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The Modulatory Effects of Transcranial Alternating Current Stimulation on Brain Oscillatory Patterns in the Beta Band in Healthy Older Adults

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Abstract: In the last few years, transcranial alternating current stimulation (**tACS**) has attracted attention as a promising approach to interact with ongoing oscillatory cortical activity and, consequently, to enhance cognitive and motor processes. Previous studies using tACS in the younger population have shown mixed results and cannot easily be transferred to the older population. In fact, the effects of tACS on cortical oscillations in healthy aging participants have not yet been investigated in great detail, particularly during movement. This study aimed to examine the after-effects of 20 Hz and 70 Hz High-Definition tACS (**HD-tACS**) on beta oscillations both during rest and movement. We found that 20-Hz HD-tACS induced a significant reduction in resting-state beta signals for electrodes C3 and CP3, while 70 Hz did not have any significant effects. With regards to Movement-Related Beta Desynchronization (**MRBD**), 20 Hz tACS led to more negative values, while 70 Hz tACS resulted in more positive ones for electrodes C3 and FC3. These findings suggest that HD-tACS can modulate beta brain oscillations with frequency-specific effects. They also highlight the focal impact of HD-tACS, which elicits effects on the cortical region situated directly beneath the stimulation electrode.

Keywords: tACS; aging; MRBD; EEG; movement; beta oscillations; non-invasive brain stimulation

1. Introduction

Advanced age often comes with a decline in sensorimotor control and functioning that affects the ability to perform activities of daily living. Indeed, it has been shown that movements become slower and/or less accurate and more cognition-dependent as we age [1]. Motor declines overall significantly impact motor independence, which is essential for older adults' quality of life and interactions with their environment. Additionally, behavioral evidence indicates that aging is frequently linked to slower movements, reduced capacity for learning new motor skills, and diminished ability to adjust a movement plan after initiation [2]. In fact, Mild Parkinsonian Symptoms (**MPS**), such as rigidity, bradykinesia, and tremor, are commonly diagnosed during clinical examination of older adults who do not have a diagnosed neurological disease [3].

Advances in neuroimaging techniques have made contributions to a better understanding of the aging brain. For instance, aging impacts brain structure such as a decrease in gray and white matter volume, along with an increase in cerebrospinal fluid in ventricles, fissures, and sulci [4,5]. These aging-related processes affect almost the entire cortex and underlying white matter, with a steeper decline in the primary motor cortex (**M1**) and frontal subcortical white matter [6]. For instance, aging is also associated with a complex pattern of atrophy [7], demyelination [8], free tissue water, and iron reduction within somatosensory and motor areas [6]. Moreover, age-related atrophy of motor cortical regions and corpus callosum has been shown to coincide with motor declines such as balance gait deficits and coordination deficits [9].

Sensorimotor cortex oscillations measured by electroencephalography (**EEG**) in the beta band (13-30 Hz) are a predominant feature of movement production and have been shown to be generated by local field potentials within the motor cortex [10]. Beta oscillations exhibit a robust pattern of movement-related changes, such as pre-movement beta Event-Related Desynchronization (**ERD**), Movement-Related Beta Desynchronization (**MRBD**), and post-movement beta Event-Related Synchronization (**ERS**) or beta rebound [11,12]. In terms of how these oscillations relate to motor performance, an association between MRBD and the accuracy at which subjects performed a bimanual task has been demonstrated, where subjects with more negative MRBD values exhibited worse task performance [13]. Greater MRBDs were also shown to correlate with a longer movement duration to complete a finger-tapping sequence [14]. In older adults, a greater (i.e. more negative) MRBD in both motor and premotor areas has been observed in subjects performing cued finger button presses [15,16] and handgrip tasks [13]. Additionally, older age has been associated with greater baseline beta power (15-29 Hz) at rest [17], suggesting that the alterations in brain structure and biochemistry during aging could be the reason behind the observed altered neural activation patterns. Given the association between movement production and beta band features, there is a high interest in modulating these oscillations non-invasively to improve motor ability and performance in older adults.

