Original Research

Effect of Transcranial Direct Current Stimulation on an Individual's Ability to Learn to Control a Brain-Computer Interface

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Published online: 09 October 2019

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Abstract

Brain-computer interfaces (BCI) are systems which enable direct communication between a brain and an external device by translating electrical brain activity into meaningful output. The motor learning period is the time it requires for an individual to learn. This influences the development of BCIs since they are used by individuals with motor impairments. BCIs relying on manipulating the sensorimotor rhythm (SMR) through motor imagery have lengthy learning periods, which present a significant barrier to using this technology. Transcranial direct current stimulation (tDCS) presents an opportunity to improve motor learning periods by facilitating the manipulation of SMR (1). The aim of this study was to validate previous studies that found significantly reduced motor learning period with combined use of tDCS and BCI. Also, to evaluate if the high density system with targeted anodal tDCS had similar effects compared to non-targeted tDCS. Following tDCS, the participants used a SMR-based BCI to move a falling ball to targets that appeared on each side of the screen. The effect of tDCS was assessed by comparing task accuracy and SMR change during motor imagery between groups. The experimental group was significantly more accurate in controlling the BCI than the control group (p = 0.021); however, there was no significant difference between groups in the SMR change upon motor imagery (p = 0.22). Results demonstrate that the high density system with targeted tDCS had similar effects than non-targeted tDCS. Moreover, 1mA of tDCS with the targeted system had similar effects of those to 2mA non-targeted system.

🏷 **Tags**: tDCS, BCI, motor imagery, primary motor cortex, EEG

Introduction

Brain-Computer Interface (BCI)

BCIs are a means of directly connecting brain activity to computer interfaces, allowing individuals to communicate or control computers without having to interact with them through physical movements. BCIs are typically used as communication and control systems (2), predominantly targeted at individuals with minimal communicative abilities due to motor impairments, such as amyotrophic lateral sclerosis (ALS) and muscular dystrophy (2). They have been linked to multiple neuroimaging methods such as electroencephalogram (EEG), magnetoencephalography (MEG), electrocorticography (ECoG), intracortical recording, functional magnetic resonance imaging (fMRI) and near-infrared spectroscopy (NIRS) (3). BCIs using EEG records neurophysiological signals from the brain and translates them into meaningful interactions with the external environment, such as controlling technological devices (4).

Many BCI systems are based upon endogenous brain responses to external stimuli, such as flashing lights (e.g. visual-evoked potential BCIs) or oddball stimuli (e.g. P300-based BCIs). The usability of these systems is high; approximately three-quarters of subjects trained on a P300-based BCI were able to gain a functional use of the system (5).

In contrast, sensorimotor rhythm (SMR)-based BCI systems are based upon user-driven changes in the event-related desynchronization (ERD) and event-related synchronization (ERS) in the sensorimotor brain area. ERD is the decrease in the synchrony of the neurons and ERS is the increase in the synchrony of the neurons (7).

At rest, this brain area produces a characteristic rhythm between 8-12 Hz – often known as the mu-rhythm – through ERS (6). When an individual moves, an ERD is produced, thereby decreasing the amplitude of the mu-rhythm (7). Motor imagery -- imagining a specific movement or sequence of movements, without performing the task itself (8) -- can also produce an ERD (9),(7),(6). SMR-based BCIs track the change in the power of the mu-rhythm and translate this into an interface command, typically the left-right movement of cursor on a computer screen. Controlling a SMR-based BCI requires users to generate ERD and ERS on command (10),(11).

Transcranial Direct Current Stimulation

Anodal high-definition transcranial direct current stimulation (HD-tDCS) is a non-invasive brain stimulation technology used to modulate the excitability of the brain using weak electric polarizing currents of 1-2mA (12). HD-tDCS has high spatial resolution and focalization (13) The effect of tDCS on cortical excitability have been shown to last up to 90 minutes in humans (14). This technology is also a relatively inexpensive tool that can be used at home once the system has been calibrated (15). tDCS is currently being used as a therapeutic treatment of conditions such as stroke, Parkinson's disease, chronic pain conditions, anorexia nervosa, depression (16),(17),(18),(19),(20). In the context of stroke, the application of tDCS over the somatosensory area has been shown to increase accuracy in SMR-BCI tasks (21),(12) as well as the ERD of the mu rhythm (1). tDCS has also been shown to improve the ability to perform motor imagery in both healthy and patient populations (21), (22), (23). In these individuals, tDCS applied to either the primary motor cortex (M1) or the dorsolateral pre-frontal cortex was shown to improve the non-dominant hand writing time and legibility through practice of motor imagery tasks when compared to sham stimulation (24).

