5/FC5, and FC3 between C3/FC3 at a radius of 3.5 cm from the center electrode. Conductive gel was used to reduce electrode offset below 25 mV for EEG electrodes and impedances under 3 kΩ for tDCS electrodes.

B. Experimental Procedure

Subjects were seated comfortably in a chair 90 cm from an LCD monitor where experimental stimuli were displayed. Subjects were instructed to relax, keep their head still, and to blink as little as possible during the trials. BCI2000 was used to present experimental stimuli.

Subjects performed 4 runs of 18 trials of left/right motor imagery, where subjects were instructed to kinesthetically imagine opening and closing their right or left hand for 4 seconds based on the location of the target on the screen. Following this, the tDCS system was turned on and stimulation was started. The total stimulation time was 20 minutes and subject performed 5 runs of 18 trials during the stimulation. Following this, the tDCS device was turned off. The subject then immediately performed 4 runs of 18 trials each. The subject then sat quietly for 15 minutes and following this performed 4 runs of 18 trials each.

C. Signal Processing

Raw data collected with BCI2000 was processed using custom software utilizing EEGLAB [12] in Matlab (The Mathworks, Inc., MA, USA). Data was high pass filtered at 1 Hz and low pass filtered at 110 Hz. The mean of each channel was removed. Bad channels were removed and electrodes were re-referenced to the common average reference. ICA was used to remove eye movement, eye blink and muscle artifacts. Data was epoched into trials and epochs contaminated with noise not removed by ICA were discarded. Data was then downsampled to 256 Hz and transformed using a 1Hz window Morlet Wavelet. Power values were normalized for each subject and each electrode to the pre-stimulation baseline power.

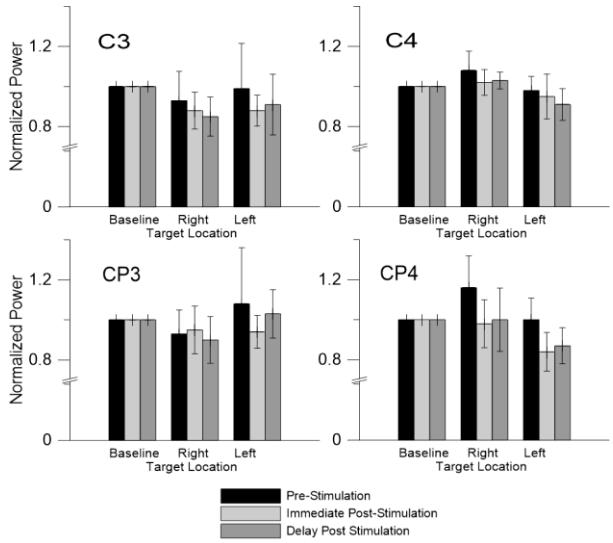

Fig 1: Power normalized to baseline during motor imagery performance for left and right targets during pre-stimulation, immediate post-stimulation, and 30 minutes post-stimulation in the alpha (8-13 Hz) band. Error bars indicate standard deviation.

D. Finite Element Model of HD-tDCS

ANSYS version 14 (ANSYS Inc., PA, USA) was used for 3D modeling and Finite Element Method (FEM) based electromagnetic simulation [13], [14]. The DUKE head model from the Virtual Family [15] was imported and constructed in ANSYS, by a hexahedral element with the mesh size of $2\times2\times2$ mm^3. Head segmentation (19 head tissues) was provided along with the head model, and their corresponding conductivity values were taken from literature [16] and assumed to be isotropic. Electrodes were localized on the scalp surface in the configuration as performed in our experiments with the surround electrodes located 3.5cm from the center electrode. 2.0 mA of current was injected into the center electrode while surround electrodes each had -0.5 mA of current injected.

