NeuroRegulation

Developing and Applying a QEEG-Informed Transcranial Electrical Stimulation Protocol to Remediate Stuttering in Adults Who Stutter

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Abstract

Introduction. This study investigated whether personalized transcranial direct current stimulation (tDCS) protocols informed by quantitative EEG (qEEG) patterns could enhance treatment outcomes in adults who stutter (AWS), addressing individual variability in neural activity. **Methods.** Twenty male AWS participated in a double-blind, randomized controlled trial. EEG signals were recorded during speech tasks to differentiate neural substrates of fluent and stuttered speech. Over 10 days, participants received 10 sessions of speech therapy combined with tDCS. The experimental group received personalized tDCS (2 mA for 25 min per session) targeting regions identified through qEEG, while the control group received standard stimulation over FC5. Behavioral outcomes (SSI-4 scores) and EEG metrics were compared pretreatment, posttreatment, and at 3-month follow-up using repeated-measures ANOVA. **Results.** Both groups exhibited significant reductions in stuttering severity posttreatment. However, the study group maintained greater fluency at follow-up ($p < .001$). EEG analysis revealed that the study group demonstrated enhanced delta power in the FC5, reduced phase coherence in motor-auditory-somatosensory networks, and greater suppression of high-frequency bands in the personalized group. **Conclusion.** Personalized, qEEG-informed tDCS protocols yielded more sustainable fluency improvements than conventional tDCS, highlighting the potential of individualized neuromodulation strategies for treating stuttering.

Keywords: stuttering; qEEG; individualized tDCS; power spectrum; phase coherence

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Introduction

Stuttering is a neurodevelopmental disorder (Smith & Weber, 2016) that affects nearly 5% of children and persists into adulthood in approximately 1% of cases (Yairi & Carrico, 1992). This condition is characterized by involuntary prolongations, repetitions, hesitations, and blocks at the sound, syllable, and word levels, which interfere with normal speech (Jiang et al., 2012). Despite its detrimental effects on the quality of life, interpersonal relationships, and socioeconomic opportunities of adults who stutter (AWS; Craig et al., 2009; Farrahi et al., 2021; Yaruss, 2010), stuttering presents challenges from both clinical and pathological perspectives. Several neuroimaging studies investigating the neurological basis of stuttering suggest that it is a motor speech disorder with structural and functional abnormalities in specific brain subsystems. Although meta-analyses have yielded robust findings, several controversial issues remain (Etchell et al., 2017). These controversies are primarily due to group studies conducted on individuals who stutter. For example, Wymbs et al. (2013) conducted a study using functional magnetic resonance imaging (fMRI) while four AWS read single words. They found that while the level of within-subject agreement during the task was very high across two separate sessions, the level of between-subject agreement was low (Wymbs et al., 2013). This discrepancy is also evident in electrophysiological studies of stuttering. Research comparing the electroencephalogram (EEG) patterns of fluent and dysfluent states in AWS has indicated differences in brain regions, EEG frequency bands, and functional connections (Mersov et al., 2018; Sengupta et al., 2017; Vanhoutte et al., 2016). Stuttering research often distinguishes between those who stutter (the stuttering “trait”) and the behavior of stuttering (the stuttering “state”; Belyk et al., 2015). However, it provides limited insight into how brain activity in AWS may differ during episodes of stuttering compared to their brain activity during fluent speech.

Some studies on stuttering have introduced specific, different, and sometimes conflicting neurophysiological markers in the frequency ranges of theta (Ghaderi et al., 2018), alpha (Jenson et al., 2018; Saltuklaroglu et al., 2017; Wells & Moore, 1990), and beta (Jenson et al., 2018; Mersov et al., 2016; Saltuklaroglu et al., 2017). Alpha and beta oscillations have been examined more frequently than other frequency bands due to their roles in evaluating sensory feedback (alpha) and in the forward modeling of motor speech programs (beta; Korzeczek et al., 2022). Given the discriminative role of speech preparation in stuttering, many studies have compared the neural activities preceding fluent and disfluent utterances in AWS. Vanhoutte et al. (2016) found that the amplitudes of the contingent negative variation (CNV) before stuttered speech in AWS were not significantly different from the CNV amplitudes observed during fluent speech in fluent speakers. This finding suggests that the neural activity associated with stuttered speech in AWS exhibits similar signal properties to that of fluent speech, indicating the existence of compensatory mechanisms that support fluent speech despite the presence of stuttering (Vanhoutte et al., 2016). While some investigations have demonstrated poor alpha and beta suppression preceding speech in individuals with a stuttering trait (Bayat et al., 2024; Jenson et al., 2020; Jenson et al., 2018), Korzeczek et al. (2022) reported that stuttering severity in AWS is correlated with low beta power during prefluent speech preparation (Korzeczek et al., 2022). In a previous study from this lab on the same sample, the possibility of associating the delta band with the fluency state was also raised (Bayat et al., 2024).

Motor imagery (MI) is a condition examined within stuttering research. Imagined stuttering demonstrates a significant correlation with overt stuttering during a paragraph reading task in a positron emission tomography (PET) study (Ingham et al., 2000) and in a word production task during fMRI (Wymbs et al., 2013), particularly in terms of regional brain activation. Both imagined and overt stuttering display comparable neural activation patterns, especially within motor-related brain regions such as the supplementary motor area (SMA), anterior insula, and cerebellum (Ingham et al., 2000). This finding suggests that the neural underpinnings of stuttering involve these areas, regardless of whether the stuttering is real or imagined. Studies with high temporal resolution to investigate imagined speech indicate that, although covert (i.e., imagined) speech does not require movement or produce reafferent feedback, forward models of speech targets and anticipated sensory feedback are activated through a purely internal loop (Tian & Poeppel, 2010, 2012).

However, due to the neurological nature of the disorder, neural modulation, which is widely used in patients with neurological diseases (Paik, 2015), is an option being considered for stuttering. Applying transcranial electrical stimulation (tES) is one of these techniques that has shown remarkable results in speech and language disorders. Busan et al. (2021) presented noninvasive brain stimulation as a promising intervention to enhance brain function in individuals who stutter. In evaluating recent advancements in stuttering treatment, they explored how direct brain interventions can improve speech fluency (Busan et al., 2021). Brignell et al. (2020) included transcranial direct current stimulation (tDCS) among effective intervention approaches for stuttering (Brignell et al., 2020). The combination of neuromodulation techniques with more “conventional” interventions has been proposed as an emerging strategy for treating stuttering (Busan et al., 2021). Several studies have investigated the use of tDCS in AWS. Some of these studies employed a single-session design (Chesters et al., 2017; Garnett et al., 2019; Karsan et al., 2022; Tezel-Bayraktaroglu et al., 2020; Yada et al., 2019), while others used a multisession design (Bakhtiar et al., 2023; Bashir & Howell, 2015; Chesters et al., 2018; Moein et al., 2022). Two of the single-session studies reported improvement in stuttering severity (Karsan et al., 2022; Yada et al., 2019). Chesters et al. (2018) succeeded in reducing the rate of stuttering during conversation and reading by performing five sessions of anodal stimulation on the FC5 electrode, although this fluency was not stable during conversation after 6 weeks (Chesters et al., 2018). Moein et al. (2022) and Bakhtiar et al. (2023) also reported promising results from using tDCS along with fluency shaping techniques (Bakhtiar et al., 2023; Moein et al., 2022).

The application of tDCS is grounded in a thorough understanding of the neurological basis of the disorder in question. Studies have shown that tDCS, particularly when administered over the left inferior frontal cortex, can significantly reduce stuttering in adults. This improvement is often noted when tDCS is combined with fluency-inducing techniques, such as choral and metronome-timed speech (Chesters et al., 2018; Moein et al., 2022; Taherifard et al., 2021). It demonstrates potential as an adjunctive treatment for stuttering by enhancing neural plasticity and modulating the brain networks, such as the left inferior frontal cortex, anterior insula, and SMA, which are involved in planning and initiating speech movement (Chesters et al., 2018; Chesters et al., 2021). Although tDCS can lead to significant improvements in speech fluency—especially when combined with behavioral therapies—(Garnett et al., 2019), additional research is necessary to optimize its application and understand its long-term benefits.

Quantitative EEG (qEEG) is a powerful and sensitive tool for identifying maladaptive brain activity patterns. Behavioral and mental states—such as reading, visualizing, or speaking—are assumed to be distinct yet uniform, involving similar perceptual and cognitive operations whenever they occur. It is also assumed that each mental operation presents a unique EEG and biochemical profile, reliably reproduced whenever the task or mental state occurs (Kaiser, 2007). QEEG employs advanced techniques for EEG signal feature extraction, including the analysis of specific frequency bands and connectivity (Popa et al., 2020). Implementation of qEEG in clinical practice has the potential to guide personalized medicine, which involves tailoring treatments to individual patients (Keizer, 2019). Since the effects of tDCS depend on the brain's baseline status at the time of application, individual patients exhibit considerable heterogeneity in treatment outcomes (Woods et al., 2016). Bjekić et al. (2022) reported that extracting the dominant theta-band frequency, based on associative memory-evoked EEG changes, can reliably be used to personalize oscillatory tES techniques (Bjekić et al., 2022). Personalized channel selection in brain-computer interfaces (BCI) based on EEG creates a direct pathway between the brain and external devices (Ma et al., 2023). Mastakouri et al. (2017) proposed personalized EEG-informed BCI training and tES to derive optimal tES parameters, such as stimulation site and frequency, as a novel intervention for motor performance rehabilitation in stroke patients (Mastakouri et al., 2017). Depending on the specific stimulation parameters, various tES paradigms can be differentiated, including tDCS and transcranial alternating current stimulation (tACS). Parameters such as intensity, duration, frequency, and the location and size of electrodes enable control over both the immediate effects and aftereffects of tES (Paulus et al., 2013). Brain imaging techniques are utilized to evaluate alterations in both the structure and function of specific brain regions prior to and following tDCS, in addition to measuring the propagation of the tDCS signal throughout the brain (Bashir & Howell, 2015). Garnett et al. (2019) examined the neural effects of high-definition tDCS (HD-tDCS) on the SMA employing fMRI.