Transcranial alternating current stimulation (**tACS**) is a type of non-invasive brain stimulation method (**NIBS**) that can alter oscillatory brain rhythms through synchronization of neural networks in a frequency-dependent manner [18]. This technique is believed to entrain endogenous brain oscillations through the synchronization of two oscillatory systems occurring when a driving external oscillatory force coordinates with another oscillating system [19–22]. When applying tACS, the stimulation location, intensity and frequency need to be sleected, as they in turn determine its effects [23]. In terms of frequency, beta tACS (20 Hz) has been shown to slow movement in healthy individuals [21,24]. A reason for this may be that, in the cortex-basal ganglia circuit, beta activity is associated with promoting tonic rather than voluntary movement [25,26]. Also, motor impairments in Parkinson's disease (**PD**) have also been linked to elevated beta-band activity in the motor cortex and subthalamic nucleus [27]. In terms of intensity, it is generally set between 1 mA and 2 mA because it is well-tolerated, it has been shown to modulate cortex excitability and alter cognitive function [28]. Additionally, using higher intensities raise concerns about safety and side effects [29]. In terms of location, acquisition of motor skills is linked to a number of cortical and subcortical brain regions, but among these, M1 is thought to play a central role [30–33], making it a popular target for neurostimulation. Behaviorally, the use of 20 Hz tACS slows voluntary movement while 70 Hz tACS enhances motor learning along with an increase in beta-band power [24]. However, the influence of tACS on aging-related brain neural activity has not yet been studied.

Other NIBS techniques, such as transcranial direct current stimulation (**tDCS**) have been shown to have greater effects on motor performance when applied during a motor task, a technique often called *online stimulation*, compared to before the motor task [34]. There is also evidence that applying tDCS during practice triggers effects that outlast the stimulation period and facilitate neuroplasticity [28]. Previous studies have reported mixed effects regarding the effects of NIBS on young adults, and these results cannot be easily transferred to older adults. The stimulation sites and frequencies that modulate brain oscillatory activity in young adults may not result in the same effect in older adults, and functional reorganization of the aging brain may be an explanation [35].

In recent years, standard double electrode tACS showed limitations in controlling the stimulation focus and intensity. The use of different electrode montages, such as High-Definition tACS (**HD-tACS**), has allowed more precise stimulation control. Prior research has indicated that HD-tACS yields a more pronounced focalization of its effects through multiple smaller electrodes, possibly resulting from reduced distribution of the electrical field compared to conventional tACS [36,37]. Notably, online HD-tACS, applied during a motor task, induces phase- and frequency-dependent effects on the cortical excitability [38,39].

After-effects on brain oscillations are a common outcome following tACS [40]. For instance, 10 Hz tACS stimulation of the parieto-occipital area resulted in an enhancement of the EEG-recorded alpha amplitude during the stimulation and this effect was seen to last at least 30 minutes after a

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10-minute stimulation period [41,42]. Other NIBS techniques, such as tDCS, have induced long-lasting excitability elevations in the human motor cortex [43], and in animals, a stimulation period of 5 to 30 minutes causes an effect lasting for hours after the end of stimulation [44].

This study aims to explore the after-effects of 70 Hz and 20 Hz HD-tACS on beta brain oscillatory patterns in healthy older adults. Based on the previously mentioned effect of tACS, we hypothesized that 70 Hz tACS would reduce resting-state beta power and promote a more positive MRBD (lower desynchronization). Conversely, we hypothesized that 20 Hz tACS would decrease beta power at rest and induce more negative MRBD (higher desynchronization).

2. Materials and Methods

2.1. Participants

In this single-blinded sham-controlled study, 15 healthy individuals (7 males, 8 females) over 65 years old (age criteria as suggested by the Organization for Economic Co-operation and Development [45]) were recruited via advertisements. All participants signed a written informed consent and were compensated for their participation. Inclusion criteria included having right-hand dominance as assessed through The Edinburgh Handedness Inventory [46] and scoring higher than 3 in the Mini-Cog Test [47]. We excluded subjects who had a personal history of neurological and psychiatric disorders, had any contraindications related to tACS assessed through our NIBS safety questionnaire, and had received tDCS or tACS in the previous three months. Participants completed the following motor performance upper limb screening assessment tests: Box and Block Test (**BBT**) [48], Purdue Pegboard Test (**PPT**) [49], and Handgrip Strength (**HGS**) [50].