The objective of this study was to determine whether anodal tDCS applied to the sensorimotor area over ten training days could improve the speed at which individuals learned a BCI task, in comparison to a sham tDCS group. Our hypothesis was that anodal tDCS over the sensorimotor area would improve an individual's ability to perform motor imagery, as measured by the change in accuracy of the BCI task. Additionally, we hypothesized that the experimental group would show a greater difference in EDS-ERD amplitude compared to the control group.

Methods

Subjects

Nine control participants, 6 females, 3 males, mean age = 24.7 ± 7.1 years and nine experimental participants, 5 females, 4 males, 23.4 ± 3.0 years (sex/gender: p = 0.629; age: p = 0.315) were recruited through posters on Facebook pages. Participants were included if they were: 1) healthy individuals between the ages of 18 and 50; 2) right-handed; and 3) naïve to the BCI task used in the experiment. Participants were excluded if they: 1) had any clinical condition (e.g. ADHD) and/or were taking medications affecting the nervous system and/or the ability to learn; and 2) had any contraindications regarding tDCS (e.g. metal cranial implants).

Study Design

A pre-post single-blind design was used in this study. Participants attended 10 BCI training sessions over a 2-week period, with sessions divided into 2 periods of 5 consecutive days (Figure 1). On the first visit, subjects completed an Information and Consent Form, Safety Questionnaire for tDCS, and a Sleep and Activity Questionnaire.

Participants were assigned to the control or experimental group, ensuring that both groups were similar in age and gender. To deliver the tDCS stimulation and record EEG data, a 128 EEG channel electrode net (Electrical Geodesics Inc., Eugene, Oregon, USA) in combination with a Geodesic Transcranial Electrical Neuromodulation (GTEN) system (Electrical Geodesics Inc., Eugene, Oregon, USA) was used. Prior to each session, electrode impedances were reduced to below 50 kΩ. In addition, six cathodal and six anodal channels over the left and right hemispheres respectively were selected, thus allowing for a high density tDCS targeted system. Subsequently, the SMR-based BCI was calibrated by asking participants to rest for 1 second, and then to imagine squeezing their right hand. This was done every experimental session. The electrode over the left somatosensory cortex that showed the greatest amplitude difference between ERD and ERS was selected for the remainder of the experiment. During the experiment, participants received 22.5 minutes of sham tDCS (control group) or anodal tDCS (experimental group), and then completed the BCI task. Upon completion of the BCI task on the first day, participants from both groups were familiarized with the sensation of the tDCS for 3.5 minutes. In sessions 2-10, for every session, the electrode over the left somatosensory cortex that showed the greatest amplitude difference between ERD and ERS was selected. At the end of each session, 3D electrode locations were recorded using a dual-axis camera (Electrical Geodesics Inc) to ensure similar electrode net placement across sessions.
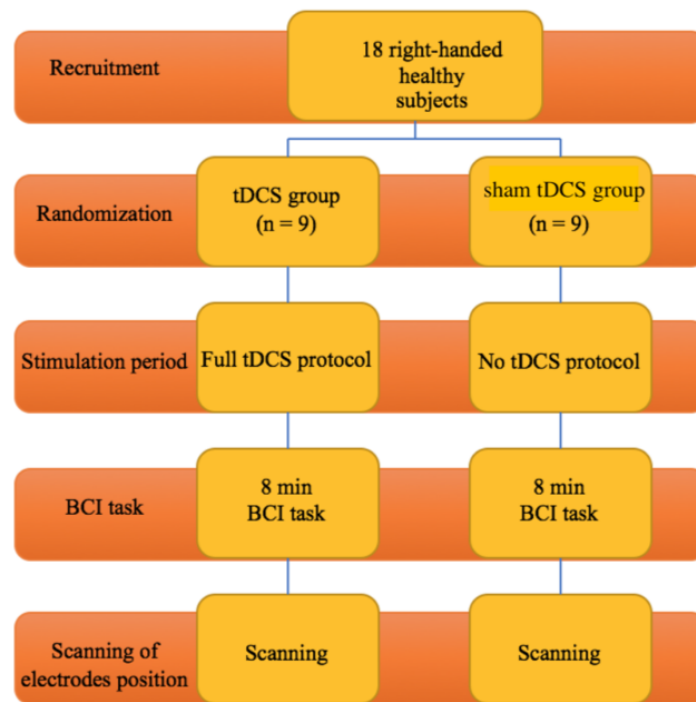