Given the significant individual and group differences in relevant research results, we planned to develop personalized protocols based on the qEEG patterns of AWS to enhance the effectiveness and durability of the outcomes from the tDCS protocols. We hypothesized that AWS have distinct neural circuits underpinning fluent versus dysfluent states, prompting us to develop a protocol to differentiate the neural substrates of fluent and stuttered speech by analyzing EEG signals recorded during two speech tasks. Based on the findings of the previous study assessing the brain dynamics of speech preparation (SP) and imagined speech (IS) production in AWS (Bayat et al., 2024), we aimed to improve the quality and quantity of recorded signals by analyzing neural oscillations associated with both SP and IS. For each subject, we identified the areas that exhibited the most significant signal energy differences between the two fluency states. The corresponding areas were potentiated with anodal stimulation if their increased EEG activation was associated with fluent speech, while they were inhibited with cathodal stimulation if their increased EEG activation was associated with stuttering. Participants in the study group received personalized stimulation, while those in the control group received conventional anodal stimulation of the FC5 electrode (Chesters et al., 2018). Since we utilized EEG to identify the area and modality of stimulation, we opted to investigate the effects of the intervention utilizing the same methodological approach. Due to the numerous parameters that must be controlled and conflicting data regarding the role of specific band frequencies in stuttering, we decided to limit the controlled parameters in the offline EEG to montages and current polarity, opting for tDCS instead of tACS.

We hypothesized that, consistent with previous studies, the study group would differ from the control group by exhibiting increased suppression of alpha and beta bands (Jenson et al., 2020; Korzeczek et al., 2022) and enhanced activation in the delta range (Bayat et al., 2024). One method of analyzing functional connectivity in the realm of frequency is measuring the degree of phase correlation. Phase coherence is the most common technique used to assess connectivity in studies related to language functions (Gaudet et al., 2020). To date, no study has examined the electrophysiological implications of treatment on functional connectivity. When comparing fluent and stuttered speech in AWS in terms of phase coherence, Sengupta et al. (2017) reported an increase in phase coherence across theta, alpha, and gamma bands at the left frontal electrodes while preparing for stuttered utterances. They attributed this increase in phase coherence during the preparation for stuttered utterances to heightened general activity associated with stuttering (Sengupta et al., 2017). Accordingly, successful treatment may modulate enhanced functional connectivity associated with the stuttering state, although the investigation of phase coherence in this research is primarily exploratory.

Therefore, we applied fast Fourier transform (FFT) relative band power and the FFT phase coherence algorithm in NeuroGuide to compare the two groups based on their posttreatment signal features. The study was a double-blind, randomized controlled trial with a pretest, posttest, and follow-up design.

Material and Methods

Participants

Twenty-two native Persian-speaking men (mean age: 35 ± 4 years, range 21–42) with persistent developmental stuttering participated in this study. They had no known history of hearing or other speech-language disorders, except for stuttering, and did not have any neurological disorders. Participants were screened for any safety contraindications for tDCS, particularly regarding a history of seizures. Subjects received compensation for their participation. Recruitment was conducted through convenience sampling among patients referred to our community mental health care setup who met the inclusion criteria. All subjects were right-handed (Oldfield, 1971), and their stuttering severity ranged from mild to very severe according to the Persian version of the Stuttering Severity Instrument, Fourth Edition (SSI-4; Tahmasebi et al., 2018). The signals of two subjects were excluded from the analysis due to technical issues, and the data from the remaining 20 subjects were analyzed. It should be noted that the initial qEEG data used in this study was the same as that from our previous study (Bayat et al., 2024), but this study focused on qEEG-informed tDCS, while the previous research speculated on qEEG features in a stuttering state. Written informed consent was obtained from all participants. The institutional ethical review board at Shiraz University of Medical Sciences (IR.SUMS.REC.1397.712) approved all experimental procedures.

Experimental Procedure

The flow diagram of the experimental design (Figure 1) provides an overview of the study procedure. Neural correlations of fluent and nonfluent speech were investigated and compared by analyzing EEG signals during speech tasks in two stages: first, at the beginning of the study to determine the regions of interest (ROIs) for tDCS, and second, at the end of the study to evaluate the intervention results. The speech tasks and signal acquisition process were similar in both stages. Figure 2 illustrates the process of obtaining, preprocessing, analyzing, and comparing the signals, as well as assigning subjects to the study and control groups.

Verbal Stimuli and Experimental Tasks

We employed 50 verbal stimuli (e.g., “Name your two friends” and “Say two fruits that start with the letter A”) as the outcome measure during the signal acquisition phase. This research represents a continuation of our previous studies, and the verbal stimuli as well as their presentation methodology are consistent with the procedures outlined by Bayat et al. (2023). The experimental design comprised five blocks, with 50 verbal stimuli presented in a random order within each block, resulting in 250 trial.

Figure 1. *Flow Diagram of Experimental Design.*

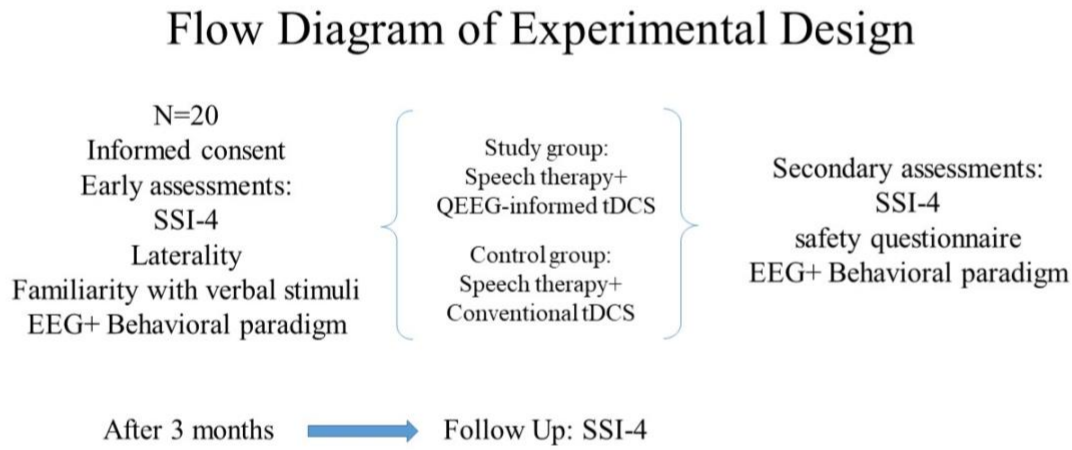
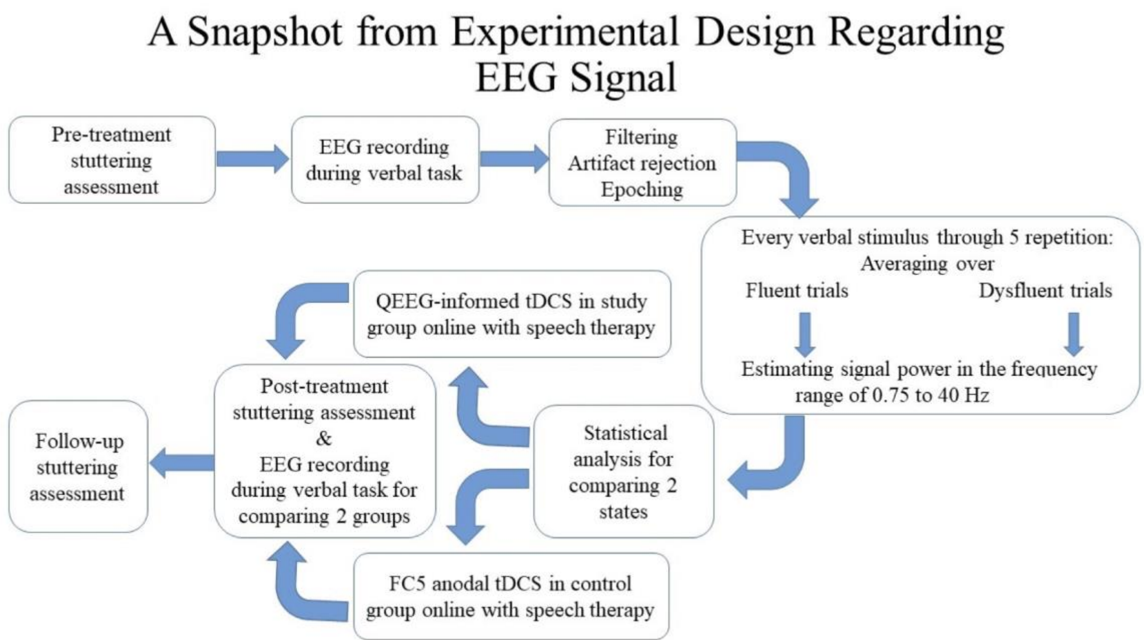


Figure 2. *A Snapshot From Experimental Design Regarding EEG Signal.*



Each verbal stimulus was presented five times to increase the likelihood of achieving the desired conditions through adaptation or sensitization. This repetition aimed to elicit both fluent and disfluent responses from the participants. The adaptation or sensitization effect benefits our task by ensuring that the person experiences both stuttered and stutter-free trials during the five repetitions. Therefore, we do not want to control for this. A schematic representation of each trial and block is illustrated in Figure 3. This methodology enabled us to compare the signals of identical utterances while differentiating them based on fluency.