2.2. Experimental Design

The paradigm flow is shown in Figure 1. There were 3 experimental sessions in addition to the eligibility visit. A 64-channel EEG system (Brain Products, Germany) was used to collect data, and stimulation was delivered using an EEG-compatible HD-tACS device (Soterix Medical, NJ, USA). Baseline EEG at rest was recorded for 5 minutes, during which participants were seated in front of a screen displaying a centered white cross. They were asked to relax, look at the cross and stay as still as possible. During the handgrip task, participants held a grip force response dynamometer with their right hand. They were required to squeeze a hand-clench dynamometer (BIOPAC, Goleta, CA, USA) which produced a linear force measurement output based on the pressure applied with the hand. A blue bar moved up and down according to the gripping force produced by the participants, who were asked to reach a red bar higher up as fast and accurately as possible. The force required to reach the target was 15% of their maximum voluntary contraction (**MVC**), which they had to hold for 4 seconds with an interval of 8-10 seconds resting between each handgrip (Figure 2). The handgrips were repeated 50 times (10 min). The hand dynamometer was connected to a BIOPAC system that converted the input to electrical signals. The signals were then transferred to a recording computer that displayed the force that was being applied by the participant in real-time. Participants practiced until they understood the goal of the task (10 trials).

After the baseline recordings, participants received either 20-Hz, 70-Hz HD-tACS or sham stimulation for 10 minutes while repeating the 50 handgrip task as EEG signals were simultaneously recorded. The order in which participants received the type of stimulation was randomized. After the stimulation ended, participants were asked to fill out a questionnaire to monitor the following possible adverse effects of tACS: headache, neck pain, scalp pain, tingling, itching, burning sensation, skin redness, sleepiness, trouble concentrating, and acute mood change [51]. Participants were also asked if they thought the stimulation was active or sham to assess for protocol blindness. EEG recordings (rest and handgrips) were repeated 15 and 45 minutes after stimulation ended.

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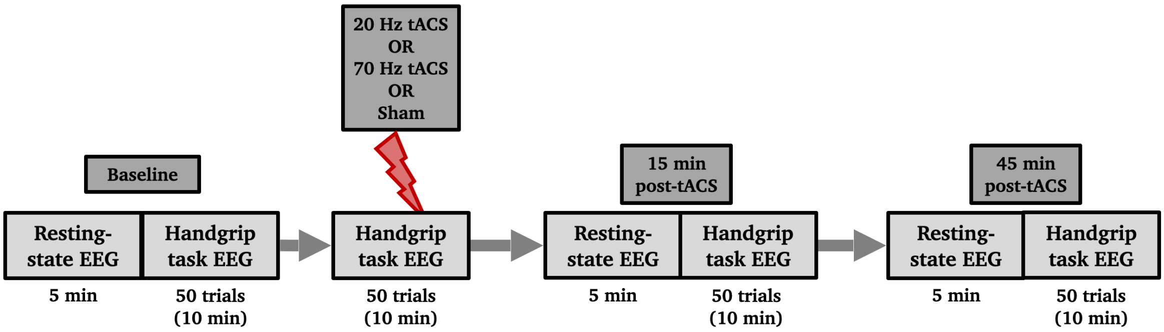

Figure 1. Schematic of the experimental timeline. Participants attended three different experimental sessions (each separated at least by a week). Two active HD-tACS sessions (20 Hz and 70 Hz) and one Sham (control) session. Sessions were counterbalanced across participants. Each session started with the Baseline EEG recording of 5 minutes of resting-state followed by 50 trials of the handgrip task. After that, active or Sham tACS stimulation was applied while performing another 50 trials of the handgrip task. Resting-state EEG and handgrip task EEG were performed again 15 minutes and 45 minutes post-tACS/sham.