Figure 1. Flowchart outlining the study design. 18 right-handed healthy subjects were recruited. They were randomized into two groups, the tDCS and the sham group. The tDCS group underwent the stimulation protocol while the sham group did not receive stimulation and rested for the same amount of time. Once the stimulation period or the rest period terminated, the subjects completed the BCI tasks followed by the scanning of electrodes position.

BCI Task

Participants were asked to control a small white ball (Figure 2), which appeared at the beginning of each trial at the top-center of their screen, and dropped to the bottom in approximately 9 seconds. Targets covering the left-half or the right-half of the screen

appeared at the bottom indicating which direction the user should direct the ball (Figure 2); participants were instructed to hit the green target. The ball direction was controlled by the total power of the EEG spectrum between 8-14 Hz over the electrode selected during the calibration period. Right hand motor imagery decreased this amplitude, causing the ball to move to the right; relaxation increased this amplitude, causing the ball to move to the left. The overall protocol lasted approximately 8 minutes. The BCI task was loaded through MatLab® (MathWorks Inc., United States) and was composed of 5 blocks of 10 trials each, with a 5-second break in between each block, and a 1-second break between each trial.
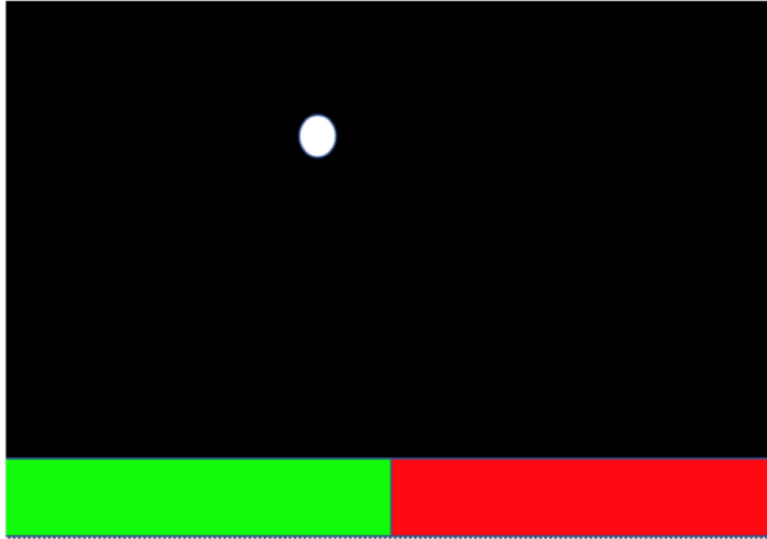
Figure 2. Brain Computer Interface task. The aim of the task was to move the ball towards the green target. In order to move the ball towards the left green target, the subjects had to relax.