Initially, participants read a verbal stimulus in silence and prepared to answer it, after which they responded aloud. They were then required to determine whether they had stuttered. In the next two steps, they were asked to imagine and whisper their own utterance (the aloud answer) while maintaining the same fluency state (Figure 3). Each trial consisted of five icons representing five events presented on a PC screen using Psychtoolbox-3 through open-source MATLAB and GNU Octave functions for vision and neuroscience research (available at https://psychtoolbox.org/docs /DownloadPsychtoolbox). Details of the executive procedure can be found in the study by Bayat et al. (2023).

To compensate for the low number of qualified trials, we decided to use the two conditions of SP and IS (among four speech-related tasks) in each trial to indirectly increase the number of available signals. This approach effectively doubled the opportunities to observe neural activity associated with both fluent and disfluent speech states. Although SP and IS tasks involve different cognitive and motor demands, they share overlapping neural mechanisms related to speech preparation and internal monitoring (Bayat et al., 2024). Two events associated with two tasks within specified time windows were selected for analysis and comparison: 2.5 s before the appearance of the speaker icon, designated as SP, and the middle 2 s of the imagining task, referred to as the IS.

During the IS condition, participants were instructed to visualize their responses from the previous step, using the same words and fluency (Figure 4). The first 500 ms of IS were excluded to reduce interference from the previous section. Similarly, the last 500 ms of IS were removed to ensure that the intended activity was captured before it finished. The sections of answering aloud and whispering were excluded from the final analyses. The whisper section (WH) was excluded at this stage due to insufficient evidence in the literature to support a correlation between the whispering condition and the other two conditions used. Additionally, because speech, especially stuttering, generates significant noise in the EEG, most studies prefer not to include parts of this section in signal analysis. A well-trained speech-language pathologist (SLP) supervised the procedure. If a participant responded with incorrect timing (e.g., answering before the relevant cues appeared) or exhibited excessive movement during task execution, the trial was classified as a rejected trial. The same protocol applied to trials in which questions were left unanswered by the examinees. A second SLP reviewed the recorded videos of the subjects during the task and assessed the trials based on their fluency offline. The video recordings were adjusted to ensure that the faces and upper bodies of the participants were visible to the SLP. A trial was considered dysfluent only if the subject and both SLPs agreed on this assessment. Similarly, trials were classified as fluent if the subject and both SLPs deemed them fluent. Participants' own assessments of the presence or absence of stuttering were effective in determining trial classifications. Before the test, participants were instructed to identify even an inaudible uncomfortable feeling as stuttering. Conversely, speech therapists were asked to label only those trials they were completely certain about as stuttering. This approach helped manage the common issues associated with stuttering-state studies. We accepted the risk of mistakenly rejecting some trials to ensure that trials in which the subject pressed the button incorrectly would not be classified as stuttered trials and would be excluded from the experiment. Participants were familiar with the verbal stimuli 1–3 weeks before the experiment. The signals from two subjects were not properly marked due to problems with the parallel port systems, resulting in their exclusion from the analysis. Consequently, the data from the remaining 20 subjects were analyzed.

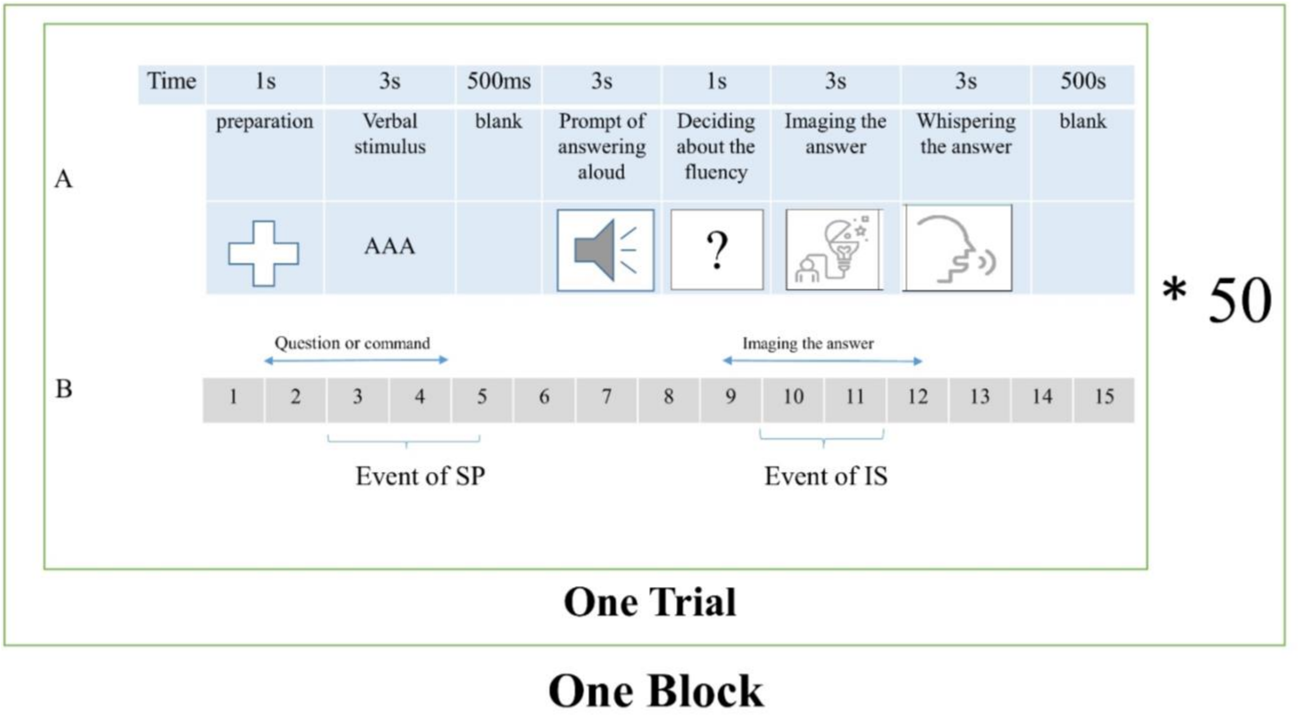
Behavioral Evaluation

Stuttering severity was evaluated using the Persian version of the SSI-4 tool (Tahmasebi et al., 2018). This reliable and valid norm-referenced assessment tool is suitable for both clinical and research purposes (Riley & Bakker, 2009). The SSI-4 measures stuttering severity across four areas of speech behavior: frequency, duration, physical concomitants, and the naturalness of the individual's speech. Two qualified independent SLPs assessed the stuttering severity. The first SLP was trained by the second SLP, who developed the Persian version of the SSI-4, to evaluate and score stuttering severity and related behaviors. The stuttering evaluation process was conducted in three stages: 7–10 days before the intervention, 3–7 days after the intervention, and 3 months after the intervention during the follow-up stage.

Electrical Neuroimaging-EEG Acquisition

EEG data were recorded at a sampling rate of 512 Hz using a 64-channel BrainVision amplifier (Mulgrave, Australia). The active electrodes were arranged on an elastic cap according to the standard 10–10 electrode positioning system, and the electrical impedance of the scalp electrodes was maintained below 5 kΩ. Subjects were instructed to minimize eye blinks and body movements during the three events: SP, IS, and WH. Approximately 2-min pauses between blocks were implemented to prevent fatigue and muscle tension. The parallel port system delivered a marking pulse with the display of each icon to synchronize EEG signals for accurate offline analysis. The procedure of EEG recording was accomplished in two stages, 1–2 weeks before and 5–10 days after the intervention.

Figure 3. *The Study Design Representing a Trial and a Block. A) Timing of Tasks and Their Corresponding Icons B) Time Windows of Investigated Events.*



Quantitative Analysis of the EEG (qEEG)

Filtering, Artifact Rejection, and Separation of Fluent and Disfluent Trials. Once these trials were evaluated by SLPs and participants, MATLAB R2018b software was used to separate the signal and analyze its power for fluent versus disfluent conditions. These tasks were entirely separate, with human evaluation ensuring accuracy in classification and MATLAB providing precise, objective segmentation for signal analysis. EEG signals from each subject were imported into MATLAB-based EEGLAB_14_1_2b open-source software (Delorme & Makeig, 2004). The signals were then filtered using a fifth-order Butterworth bandpass filter with cut-off frequencies of 0.75 Hz and 45 Hz. The average montage for EEG acquisition was utilized.

Training for independent component analysis (ICA) was performed using the “Runica” algorithm in EEGLAB. This process resulted in 64 independent components corresponding to the number of recording electrodes. ICA is an effective tool for unmixing volume-conducted EEG signals, yielding components from both neural and nonneural sources, such as muscular and artifactual sources (Bell & Sejnowski, 1995; Olbrich et al., 2011). Training was stopped by default when the weight change was below 1×10^{-7} for more than 33 channels.