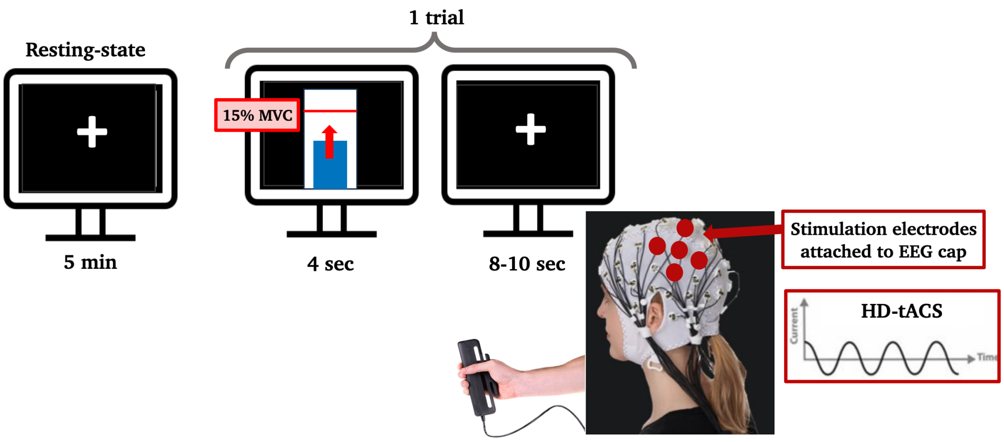

Figure 2. Computer screens showing the paradigm to the participant. During resting-state the participant looked at a black screen with a white cross in the center for 5 minutes. During the handgrip task, one trial consisted of reaching a threshold (red line inside the white bar), which was set to 15% of their maximum voluntary contraction, with a dynamometer using their right hand and staying on that threshold for 4 seconds. Each trial was followed by an inter-trial resting interval of 8 to 10 seconds. The stimulation electrodes that delivered HD-tACS were positioned on 5 recording electrodes over left M1 (anode: C3, cathodes: FC5, FC1, C3, CP5, CP1).

2.3. Data Acquisition and Pre-Processing

EEG signals were amplified and sampled at 2500 Hz. All electrodes were referenced to FCz. Electrode impedances were kept below 20 kΩ. HD stimulation was delivered by a current regulator (Soterix Medical, Germany). The EEG cap covered the individual's entire scalp, but the stimulation was delivered at pre-selected electrodes over the left sensorimotor cortex, with a target on M1 [28]. The anode was located in electrode C3 and the cathodes in FC5, FC1, CP5, and CP1. The stimulation lasted 10 minutes and it was delivered in the form of a sinusoid waveform with a peak-to-peak value of 1 mA and frequencies of 20 Hz and 70 Hz.

Offline EEG data were pre-processed using the Brainstorm MATLAB toolbox [52]. Electrodes with atypical power spectrum density were rejected for analysis. The rest of the EEG data were filtered (0.5 - 100 Hz bandpass, 60 Hz notch), resampled at 250 Hz, and re-referenced to an average reference. Noisy segments (e.g., muscle, head and jaw movement artifacts) were rejected by visual inspection.

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Independent components analysis (ICA) was used to identify and remove eye movement, muscle, and heart artifacts. Criteria for rejection included components' topography and time-history [53].

2.4. Data Analysis

2.4.1. Behavioural Scores

Desrosiers et al. [54] developed predictive equations for PPT scores based on normative data resulting from their study. The normative data portion of the study involved 360 healthy participants over the age of 60 years. Student's t-tests were used to determine if the predicted scores were significantly different from the scores obtained by our participants.

2.4.2. Resting-State EEG

Signals from the resting state recording were epoched in 5 s segments (n= 51 ± 5.95/recording). EEG signals were convoluted using a Morlet wavelet transformation with a frequency range from 1 Hz to 55 Hz in 1 Hz steps [55]. We further analyzed signals from electrodes FC3, FC1, C5, C3, C1, CP5, CP3, and CP1 since we were interested in the HD effects over M1 and surrounding areas. We excluded FC5 since it was a faulty electrode. The beta frequency was extracted (15 - 29 Hz) and averaged in this range to calculate the beta power at rest.