Transcranial Direct Current Stimulation Protocol

High-definition tDCS (HD-tDCS) uses smaller electrodes placed in flexible configurations for neuromodulation of focused brain regions. To maximize the accuracy of the stimulation, Finite-Difference-Method (FDM) modelling was used to determine the optimal stimulating electrode configuration for the focal brain stimulation of the right hand representation on the motor cortex (Talairach coordinates: x = -37; y = -21; z = 58). In the software Reciprocity (Electrical Geodesics Inc., Eugene, Oregon, USA), a pre-loaded MRI from a canonical 40 year old male head model was used to generate the forward and inverse models of source-localized activity using FDM and sLORETA (standardized low resolution brain electromagnetic tomography) modelling (25),(26). The principles of Reciprocity are fully described in other publications (27). Briefly, the software relates the graduate of the potential (Ψ) at a spatial point (r) generated by an external source (I) with the surface potential (Φ) generated by a current diploe (d) placed at the same position (r) (28). We set the total target electrodes to 12, yielding 6 possible current injection pairs In. For each pair, the software injects a fraction an of the total targeted stimulation Imax (in our case, 1 mA; such that In = anImax). The software iteratively solves for an across all current injection pairs In, focusing on the electrodes with the maximum ΔΦ. In this way, the 6 anodal and 6 cathodal electrodes are selected, along with the current distribution (an) for each current injection pair (In). The results of the Reciprocity analysis selected the following electrodes on a Hydrocel Geodesic EGI 128-channel net: 6, 7, 13, 30, 31, 106 were selected as anodes, and 40, 41, 46, 47, 50, 51 were selected as cathodes. Lidocaine, a topical numbing agent, was manually applied to the selected 6 anodes and 6 cathodes to decrease the sensation induced by the stimulation. tDCS stimulation ramped up from 0 to 1 mA over 2 minutes,

was maintained at 1 mA for 20 minutes (for the tDCS group only), then ramped-down to 0 mA over 30 seconds (19). In order to maintain blinding, the control group received tDCS during the ramp-up and ramp-down of the protocol only.

Data Processing

EEG processing. EEG was pre-processed in EEGLAB (Swartz Centre of Computational Neuroscience, San Diego, California). All data were bandpass filtered between 1-50 Hz, and notch filtered at 60 Hz. Data was manually cleaned by removing all artifactual channels and all epochs with artifacts, and converted to average reference. Briefly, artifactual channels were defined as those which deviated with respect to the general trend of the majority of the channels or appeared to be flat and not having recorded any data. Artifacts were defined as changes in amplitude or frequency that did not follow the patterns of resting or active EEG. Eye blinks were detected using an Independent Component Analysis (ICA). The eye blinks were removed by selecting the components with high activity in the eyes region of the scalp map projections provided by the program. Once the EEG was cleaned, the data from the electrode showing the greatest ERD was extracted.

Mu Rhythms. A fast Fourier transformation was used to calculate the spectral power of the EEG data from the selected electrode. The frequency with the highest power between 8 and 14 Hz was manually selected, and a window of 2 Hz was imposed on both sides of this frequency. This window was selected empirically after pilot testing to maximize the difference in the area under the curve (AUC) of the power at the peak frequency. The total area under the curve within this selected window was used to estimate the total ERD and ERS for each session. As a result of non-physiological artifacts in the EEG, no peaks were found on any days for two out of the 18 subjects, and therefore, those subjects' data was excluded from the dataset.

Accuracy. Three variations of accuracy were calculated: 1) right-target accuracy; 2) left-target accuracy; and 3) total accuracy. Accuracies were averaged across participants in each group for each day, creating a trajectory of task performance across the 10-day experiment (Figure 3).
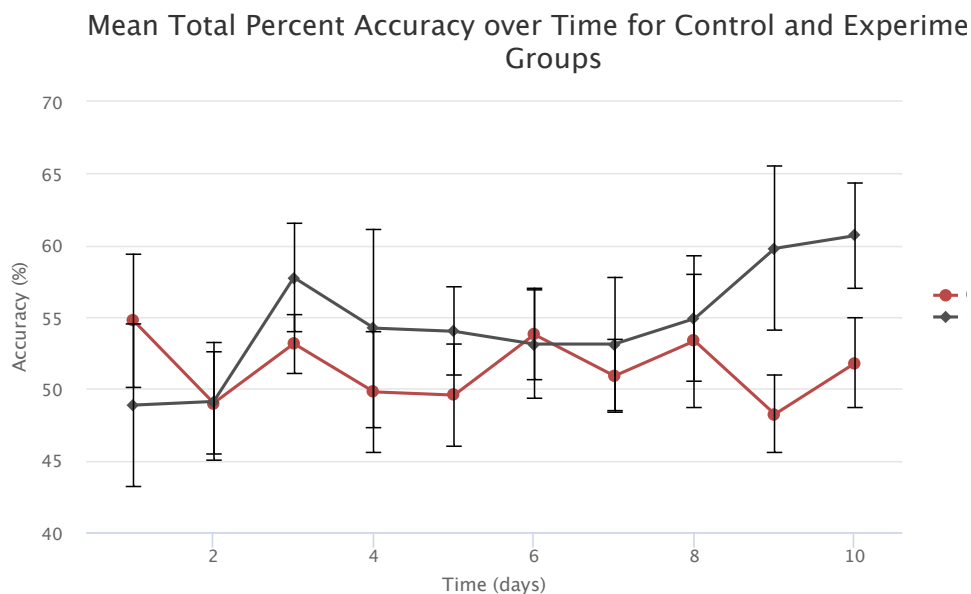
Figure 3. Mean Total Percent Accuracy over Time for Control and Experimental Groups. Behavioral improvement. The tDCS group showed greater mean total percent accuracy over time compared to the sham group (p = 0.021).