The ADJUST.1.1.1 tool (an open-source EEGLAB plugin available at https://sccn.ucsd.edu/wiki/Plugin_list_all), was employed to detect nonneural sources. ADJUST is a fully automated algorithm that identifies independent components of artifacts by combining stereotyped spatial and temporal features. These functions are optimized to capture blinks, eye movements, and frequent discontinuities in the feature selection dataset. The algorithm then provides a fast, efficient, and automated process for applying ICA to remove artifact (Mognon et al., 2011). After the removal of the detected components utilizing the ADJUST plug-in, an experienced neuroscientist performed a visual inspection to identify components resulting from muscle tension artifacts and faulty channels. Components associated with blinking or speaking tension were also eliminated. Channels identified by the neuroscientist as faulty were interpolated employing a spherical method for each participant.

Items for which responses across the five repetitions included both fluent and disfluent utterances were extracted, while those that were consistently fluent or disfluent across all five blocks were excluded. This approach allowed for a comparison of neural activation between two brain states differing by a singular feature (e.g., fluency vs. disfluency), while sharing linguistic aspects (e.g., the verbal stimulus as well as the phonological and semantic components of the response). The signals accompanying fluent and stuttered responses were subsequently averaged separately for each verbal stimulus across the two tasks (i.e., SP and IS) at each electrode. This approach aimed to equalize the sample sizes of the fluent and stuttered utterances. The number of accepted verbal stimuli (items) for each participant is presented in Table 1.

ROIs, Current Types, and Signal Processing for Individualized Stimulation

In this stage, electrodes were allocated to 13 areas based on their involvement in speech and language. This allocation was accomplished using an algorithm introduced by Giacometti et al. (2014) for assigning scalp coordinates of EEG electrodes to cortical areas. We applied a reference table that correlates electrode proximity with cortical regions, consistent with our 64-channel system and the international 10–10 system. We then selected the ROIs for stimulation by comparing the signals associated with fluent and stuttered utterances from the electrodes in these 13 areas. This comparison was based on the accepted trial signals for each participant. The selected 13 areas along with their corresponding scalp electrodes are provided in Table 2. The average signal power in the frequency domain was computed for fluent versus dysfluent items between 0.75 and 40 Hz using FFT in MATLAB software. Band-power features, which represent the power of the EEG signal in specific frequency bands, are commonly used to identify and classify mental states. Kuremoto et al. (2017, 2018) proposed a feature extraction method using FFT and selected discriminant coefficients by maximizing the area under the curve (AUC) of the receiver operating characteristic (ROC) to differentiate among various classes of mental tasks. Variations in all EEG bands during both fluent and dysfluent speech production have been observed (Ghaderi et al., 2018; Jenson et al., 2020; Mersov et al., 2016; Piai & Zheng, 2019; Rimmele et al., 2018; Saltuklaroglu et al., 2017; Sengupta & Nasir, 2016). In this study, the power of all signals in the frequency range of 0.75 to 40 Hz was considered as the EEG feature, without separating them into specific frequency bands, which include delta, theta, alpha, beta, and gamma EEG bands. The power of EEG signals corresponding to fluent and dysfluent responses was compared for each participant using paired *t*-tests and the AUC of the ROC. This estimation was performed across 13 areas to identify ROIs. The procedure for selecting the ROIs involved finding the area with a) the lowest *p*-value resulting from the aforementioned *t*-test, provided there were significant differences ($p < .05$), and b) the highest AUC, if any measures exceeded 70% in the SP condition. An alternative option was to select an area that demonstrated both the lowest *p*-value and the highest AUC in the two conditions of SP and IS (a total of four measures) for participants who did not meet the two criteria. This area was determined based on a voting process among the specified measures, with the area receiving the most votes chosen as the ROI. The type of current applied in the study group (anodal or cathodal) was selected according to the sign of the difference in the measurement of fluent and dysfluent signal power in the selected ROI for each participant. In other words, participants in the study group received anodal stimulation if the sign of the difference indicated that the signal power associated with fluency was higher. Conversely, they received cathodal stimulation if the signal power associated with stuttering was higher. The control group received anodal stimulation over the FC5 electrode.

Table 1
The Number of Accepted Items of Verbal Stimuli From the Total of 50 Presented Verbal Stimuli in Each 20 Participants

Participant Number	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20
Total Number of Accepted Verbal Stimuli	34	15	17	40	28	8	35	15	11	41	32	13	24	17	25	21	13	9	14	25

The priority of the SP condition was related to its determining role in stuttering (Mersov et al., 2018; Mersov et al., 2016; Vanhoutte et al., 2016). In determining the ROIs, when no area met the criteria of a significant paired *t*-test and an AUC above 70%, we used the signals from the two studied conditions, SP and IS. To compare the correlation between the two conditions, we separately calculated the differences in average signal power for fluent and dysfluent items across both conditions for all participants within their ROIs. The Pearson correlation coefficient was used to compare the obtained values and investigate the correlation between the two conditions of SP and IS. This procedure was completed for both pre- and posttreatment signals.

Stimulation and Online Speech Therapy

Participants were randomly assigned to study and control groups in this stage. Both groups received stimulation at 2 mA for 25 min using 5x5 cm electrodes while applying fluency shaping methods. Because the study was double-blind, each participant experienced both personal and conventional montages. This design ensured that neither the participants nor the SLP could identify which group the individuals were assigned to based on the montage used. By using NeuroStim 2 (made in Iran), the personalized tDCS protocol in the study group delivered real stimulation, while the control group received real anodal stimulation applied just over the FC5 area. While one montage delivered real stimulation, the other was set to sham. The cathode electrode was positioned on the right supraorbital ridge for conventional montages and on the opposite deltoid (extra cephalic) for individualized montages. No significant adverse effects were reported during or after the stimulation sessions. Participants were closely monitored, and while a few individuals reported minor discomfort, such as mild itching or tingling, this aligns with findings from previous studies on tDCS.

A certified SLP recorded the instructions and materials for fluency shaping methods, including easy onset and prolongation. In the first session, the subject watched a videotape in which the SLP explained these methods for 10 min. During the next 10 min, participants watched a prerecorded videotape of a speaker demonstrating the described techniques. In the final 10 min, the participant listened to a prerecorded voice and attempted to produce speech in unison with it. At this stage, an independent SLP corrected any issues related to the implementation of the method. From the second session to the end of the 10 sessions, prerecorded speech tasks containing the mentioned methods were used as therapeutic materials, and participants were asked to repeat the same patterns. The tasks began with single words and progressed to 8-word sentences. The SLP supervised the procedures and provided corrections as needed. Speech therapy was conducted online with tDCS.

Table 2

13 Selected Areas and Corresponding Scalp Electrodes According to 10–10 International System Consistent With 64-Channel EEG Recording According to the Algorithm Introduced by Giacometti et al. (2014)

Areas number	Laterality and name of the areas	Correspond electrodes
1	Right Pars Opercularis	32 & 9
2	Left Pars Opercularis	28 & 6
3	Right Triangularis	5 & 32
4	Left Triangularis	1 & 24
5	Right Superior Temporal Gyrus	14 & 41
6	Left Superior Temporal Gyrus	10
7	Right Inferior Parietal Cortex	22 & 45
8	Left Inferior Parietal Cortex	20 & 42
9	Right Banks Superior Temporal Sulcus	41 & 23
10	Left Banks Superior Temporal Sulcus	37
11	Right Precentral Gyrus	13 & 35
12	Left Precentral Gyrus	11 & 34
13	Paracentral Lobule	34, 12, & 35

Signal Analyses

NeuroGuide software compared FFT relative power and FFT phase coherence measures in the fluent state before and after the intervention between the study and control groups across the 19 default electrodes of the software, following the 10–20 system. The comparison at the pretreatment stage aimed to investigate the homogeneity of the two groups. NeuroGuide software (NG 3.0.9; Applied Neuroscience, St. Petersburg, FL, USA) is a sophisticated tool developed for analyzing frequency power and connectivity measures of EEG signals (available at https://appliedneuroscience.com /neuroguide/ and https://anineuroguide.com/; Cannon et al., 2012).

The FFT method employs mathematical techniques to analyze EEG data. Characteristics of the acquired EEG signal are computed using power spectral density (PSD) estimation to selectively represent the EEG samples signal (Al-Fahoum & Al-Fraihat, 2014). A phase correlation coefficient indicates the amount of variance in one signal that can be explained by another signal, and it is a normalized quantity ranging between 0 and 1 (Bastos & Schoffelen, 2016). Phase coherence is the most common technique used to measure connectivity in research on functional brain communication related to language functions (Gaudet et al., 2020). Since the study aims to select the appropriate type of individualized stimulation based on patterns of neural activity associated with fluent speech, we planned to compare the two groups in terms of neural activity during fluency. Subsequently, the posttreatment analysis compared the two groups in the fluent state to investigate the effects of the two protocols on the neural circuits involved in fluency.