2.4.3. Motor Task EEG

Signals were epoched from 1 second before to 8 seconds after the appearance of the blue bar that triggered the initiation of the handgrips. The first 5 trials were rejected for each subject and each recording. Additionally, trials that were once again visually inspected and if still contaminated with artifacts they were manually rejected (n= 35 $\pm$ 6.18 trials/recording). EEG signals were then examined in the time-frequency domain using a Morlet wavelet transformation with a frequency range from 1 Hz to 55 Hz in 1 Hz steps [55]. Time-frequency maps were averaged within the beta band (15 - 29 Hz). MRBD was calculated as follows:

$$MRBD = \frac{P(t) - B}{B} x100\% \tag{1}$$

Where *P(t)* is the absolute power at time *t* and *B* is the mean power of the baseline, which was defined as 0.9 - 0.1 seconds before the start of each trial. MRBD% was averaged over 0.5 - 3.5 seconds after the appearance of the visual cue during sustained contraction at 15% MVC [56]. The analyzed signals were focused on the same electrodes as the resting-state analysis. The EEG signals recorded during active tACS were contaminated with very large artifacts, therefore they were not included in the analysis.

2.5. Statistical Analysis

Repeated-measures analysis (**rmANOVA**) tests were conducted for beta features during rest and movement with an a-level of 0.05 with factors Stimulation (20 Hz, 70 Hz and Sham) and Time (Baseline, post-15 min, and post-45 min). Post-hoc Bonferroni corrected [57] t-tests were also used to test for differences across time (before applying stimulation vs. 15- and 45-min post-stimulation).

3. Results

3.1. Behavioral Assessment

The results from the behavioural assessments are reported in Table 1. The predicted scores for each of the participants based on their age were calculated according to Desrosiers et al [54] (Table 2). There was no significant difference between the predicted scores of this model and the scores of our participants.

Table 1. Subject characteristics and behavioral scores

	Mean	Min - Mix
Sex	$7M/8F$	
Age (years)	69.7 ± 4.2	$65 - 78$
Handedness (/100)	94.6 ± 6.8	$80 - 100$
MiniCog (points)	4.7 ± 0.5	$4 - 5$
BBT Right Hand (blocks)	56.4 ± 5.5	$46 - 65$
BBT Left Hand (blocks)	57.2 ± 6.9	$43 - 65$
PPT Right Hand (pins)	13.4 ± 2.3	$9 - 17$
PPT Left Hand (pins)	12.0 ± 2.4	$6 - 15$
PPT Both Hands (pins)	10.4 ± 2.0	$6 - 14$
PPT Assembly (pins)	26.7 ± 5.1	$17 - 34$
HGS Right Hand (kg)	32.7 ± 11.3	$17.6 - 55.6$
HGS Left Hand (kg)	30.7 ± 9.5	$15.6 - 54.3$

Table 2. Participants' PPT scores (predicted from Desrosiers [54] model vs. real scores)

PPT subtests	Predicted	Real	*p*-value	*Cohen's d*
Right Hand	12.8 ± 0.8	13.4 ± 2.3	0.136	0.326
Left Hand	12.0 ± 0.7	12.0 ± 2.4	0.439	0.048
Both Hands	9.8 ± 0.6	10.4 ± 2.0	0.128	0.358
Assembly	26.7 ± 2.2	26.7 ± 5.1	0.493	0.005

3.2. Participant Blinding

After the end of the stimulation, participants were asked if they believed they had received active stimulation and, if they believed they did, at which frequency. For all the sessions, 33% of participants correctly identified sham stimulation, and 47% and 33% correctly identified the 20 Hz and 70 Hz active stimulation, respectively. Cochran's Q test did not reveal significant differences between conditions (p = 0.716).

3.3. Effects of tACS on Resting-State Beta Power

For resting-state beta power at electrode FC3, there was no main effect of *Time* (F = 0.751, p = 0.480), no main effect of *Stimulation* (F = 0.303, p = 0.740), and no *Time* X *Stimulation* interaction (F = 0.207, p = 0.933).