Statistical Analysis. Linear mixed models (SPSS Statistics Data Editor, IBM Armonk, NY: IBM Corp), were used to assess the interactions amongst the mu rhythm at rest, mu rhythm during right hand motor imagery, ERD, total percent accuracy, total right percent accuracy, and total left percent accuracy. The latter six were used as the dependent variable when analyzing the interaction with group and day. One-tailed unpaired t-tests were performed to determine if tDCS had a positive effect on total percent accuracy using the means of total percent accuracy over the 10 days of both groups, and the participant's ability to perform total right and left percent accuracy for both groups. A one-tailed unpaired t-test analysis was done to compare the two groups in terms of ERD over time. The daily ERD of each subject in their respective groups were averaged together. To assess the power of the mu rhythms at rest, a two-tailed unpaired t-test was conducted. For each subject, the mu-rhythm values at rest over time were averaged together. An unpaired t-test was conducted using the averaged values of the experimental group to the control group. The effect of tDCS and sham stimulation on mu rhythms during right hand motor imagery was assessed using an one-tailed unpaired t-test. This analysis was conducted using the averaged mu rhythms during right hand motor imagery over time for each subject in both groups. For all statistical analyses the level of significance (α) was set at 0.05.

Results

Participant characteristics

A summary of the demographic characteristics of both groups is presented in supplementary material. There were no statistically significant differences in gender and age between the experimental and control groups ($p = 0.63$; $p = 0.31$).

The effect of tDCS on performance accuracy

Participants in the tDCS group achieved a higher mean total accuracy in the BCI task than those in the sham group (t-value = 2.17, $p = 0.021$) (Figure 3). Additionally, tDCS had a significant effect on the individual's ability to perform right hand motor imagery (t-value = 3.56, $p = 0.001$). There was no statistical difference in terms of improvement over time between groups (t-value = -2.88, $p = 0.995$).

The effect of tDCS on mu rhythms

There was no difference in ERD between the control and experimental groups ($p = 0.22$). The mu rhythms at rest and during motor imagery varied over time for subjects in both control and experimental groups. The average power of the mu rhythms at rest and during right hand motor imagery was not different between groups (t-value = 1.67, $p = 0.116$ and t-value= 1.42, $p = 0.088$, respectively).

Mu-rhythms vs. accuracies

There was no significant relationship between ERD and accuracy: mu-rhythms at rest with total left accuracy ($p = 0.857$); mu-rhythms during right hand motor imagery with total right accuracy ($p = 0.964$); and the difference between the mu-rhythm at rest and during motor imagery with the total accuracy ($p = 0.342$). We found an interaction between Group and Day for mu rhythms at rest (F(2.149), $p = 0.020$) and ERD (F(3.821), $p = 0.001$). Over the ten session, the resting mu rhythm of the tDCS group increased overtime. However, the resting mu rhythm of the sham group remained relatively similar for majority of the ten sessions. We did not find significant interactions

between Group and Day for the other dependent variables (mu rhythm during right hand motor imagery, p = 0.923; total right percent accuracy, p = 0.456; total left percent accuracy, p = 0.935; and total percent accuracy, p = 0.163).

Discussion

The aim of this study was to validate previous studies which evaluated the effect of tDCS on controlling an SMR-based BCI however, using a high density targeted tDCS system at 1mA. Motor imagery was used to create an ERD to affect the amplitude of the sensorimotor rhythm; this change in amplitude was translated into the left and right movement of a ball on a computer screen. tDCS applied over the left sensorimotor area led to significant improvements in task accuracy, as well as accuracy in hitting the right target, which was associated with right hand motor imagery. These behavioral improvements, however, were not associated with changes in the amplitude difference of the SMR induced by motor imagery.