Statistical Analyses

The homogeneity of the two groups was investigated regarding age, education, and stuttering severity using an independent *t*-test or its nonparametric equivalent before the intervention. The stuttering severity of the two groups was compared across three time points—pretreatment, posttreatment, and follow-up stages—using repeated measure ANOVA (rmANOVA). In this analysis, the three time points served as the within-subject factor with three levels, while the stimulation protocol was the between-subject factor, and the dependent variable was the severity of stuttering as measured by SSI-4 scores. The interaction between time and group was significant, $F(1) = 17.95$, $p < .001$, so a 2*2 rmANOVA was applied to compare the two groups at the pretreatment versus posttreatment and a 2*2 rmANOVA to compare pretreatment versus follow-up time points. A one-way repeated measures ANOVA (rmANOVA) for pairwise comparison of the three time points was performed to compare SSI-4 measures within each group separately. The correlation coefficient for the power difference of fluent and dysfluent signals between the SP and IS conditions was calculated using the Pearson correlation coefficient. Inter-rater reliability for stuttering severity scores rated by two SLPs was assessed using Cronbach's alpha coefficient. All of the aforementioned statistical analyses were conducted using SPSS-22 software. ROIs were determined using paired *t*-tests and AUC of the ROC method in MATLAB R2018b. The analysis and comparison of EEG signals between the two groups were conducted in NeuroGuide software, utilizing an independent *t*-test corrected for multiple comparisons.

Results

Homogeneity of Two Groups

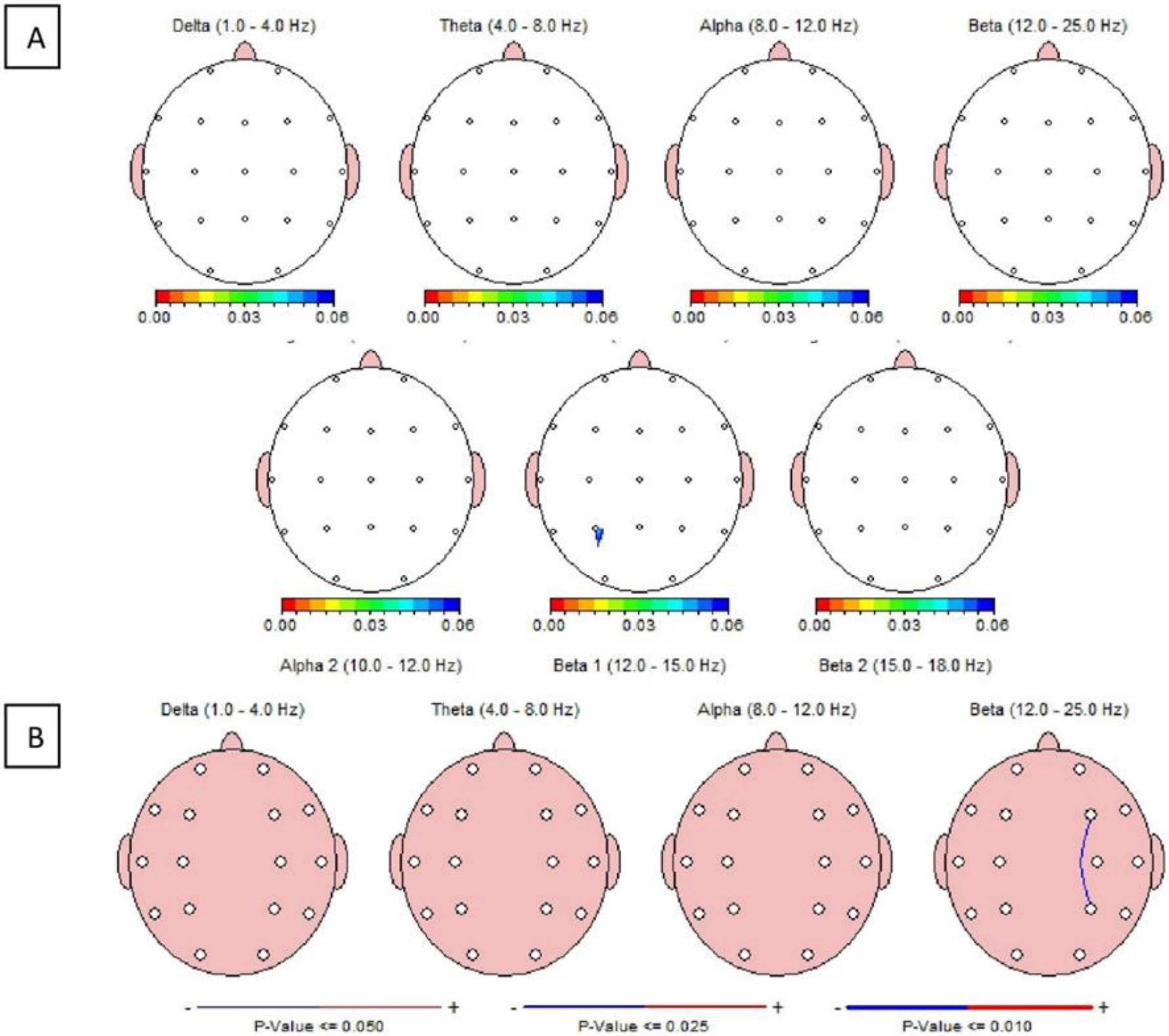
Homogeneity of Behavioral and Demographic Features. The study and control groups were homogeneous in terms of education, $t(18) = 0.20$, $p = .739$, and age, $t(18) = 0.19$, $p = .315$. This homogeneity also applied to the severity of stuttering, as measured by the SSI-4, $t(18) = 0.185$, $p = .856$. Table 3 presents the mean and standard deviation for the demographic variables and stuttering severity from the early assessments.

Signal Homogeneity. Signals from the study and control groups did not reveal remarkable differences in FFT relative power and FFT phase coherence during the conditions of SP and IS in the fluent state. These results are depicted in Figures 2 and 3, respectively.

Table 3
Mean and Standard Deviation of Age, Education, and SSI4 Scores in Two Groups

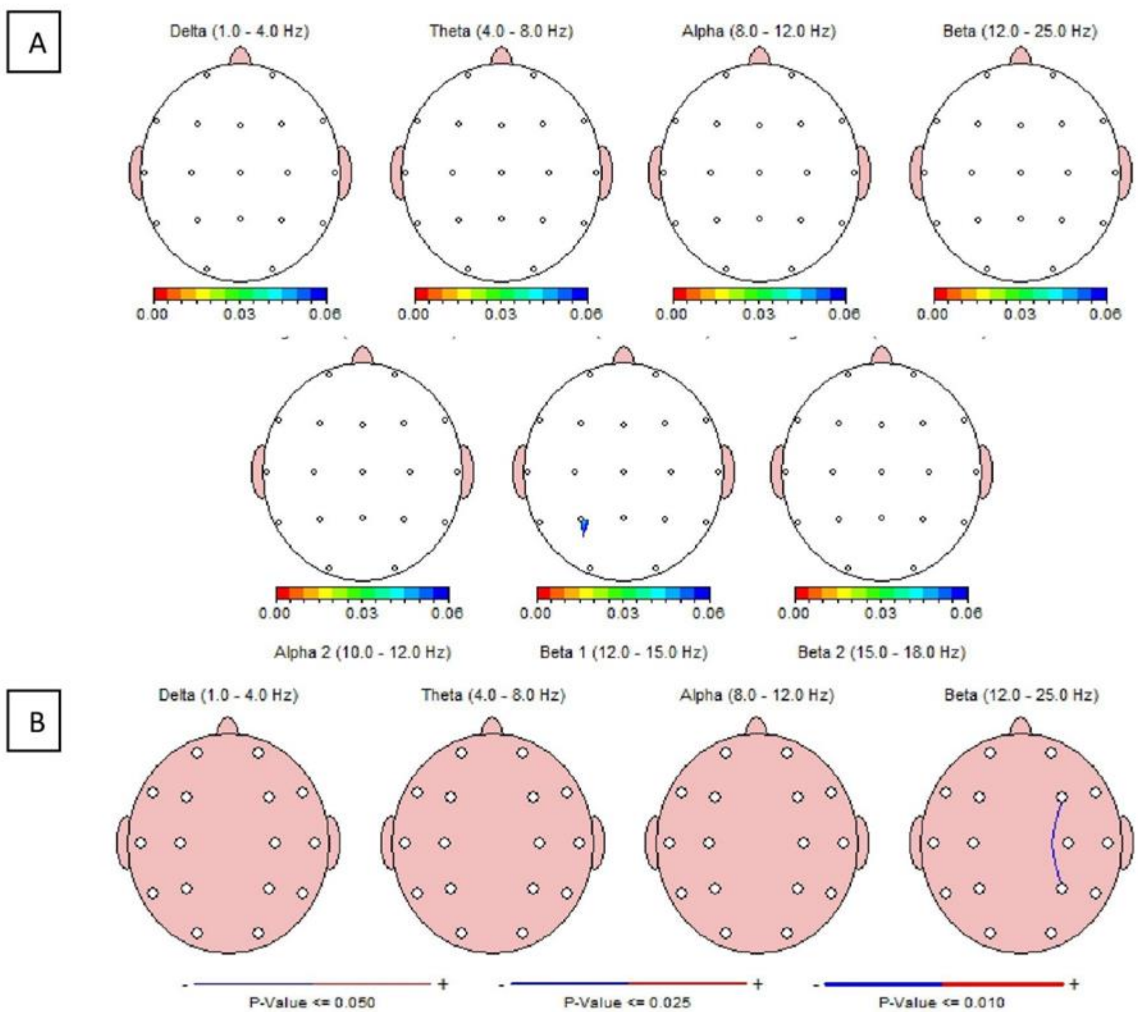
	Age (Mean ± *SD*)	Education (Mean ± *SD*)	SSI-4 (Mean ± *SD*)
Study group	35.70 ± 1.94	14 ± 2.67	32.30 ± 8.96
Control group	32.5 ± 7.76	15 ± 1.94	31.60 ± 7.97

Figure 2. *Homogeneity of Two Groups According to Pretreatment through SP Condition.*



Note. Legend: A) FFT Relative Power and B) FFT Phase Coherence.