At electrode FC1, there was no main effect of *Time* (F = 0.800, p = 0.458), no main effect of *Stimulation* (F = 1.654, p = 0.209), and no *Time* X *Stimulation* interaction (F = 0.492, p = 0.741).

At electrode C5, there was no main effect of *Time* (F = 0.258, p = 0.773), no main effect of *Stimulation* (F = 0.666, p = 0.521), and no *Time* X *Stimulation* interaction (F = 1.535, p = 0.204).

At electrode C3, there was no main effect of *Time* (F = 0.795, p = 0.461), a significant main effect of *Stimulation* (F = 0.087, $\mathbf{p = 0.016}$), and no *Time* X *Stimulation* interaction (F = 0.429, p = 0.786).

At electrode C1, there was no main effect of *Time* (F = 0.229, p = 0.796), no main effect of *Stimulation* (F = 0.112, p = 0.893), and no *Time* X *Stimulation* interaction (F = 1.458, p = 0.227).

At electrode CP5, there was no main effect of *Time* (F = 0.751, p = 0.481), no main effect of *Stimulation* (F = 0.465, p = 0.632), and no *Time* X *Stimulation* interaction (F = 1.221, p = 0.312).

At electrode CP3, there was a main effect of *Time* (F = 3.958, $\mathbf{p = 0.030}$), no main effect of *Stimulation* (F = 0.137, p = 0.872), and no *Time* X *Stimulation* interaction (F = 0.849, p = 0.500).

At electrode CP1, there was no main effect of *Time* (F = 2.206, p = 0.128), no main effect of *Stimulation* (F = 0.371, p = 0.693), and no *Time* X *Stimulation* interaction (F = 0.586, p = 0.673).

Using post-hoc t-tests of C3 and CP3 (Bonferroni corrected) (Figure 3), spontaneous beta power significantly decreased only 45 minutes post-20 Hz tACS (***p* < 0.001**).

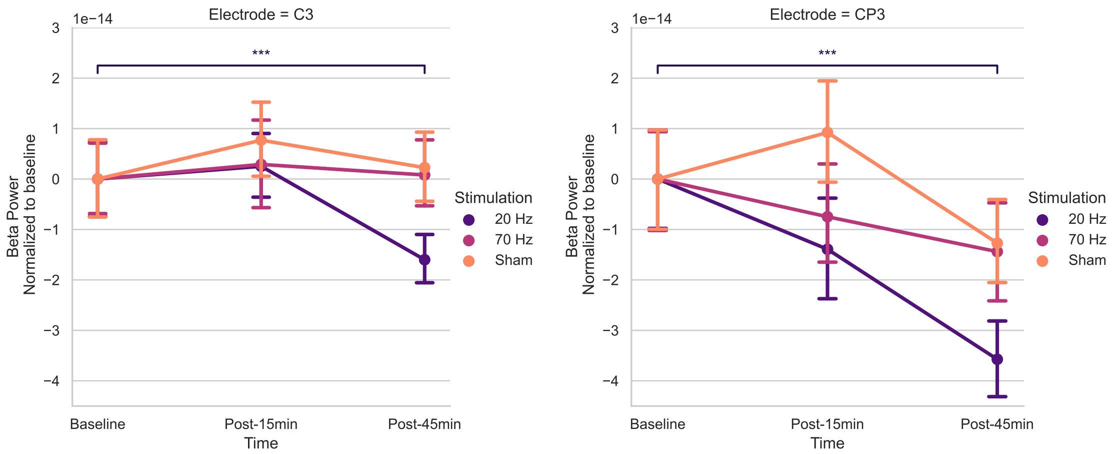

Figure 3. Effects of HD-tACS on average beta power at electrode C3 and CP3 normalized to baseline. Error bars represent SE. (*** = $p < 0.001$)

3.4. Effects of tACS on MRBD

For the results of rmANOVA on MRBD at electrode FC3, there was no main effect of *Time* (F = 0.035, p = 0.965), no main effect of *Stimulation* (F = 1.276, p = 0.294), and a significant *Time* X *Stimulation* interaction (F = 4.144, ***p* = 0.005**).