Accuracy

Accuracy is to successfully get the ball in the designated target. After 10 sessions, the tDCS group showed greater total accuracy and total right accuracy of the BCI task, compared to control. These results are in agreement with the findings of Ang et al., who demonstrated that tDCS improved accuracy in motor imagery brain-computer interface with robotic feedback for stroke rehabilitation (21). However, they contradict Wei et al., who did not find a difference in accuracy between pre- and post-anodal tDCS in healthy subjects despite getting a significant increase in ERD of the upper mu post-intervention (10-14 Hz) (29). This difference may be attributed to the length of training (1 day vs. 10 days), which may not have been enough time for participants to learn to control the BCI through motor imagery. Many studies have reported that it takes several days of practicing the BCI task to see improvements in the control of the BCI (30),(31),(32).

Mu Rhythms

There was no significant difference between the control and experimental groups in the daily mean of the mu rhythm over time. This is contrary to our hypothesis where a greater change in the mu-rhythm between rest and motor imagery was expected in the tDCS group. Previous studies have reported that when applying anodal tDCS in healthy subjects, ERD of the mu rhythm increases. This leads to a decrease in amplitude of the power of the mu rhythm and a greater difference in the power of the mu-rhythm between rest and right hand motor imagery (22),(29). In our experiment, the mu-rhythm during rest was calculated using the 1-second epoch between trials; this may have affected these results, as it has been reported that the brain requires a minimum of 7 seconds after a movement trial for the EEG to return to baseline (33). The mu-rhythm of our participants may have had insufficient time to re-establish baseline patterns, resulting in no difference between control and experimental groups. The change in mu-rhythm power has been observed in other studies with longer inter-trial intervals; Matsumoto et al., found that the ERD of the mu rhythm increased post-tDCS when they used an 8-second period inter-trial interval (22).

Limitations

The main limitations of this study were the small sample size and the limited number of participation sessions. Future studies should focus on increasing sample size and greater number of participation sessions in order to determine the effect of tDCS on one's ability to learn.

The mu-rhythm threshold of the BCI task was derived by the experimenters using the protocol described in the methods, and this needed to be re-calibrated daily. The remainder of the task was carried out with the parameters set in this calibration period, and errors at this stage could have reduced the efficacy of the participant's performance with the BCI. For example, if the threshold was set too high, the participant would experience greater difficulty moving the ball to the right target. While this limitation equally affected both groups, and therefore would not change the overall results of this study, it may have reduced the maximum potential accuracy that either group could have theoretically achieved with the BCI.

In this experiment, participants were not cued about the target location for the upcoming trial. Consequently, participants were not able to prepare for the mental task required to control the ball, and needed to generate the requisite state immediately upon the start of the trial. Participants reported anecdotally that when the target was in the same location for numerous trials, it was easier to control the movement of the ball, whereas when it was switching right and left, it made it more difficult to mentally adjust for the appearance of the next target. To decrease the difficulty of subsequent BCI experiments, we recommend cueing the participants about the upcoming trial (21) and allowing them to mentally prepare for task ahead.

Although there were no differences between both groups in terms of demographics (age and gender), other factors such as motivation, fatigue, mood, attention span as well as other psychological factors have been shown to be correlated with MI-BCI performance (34),(11),(35). These factors could explain, in part, the lack of differences observed between groups in terms of change in mu rhythm over time. Our study, having a small sample size, might have also made these inter-individual differences more prominent.

Conclusion

The application of the high density system's targeted tDCS at 1mA over the somatosensory area led to significant improvements in a user's ability to control a SMR-based BCI like that of non-targeted tDCS at 2mA. However, this improvement was not associated with an increase in the difference in power of the SMR between motor imagery and rest. Our results validate those of Ang et al., we demonstrate that tDCS can improve an individual's ability to learn a BCI by speeding up the learning process (12)(21). This suggests that this method has promise in accelerating the process of learning to control BCIs for individuals with severe motor impairments.

Acknowledgements

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