Figure 3. *Homogeneity of Two Groups According to Pretreatment Through IS Condition.*



Note. Legend: A) FFT Relative Power and B) FFT Phase Coherence.

ROIs and Current Types

The ROIs, along with the measures of *p*-value and AUC for each participant, are presented in Table 4. Participants numbered 4, 5, and 7 met the first criteria: a) the lowest *p*-values indicating significant differences ($p < .05$), and b) the highest AUC of the ROC, with measures above 70% in the SP condition. The remaining seven participants were included based on the second criteria, which selected from the lowest *p*-values and the highest AUC measures in the SP and IS conditions. The types of current applied to each participant are shown in Table 5. Of the ten participants in the study group, six received anodal stimulation, while the remaining subjects received cathodal stimulation according to the difference in the measurement of fluent and dysfluent signal power in the selected ROI (Table 5). Since the study aimed to compare the effects of personalized stimulation with conventional stimulation, the impact of the type of stimulation (anodal vs. cathodal) was not investigated.

Additionally, the Pearson correlation test revealed a high correlation between the pretreatment power of EEG signals in the SP and IS conditions across the ROIs ($r = 0.87$, $n = 20$, $p < .001$). This correlation was also evident for the posttreatment power of EEG signals ($r = 0.900$, $n = 20$, $p < .001$).

Table 4
ROIs of Stimulation, the Areas With the Highest Measures of Area Under the Curve of the Receiver Operating Characteristic (AROC) and Their Detected Measures in SP and IS Conditions, the Areas With the Lowest P-Values and Their Detected Measures in SP and IS Conditions

Groups	Participants number	Area with the highest AROC according to SP	AROC measure according to SP	Area with the highest AROC according to SP	*p*-value measure according to SP	Area with the highest AROC according to IS	AROC measure according to IS	Area with the lowest *p*-value according to IS	*p*-value measure according to IS	ROI for stimulation
Study Group	1	9	0.6332	9	0.1439	9	0.7223	9	0.0118	9
	2	3	0.6978	1	0.1052	3	0.6267	3	0.3131	3
	3	3	0.8339	11	0.3555	6	0.9377	6	0.00026	6
	4	6	0.7075	6	0.022	6	0.6513	11	0.3251	6
	5	4	0.7015	4	0.0185	4	0.6359	4	0.3139	4
	6	4	0.5625	5	0.712	5	0.6094	5	0.3955	5
	7	12	0.7119	12	0.0081	12	0.8157	12	0.00083	12
	8	7	0.875	4	0.0879	13	0.9375	13	0.0517	13
	9	3	0.7591	3	0.1483	3	0.6275	3	0.09578	3
	10	6	0.6406	6	0.0616	6	0.6563	4	0.3624	6
Control Group	11	10	0.75	10	0.0028	8	0.7813	10	0.0024	2
	12	8	0.9586	8	0.0023	9	0.999	9	0.0021	2
	13	10	0.5937	10	0.3225	10	0.5781	10	0.3057	2
	14	4	0.9343	4	3.6E-05	9	0.9792	8	7.7E-05	2
	15	10	0.5603	10	0.3224	10	0.5781	10	0.3057	2
	16	13	1	2	0.0067	13	1	13	0.0301	2
	17	4	0.9941	4	6E-06	7	1	7	0	2
	18	4	0.5961	4	0.7129	1	0.6064	5	0.3955	2
	19	8	0.9586	8	0.0023	9	0.929	9	0.0021	2
	20	10	0.612	10	0.3224	12	0.5712	6	0.3433	2

Table 5
The Current Types According to the Difference Between the Signal Power Along With the Fluent and the Dysfluent States

Type of Applied Current	DS in the IS State	DS in the SP State	Participant Number
Anode	1.118100	1.524300	1
Cathode	−.037200	−.026400	2
Anode	.180200	.238100	3
Anode	.286000	.076500	4
Anode	.022800	.061500	5
Cathode	−.426800	−.171300	6
Cathode	−.234800	−.027500	7
Anode	.005000	.000900	8
Cathode	−.008400	−.021600	9
Anode	1.176400	.573000	10

Note. DS: Difference score, SP: speech preparation, IS: imagined speech; DS shows the difference between the signal power along with the fluent and the dysfluent states in the SP and IS states.

Behavioral Results

According to a 2*3 rmANOVA for SSI-4 measures, the main effect for the three levels of the within-group subject factor (pre, post, and follow-up) was significant, $F(2) = 76.96$, $p < .001$. However, no meaningful between-group subject effect was obtained, $F(1) = 3.53$, $p = .076$. The interaction between time and group was also significant, $F(1) = 17.95$, $p < .001$. A 2*2 rmANOVA was then applied to compare the two groups at the pretreatment versus posttreatment and pretreatment versus follow-up time points. The results indicated a preference for the study group during the follow-up stage (Table 6). Additionally, the results from a one-way rmANOVA for pairwise comparisons of the three time points are presented in Table 7 for SSI-4 measures. Stuttering severity significantly decreased after the intervention and increased 3 months later, yet remained significantly lower than the pretreatment level in both groups. The trend of stuttering severity across the three time points for the two groups according to the SSI-4 tool and scatter plots illustrating individual data points for the ANOVA results are illustrated in Figure 4A and Figure 4B, respectively. High inter-rater reliability was found (Kappa = 0.91, $p < .005$) for stuttering severity scores rated by two SLPs.

Table 6
Pairwise Comparisons of SSI-4 for Stuttering Severity scores of the Three Time Section in Two Groups

		Source	df	*F*	Sig.
2*2 rmANOVA	Comparing groups at pre vs. posttreatment	Time	1	95.31	*p* < .001*
		Group	1	0.21	*P* = .65
		Time*Group	1	0.86	*P* = .37
	Comparing groups at pretreatment vs. follow-up	Time	1	68.45	*p* < .001*
		Group	1	2.61	*P* = .123
		Time*Group	1	17.96	*p* < .001*

Note. Meaningful differences were shown with *in the column of Sig.

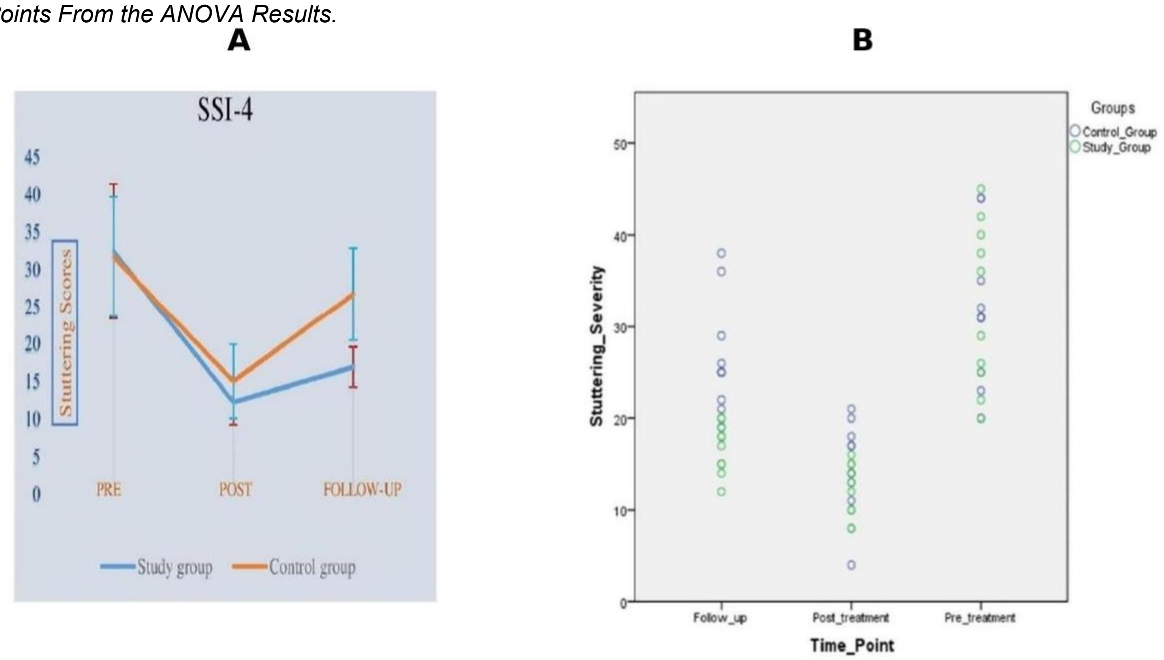
Table 7

Pairwise Comparisons of SSI-4 for Stuttering Severity Scores of the Three Time Point in Two Groups According to rm One-Way ANOVA

		Source	df	*F*	Sig.
2*2 rmANOVA	Comparing groups at pre vs. posttreatment	Time	1	95.31	*p* < .001*
		Group	1	0.21	*P* = .65
		Time*Group	1	0.86	*P* = .37
	Comparing groups at pretreatment vs. follow-up	Time	1	68.45	*p* < .001*
		Group	1	2.61	*P* = .123
		Time*Group	1	17.96	*p* < .001*

Note. Meaningful differences were shown with *in the column of Sig.