At electrode FC1, there was no main effect of *Time* (F = 0.577, p = 0.567), no main effect of *Stimulation* (F = 1.049, p = 0.363), and no *Time* X *Stimulation* interaction (F = 2.367, p = 0.063).

At electrode C5, there was no main effect of *Time* (F = 0.492, p = 0.616), no main effect of *Stimulation* (F = 0.866, p = 0.431), and no *Time* X *Stimulation* interaction (F = 1.896, p = 0.123).

At electrode C3, there was no main effect of *Time* (F = 0.616, p = 0.546), no main effect of *Stimulation* (F = 0.964, p = 0.393), and a significant *Time* X *Stimulation* interaction (F = 2.694, ***p* = 0.040**).

At electrode C1, there was no main effect of *Time* (F = 0.079, p = 0.923), no main effect of *Stimulation* (F = 1.759, p = 0.190), and no *Time* X *Stimulation* interaction (F = 1.271, p = 0.292).

At electrode CP5, there was no main effect of *Time* (F = 0.189, p = 0.828), no main effect of *Stimulation* (F = 0.827, p = 0.447), and no *Time* X *Stimulation* interaction (F = 1.561, p = 0.197).

At electrode CP3, there was no main effect of *Time* (F = 1.449, p = 0.251), no main effect of *Stimulation* (F = 1.877, p = 0.171), and no *Time* X *Stimulation* interaction (F = 2.156, p = 0.085).

At electrode CP1, there was no main effect of *Time* (F = 0.238, p = 0.789), no main effect of *Stimulation* (F = 0.340, p = 0.714), and no *Time* X *Stimulation* interaction (F = 0.686, p = 0.604).

Using post-hoc t-tests for electrodes FC3 and C3 (Bonferroni corrected) (Figure 5), a significant increase in MRBD percentage (more negative) was found 15 min post 20 Hz HD-tACS for both (FC3; ***p* = 0.039**, C3; ***p* = 0.011**). 70 Hz HD-tACS, elicited a decrease in MRBD percentage for both electrodes (FC3; ***p* < 0.001**, C3; ***p* = 0.039**), which persisted post-45 minutes only in FC3 (***p* = 0.036**). There were no significant differences in MRBD values during Sham stimulation for either of them (p > 0.05).

The timecourses for MRBD at electrodes FC3 and C3 at the three time points during the three stimulation sessions can be seen in Figure 4.

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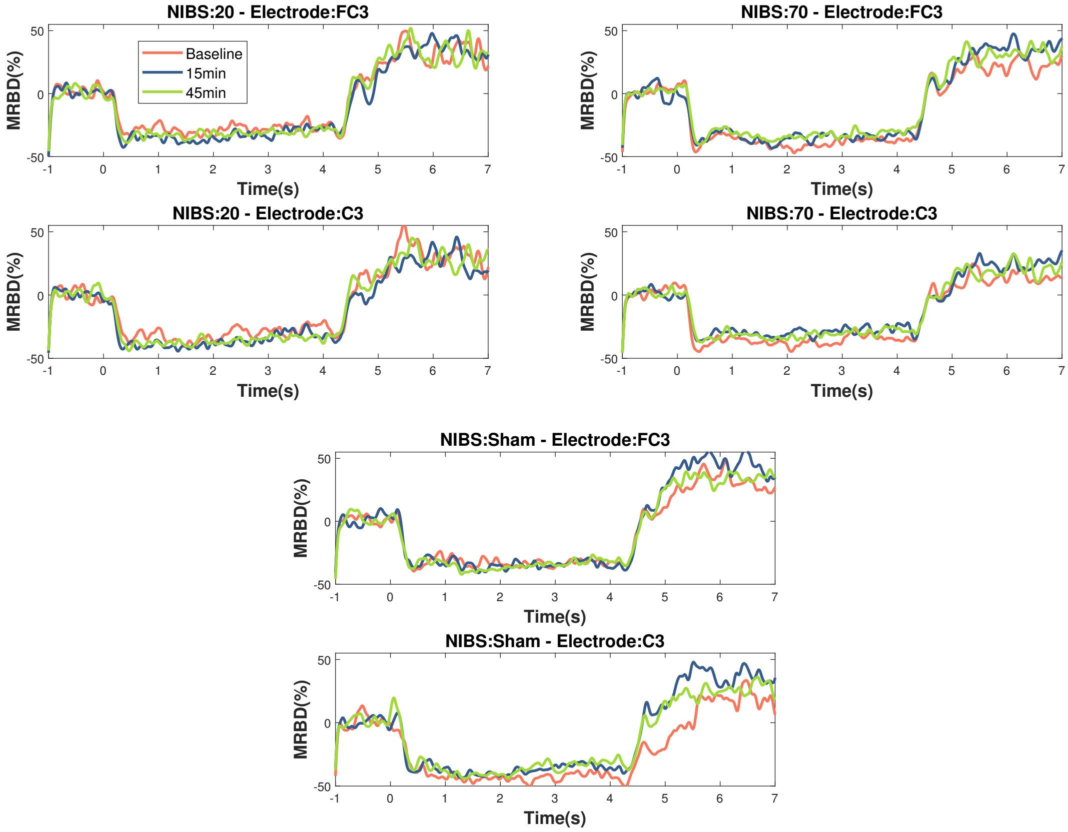