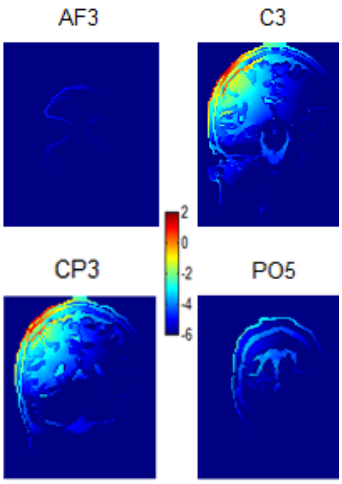

Fig 2: Coronal slices of FEM model at electrodes of interest. Colors indicate the natural log of total current density

III. Results

Stimulation altered event-related power bilaterally in the sensorimotor cortex (Figure 1). Mean alpha increase was 14% above the pre-stimulation condition for C3/CP3 and 24% above the pre-stimulation for C4/CP4 at 30 minutes post-stimulation. C3 had an increased ERD amplitude from a decrease of 7% from baseline to 15% from baseline. C4 also had an increased ERD amplitude from a decrease of 2% to 9% from baseline. The result of this is a mean increase in amplitude from 15% to 18% for the ERD signal during right hand imagery using a standard weighting of C3 vs. C4 electrodes.

The peak current density calculated within the FEM in grey matter under the electrodes of interest in the left (stimulated) hemisphere: C3: 0.161 A/m^2 CP3: 0.132 A/m^2 PO5: 0.007 A/m^2. Peak current density in white matter: C3:

0.492 A/m^2 CP3: 0.356 A/m^2 PO5: 0.005 A/m^2 (Figure 2). EEG results during stimulation in these electrodes displayed an effect suggested by the model. C3/CP3 had an average of 100% increase in EEG power during stimulation; though some of this is likely due current directly flowing to these electrodes. C4/CP4 had a 3% decrease in power during stimulation; supporting the model showing very little current flowing to the contralateral hemisphere. AF3/AF4 and PO5/PO6 showed little change in EEG signal during simulation also illustrated by the model.

IV. Discussion

The change in baseline resting data from pre-stimulation to post-stimulation supports previous work that found increased MEP and global network power post-stimulation [8], [10]. The largest effect was not found directly after stimulation, but at 30 minutes delayed from the end of a 20 minute stimulation session. Motor imagery activity had the largest power change at a thirty minute time delay, paralleling the delayed maximum effect of anodal motor cortex stimulation with HD-tDCS [8]. For both of the imagery conditions, the measured ERD signal amplitude increased, suggesting anodal tDCS could be used to increase signals used to control a brain-computer interface. More subjects are needed to evaluate statistical significance as well as compare these results to sham conditions. The results of the FEM model are similar to previous results using HD-tDCS FEM models, including the peak current density within the brain.

By combining FEM and EEG we have extended previous modeling studies and examined the effect of this localized stimulation on non-stimulated areas of the brain, including the contralateral motor cortex. Our model shows minimal current flowing to the contralateral sensorimotor cortex but there are strong interhemispheric cortico-cortico connections through the corpus callosum where each hemisphere can affect the contralateral hemisphere in an activity dependent manner. In addition, the increase in EEG power in the alpha band in non-stimulated electrodes supports a widespread connectivity specific increase in resting activity following simulation.

References

[1] M. D. Johnson, H. H. Lim, T. I. Netoff, A. T. Connolly, N. Johnson, A. Roy, A. Holt, K. O. Lim, J. R. Carey, J. L. Vitek, and B. He, "Neuromodulation for brain disorders: challenges and opportunities.," *IEEE Trans. Biomed. Eng.*, vol. 60, no. 3, pp. 610–24, Mar. 2013.

[2] T. Wagner, A. Valero-Cabre, and A. Pascual-Leone, "Noninvasive human brain stimulation.," *Annu. Rev. Biomed. Eng.*, vol. 9, pp. 527–65, Jan. 2007.