Figure 4. *A) Trend of Stuttering Severity Through the 3-Time Section in Two Groups. B) Scatter Plots Displaying Individual Data Points From the ANOVA Results.*



Note. Legend A: Pre: Pretreatment, Post: Posttreatment, and follow-up stage according to the Persian version of SSI-4.

Results of Signal Analyses

Red colors in the figures representing phase coherence indicate significant increases in the study group, while blue colors signify meaningful increases in the control group. The color maps for relative power highlight the areas with significant differences.

Results of Signal Analysis Through the Condition of SP. The study group showed a relative increase in delta band power at the FC5 electrode (equivalent to the Broca region) and an increase in beta band power over the occipital areas of the opposite hemisphere, compared to the control group. In the control group, the power of the low beta-frequency band showed a significant increase compared to the study group. Comparing the two groups in terms of phase coherence demonstrated that preparation for fluent speech in the control group was associated with enhanced inter- and intrahemispheric functional connectivity (FC) across the spectrum. In contrast, the study group exhibited increased left intrahemispheric FC through delta and theta band frequencies during preparation for fluent speech. The results are depicted in Figure 5.

Results on Signal Analysis Through the Condition of IS. In the IS condition, the difference between the two groups largely followed the pattern observed in the SP condition. In the study group, fluency was associated with increasing delta band power over the F3 and C3 electrodes. In contrast, the control group exhibited fluency accompanied by increased beta band power over the F7, FC5, O1, FC6, C4, and Cz electrodes. Additionally, in the control group, imagined fluent speech was linked to greater inter- and intrahemispheric FCs connectivity across the spectrum. The results of the comparison between the two groups in the IS condition are illustrated in Figure 6.

Figure 5. *Posttreatment FFT Relative Power (RP) and FFT Phase Coherence Through SP Condition.*

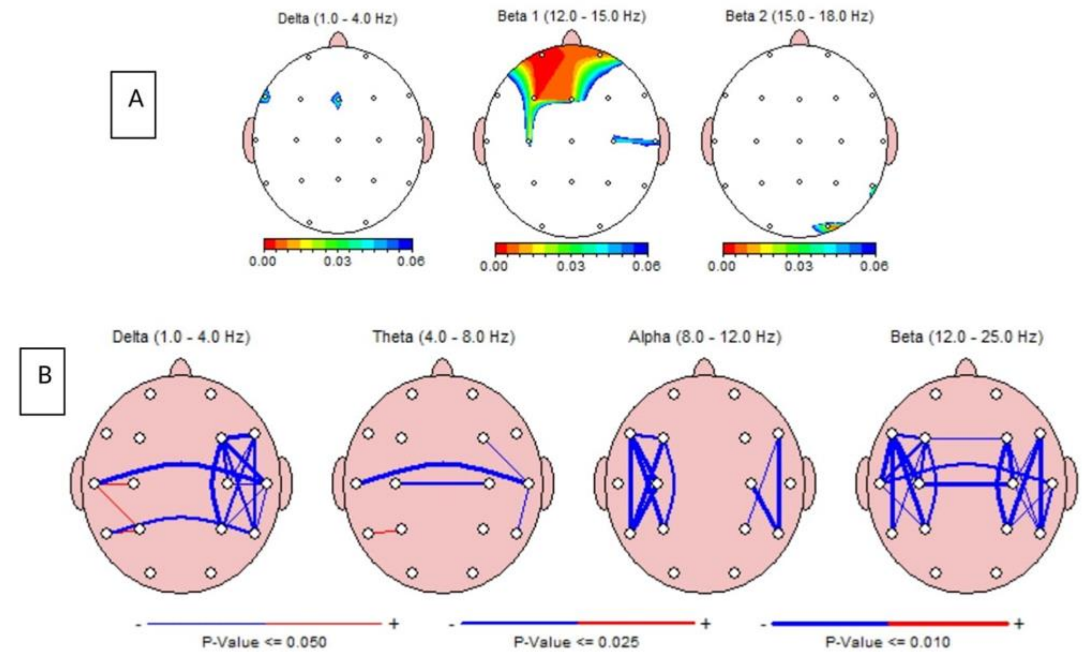
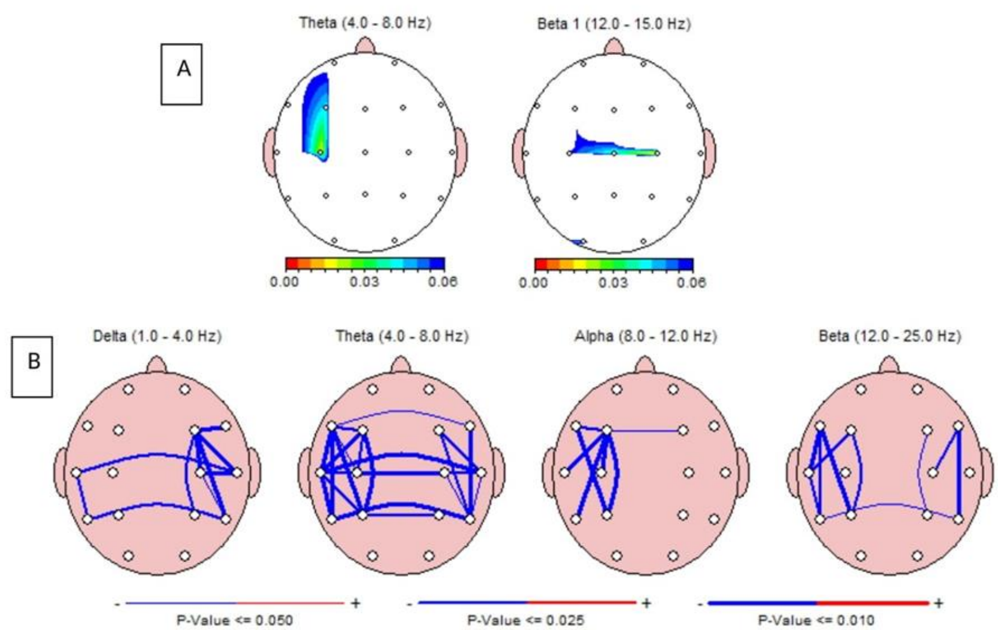


Figure 6. *Posttreatment FFT Relative Power and FFT Phase Coherence Through IS Condition.*



Note. Legend: A) increased theta relative power over F3 and C3 in the study group and increased beta relative power over F7, Fc5, Fc6, C4, Cz, and O1 in the control group. B) increased inter- and intrahemispheric phase coherence through the spectrum in the control group (blue lines).

Discussion

This study aims to investigate the effect of tDCS, optimized through EEG processing for each subject, on enhancing the effectiveness and maintenance of speech therapy outcomes in AWS. To achieve this, we compared two types of electrical stimulation in the study and control groups. The control group received anodal stimulation of the FC5 region according to a conventional pattern for stuttering, while the study group received stimulation based on a comparison of fluent and dysfluent signal power. The application of tDCS in both groups was integrated with speech therapy. We compared the two groups based on behavioral outcomes and changes in signal patterns, both immediately after 10 sessions of intervention and again after 3 months. We planned to monitor the persistence of behavioral outcomes, as relapse is a common complication in stuttering therapy following the cessation of treatment (Craig et al., 2002; Cream et al., 2009; DiLollo et al., 2002).

Behavioral Results

One of the main objectives of the present study was to investigate the effect of qEEG-based personalized tDCS protocols. This section compares the short-term and mid-term outcomes of the two types of stimulation patterns used. Comparing pretreatment and follow-up stage time points indicated a priority for the study group during the follow-up stage. This suggests that although personalized stimulation patterns may not enhance the effectiveness of the intervention, they may lead to a long-lasting effect. This finding is clinically significant for a disorder such as stuttering (Craig et al., 2009; Cream et al., 2009; DiLollo et al., 2002).

One common challenge in stuttering therapy is the time-consuming nature of behavioral interventions needed to achieve clinical results (Blomgren, 2013; Craig et al., 2002; Onslow et al., 1996). Therefore, any method associated with rapid remediation of stuttering will be highly regarded. In the present study, posttreatment outcomes showed a decrease in stuttering scores of 20.2 ± 0.94 in the study group and 16.7 ± 1.57 in the control group, according to the SSI-4. The effectiveness of tDCS increases when combined with behavioral interventions. This combination has demonstrated greater and more lasting fluency improvements than behavioral interventions alone. Chesters et al. (2018) conducted one of the first attempts to investigate the use of tDCS in AWS. They found that after 5 days of anodal stimulation of the left inferior frontal gyrus (IFG) combined with temporary fluency-inducing methods, stuttering scores were significantly reduced based on the SSI-4 tool in the study group (Chesters et al., 2018). Since the stimulation montage in the control group of our study was similar to that used by Chesters et al. (2018) in their study group, the significant difference in the results can be attributed to the variations in methodologies (Chesters et al., 2018; Chesters et al., 2017; Garnett et al., 2019). In addition, we designed a 10-session treatment over 2 weeks for our study, as opposed to the five sessions implemented by Chesters et al. (2018). Thus, the duration of the treatment and the methods used in fluency induction appear to be significant factors in remediating stuttering in the short term. One of the challenges in stuttering therapy is the relapse of the disorder after treatment cessation. A considerable amount of research has been conducted to investigate the causes and methods of prevention (Block et al., 2006; Miller & Guitar, 2009). In this context, follow-up on the results will be a crucial aspect of any treatment. While both our groups experienced a decrease in stuttering severity immediately after the intervention, the control group’s stuttering severity significantly increased after 3 months but remained lower than baseline. In contrast, Chesters et al. (2018) reported that the mean stuttering scores of the study group in the SSI-4 test returned to baseline after 6 weeks. This difference may be attributed to the fluency-inducing method and/or the duration of our intervention. Moein et al. (2022) conducted another study that reported the effectiveness of tDCS in reducing the severity of stuttering. They found that 6 days of anodal stimulation of the left superior gyrus, combined with delayed auditory feedback (DAF), effectively reduced stuttered syllables immediately and up to 6 weeks postintervention. However, this study did not incorporate a follow-up design. This suggests that personalized tDCS may lead to more durable treatment effects.

Results of Signal Analyses

Signal Analyses Through the Condition of SP. IFG plays a crucial role in stuttering, particularly in relation to speech production and motor planning (Lu et al., 2010). Research indicates that individuals who stutter demonstrate atypical activation and connectivity patterns in the IFG, which may contribute to their speech dysfluency (Zhang et al., 2022). Studies reveal overactivation in the right IFG and underactivation in the left IFG among AWS, potentially indicating an increased demand for feedforward planning and dysfunction in auditory feedback systems (Cai et al., 2024). The right IFG is believed to function as a compensatory mechanism in stuttering, possibly as part of an expanded core timing network that includes the basal ganglia and SMA (Etchell et al., 2014a). One of the expected changes after speech therapy for stuttering is increased activity in the IFG (Lu et al., 2017). Personalized tDCS targeting the IFG or its connected networks may help restore functional lateralization, thereby enhancing fluency. Our findings support this framework, as participants in the study group demonstrated greater suppression of maladaptive high-frequency activity in motor-related networks, likely reflecting improved neural coordination.

Furthermore, the study group showed an increase in the relative power of the delta band at the FC5 electrode during the SP condition (Figure 5). This increased activity at FC5 was observed in the study group, while the control group utilized anodal stimulation at FC5. One of the key advantages of personalized tDCS is its potential to optimize therapeutic outcomes by targeting specific neural circuits involved in stuttering. For example, our study found that stimulation aimed at enhancing delta-band activity in the motor preparation areas, specifically at the FC5 electrode during the SP condition, aligns with previous research indicating that delta oscillations play a role in facilitating error monitoring (Yordanova, Falkenstein, & Kolev, 2024). These delta oscillations produce error-specific signals that are essential for performance monitoring, enabling the distinction between correct and incorrect responses (Yordanova, Falkenstein, Hohnsbein, et al., 2004). In contrast, when comparing the control group to the study group during speech preparation, the relative power of the beta band in the electrodes corresponding to the left and right sensory-motor speech-related areas remained significantly higher in the control group. This suggests that despite achieving fluent speech, there has not yet been sufficient beta suppression necessary for an effective motor system (Bartolo et al., 2014; Etchell et al., 2014b; Mock et al., 2016). These results align with the theory regarding defective beta band suppression in stuttering. Accordingly, there are issues in specific brain areas of AWS, such as the basal ganglia and the SMA. These regions provide the internal timing substrate essential for planning and executing movements. Stuttering is considered an internal timing disorder characterized by modulations in the oscillatory power of the beta band (Etchell et al., 2014b). The sensitivity of beta oscillations to sensory-motor deficits resulting from stuttered speech has been well established (Jenson et al., 2020). In this context, a relapse of stuttering in this delicate and vulnerable network is predictable.

The role of the cerebellum, alongside the basal ganglia-thalamus-cortex circuit, has been demonstrated in internal timing for the smooth, rapid, and skillful execution of voluntary movements (Diedrichsen et al., 2003), speech production—especially at high speeds (Ackermann, 2008)—and in stuttering (Chang & Guenther, 2020; Neumann et al., 2018). However, opinions differ regarding the cerebellum's role in the treatment or improvement of stuttering. Kell et al. (2018) suggest that in cases where stuttering has recovered due to treatment, the cerebellum disconnects from the neocortical circuit. In contrast, Neumann et al. (2018) reported increased cerebellar activity following spontaneous stuttering recovery. The increased right cerebellar activity observed in our study group appears consistent with the findings of Neumann et al. (2018).

A few studies have explored the effect of therapy on functional brain communication, often using the fMRI technique (Kell et al., 2018; Korzeczek et al., 2021). To date, no study has examined the electrophysiological implications of treatment on functional connectivity, and only one study has compared fluent to stuttered speech in AWS in terms of phase coherence. Sengupta et al. (2017) reported an increase in the phase coherence of theta, alpha, and gamma bands at the left frontal electrodes while preparing for stuttered utterances. They attributed this increase to the heightened general activity associated with stuttering (Sengupta et al., 2017). Consequently, successful treatment may lead to a reduction in some measures of increased phase coherence. In comparing the two groups, the main feature associated with fluency in the study group was a significant decrease in greater phase coherence, which in the control group manifested in different bands and between electrodes related to motor-auditory-somatosensory areas of both hemispheres. Fluency in the control group was still associated with high phase coherence between different brain neural circuits, a characteristic of the dysfluency state (Sengupta et al., 2017). While the type and location of electrical stimulation were different and scattered across both hemispheres in the subjects of the study group, the control group had stimulation focused in the left inferior frontal region. Increased phase coherence in the delta band was observed in the left temporal-inferior parietal lobule (IPL) areas of the study group. One possible reason for the longer persistence of the behavioral results in the study group after 3 months may be the purposeful change in neural activities, leading to limited and targeted functional connectivity.

Signal Analyses Through the Condition of IS. In the study group, theta relative band power over the F3 and C3 regions was significantly higher than in the control group through the IS condition. Increases in power of frequencies below 10 Hz in the motor areas have been reported during both overt and covert normal speech (Bowers et al., 2019). In this context, the increase in theta band power may be interpreted as a successful strategy that has been reinforced in the study group. Although imagined speech does not involve any movement or generate re-afferent feedback, forward models of speech targets and expected sensory feedback are generated through a strictly internal loop (Tian & Poeppel, 2010, 2012). Since imagined speech engages the somatosensory and premotor areas, it is expected that these areas would differentiate between the two groups, shifting from the left IFG in the SP condition to coordination corresponding to the somatosensory cortex (C3) and the premotor area (F3) in the IS condition.

In the control group, an increase in beta relative power over the left and right IFG was observed during the Imagined stuttering (IS) condition. These findings align with the theory that defects in beta-band suppression are associated with stuttering. IS has been shown to correspond with overt stuttering in positron emission tomography (PET) studies (Ingham et al., 2000) and fMRI (Wymbs et al., 2013) in terms of regional brain activation. Additionally, Jenson et al. (2014) demonstrated that neural oscillations during syllable production and imagined speech are comparable. They later reported significantly decreased alpha and beta suppression in the left hemisphere in individuals with a stuttering trait during both overt and covert syllable production (Jenson et al., 2018). Consequently, the enhancement of beta suppression in the study group may be considered a marker of intervention success and is likely related to long-lasting outcomes in this group.

Regarding imagined speech, fluency was again associated with enhanced inter- and intrahemispheric FCs in the control group, similar to the SP condition. To our knowledge, no study has examined the effect of treatment on FC in the IS condition. However, the decrease in FC measures during the fluency state in this condition may also be interpretable within the framework of studies on speech preparation (Sengupta et al., 2017). Enhanced functional connectivity has been observed in various neurological disorders, such as autism (Mizuno et al., 2006) schizophrenia (Ding et al., 2019), and anxiety (Giménez et al., 2012). Considering the longer-lasting behavioral outcomes in the study group, the greater task-related FC in the control group suggests an open topic that requires further research.

Conclusion

Our findings suggest that individualized tDCS may be preferable to conventional tDCS, particularly regarding the maintenance of effects. Neural deficits associated with stuttering appear to be distributed across disparate networks. However, personalized intervention—through identification and stimulation of the individualized affected areas—likely led to similar activation patterns. We consider the increased activation of FC5, IFG, and the enhanced beta suppression over bihemispheric speech-related areas in the study group as evidence supporting this interpretation. Thus, the potentiated or inhibited neural networks may align with individual neural compensatory strategies or closely resemble expected and normal circuits. Nonetheless, we view this as a preliminary claim that requires further investigation.

Limitations and Future Directions

We recommend using a larger sample size to improve the accuracy of estimates. Additionally, we did not record a signal during the follow-up stage as per the predetermined study design. The current experimental design cannot separate the effects of tDCS from fluency-shaping training due to the absence of a sham group. Furthermore, we believe that utilizing the time-frequency spectrogram method, which takes advantage of the important benefits of EEG, may facilitate the exploration of protocols based on more efficient features. Our study acknowledges that EEG data can be contaminated during speaking tasks. Although we used the ADJUST algorithm for artifact removal, more advanced and better-validated methods, such as ASR, are now preferred. Furthermore, we did not assess the neural integrity of the components postcleaning with tools like IC_Label in EEGLAB, which is essential for ensuring data validity. Future studies should integrate these advancements to improve methodological rigor. In future studies, it is recommended that EMG traces be analyzed to effectively differentiate the components of the neural EEG signals, as this was not possible during the current study.

Author Declarations

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