[3] M. Nitsche and W. Paulus, "Sustained excitability elevations induced by transcranial DC motor cortex stimulation in humans.," *Neurology*, vol. 57, no. 10, pp. 1899–901, Nov. 2001.

[4] G. Pfurtscheller and L. Da Silva, "Event-related EEG/ MEG synchronization and desynchronization: basic principles," *Clin. Neurophysiol.*, vol. 110, pp. 1842–1857, 1999.

[5] J. Matsumoto, T. Fujiwara, O. Takahashi, M. Liu, A. Kimura, and J. Ushiba, "Modulation of mu rhythm desynchronization during motor imagery by transcranial direct current stimulation," *J. Neuroeng. Rehabil.*, vol. 7, no. 1, p. 27, 2010.

[6] P. Wei, W. He, Y. Zhou, and L. Wang, "Performance of motor imagery brain-computer interface based on anodal transcranial direct current stimulation modulation.," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 21, no. 3, pp. 404–15, May 2013.

[7] O. M. Lapenta, L. Minati, F. Fregni, and P. S. Boggio, "Je pense donc je fais: transcranial direct current stimulation modulates brain oscillations associated with motor imagery and movement observation.," *Front. Hum. Neurosci.*, vol. 7, no. June, p. 256, Jan. 2013.

[8] H.-I. Kuo, M. Bikson, A. Datta, P. Minhas, W. Paulus, M.-F. Kuo, and M. a. Nitsche, "Comparing cortical plasticity induced by conventional and high-definition 4 × 1 ring tDCS: A neurophysiological study," *Brain Stimul.*, no. October, pp. 1–5, Oct. 2012.

[9] P. Minhas, V. Bansal, J. Patel, J. S. Ho, J. Diaz, A. Datta, and M. Bikson, "Electrodes for high-definition transcutaneous DC stimulation for applications in drug delivery and electrotherapy, including tDCS.," *J. Neurosci. Methods*, vol. 190, no. 2, pp. 188–97, Jul. 2010.

[10] A. Roy, B. S. Baxter, and B. He, "High-Definition Transcranial Direct Current Stimulation Induces Both Acute and Persistent Changes in Broadband Cortical Synchronization: A Simultaneous tDCS–EEG Study," *IEEE Trans. Biomed. Eng.*, 2014.

[11] T. Iuculano and R. Cohen Kadosh, "The Mental Cost of Cognitive Enhancement," *J. Neurosci.*, vol. 33, no. 10, pp. 4482–4486, Mar. 2013.

[12] A. Delorme and S. Makeig, "EEGLAB: an open source toolbox for analysis of single-trial EEG dynamics including independent component analysis.," *J. Neurosci. Methods*, vol. 134, no. 1, pp. 9–21, Mar. 2004.

[13] X. Zhang, S. Zhu, and B. He, "Imaging Electric Properties of Biological Tissues by RF Field Mapping in MRI," *IEEE Trans. Med. Imaging*, vol. 29, no. 2, pp. 474–481, Feb. 2010.

[14] P. P. Silvester and R. L. Ferrari, *Finite Elements for Electrical Engineers*. Cambridge, UK: Cambridge University Press, 1996.

[15] A. Christ, W. Kainz, E. G. Hahn, K. Honegger, M. Zefferer, E. Neufeld, W. Rascher, R. Janka, W. Bautz, J. Chen, B. Kiefer, P. Schmitt, H.-P. Hollenbach, J. Shen, M. Oberle, D. Szczerba, A. Kam, J. W. Guag, and N. Kuster, "The Virtual Family--development of surface-based anatomical models of two adults and two children for dosimetric simulations.," *Phys. Med. Biol.*, vol. 55, no. 2, pp. N23–38, Jan. 2010.

[16] R. J. Sadleir, T. D. Vannorsdall, D. J. Schretlen, and B. Gordon, "Transcranial direct current stimulation (tDCS) in a realistic head model.," *Neuroimage*, vol. 51, no. 4, pp. 1310–8, Jul. 2010.