Figure 4. Temporal evolution of MRBD on electrodes that showed significant changes (C3 and FC3) across the different NIBS (20 Hz tACS, 70 Hz tACS and Sham). Time zero is motor task onset and Time 4 is motor task offset. 20 Hz HD-tACS induced a more negative MRBD only after 15 minutes and 70 Hz HD-tACS induced a more positive MRBD after 15 minutes in both electrodes and after 45 minutes only on FC3. No changes were significant during Sham stimulation.

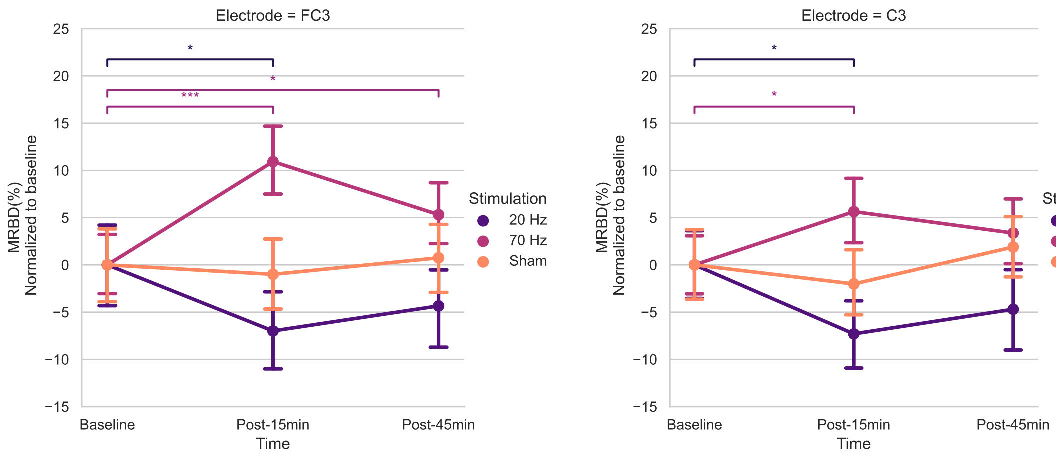

Figure 5. Effects of tACS on average MRBD percentage at electrode FC3 and C3 normalized to baseline. Error bars represent SE. (* = $p < 0.05$, *** = $p < 0.001$)

4. Discussion

This study aimed to quantitatively examine the after-effects of HD-tACS on electrophysiological features in healthy older adults. Our rmANOVA anaysis for beta power revealed a noteworthy main effect of *Stimulation* for C3, as well as a significant effect of *Time* for CP3. Conversely, no other electrodes displayed any statistically significant main effects. Upon post-hoc analysis of t-tests, a notable reduction in resting-state beta power was observed post-45 min for 20 Hz HD-tACS for both electrodes. In contrast, in the case of 70-Hz HD-tACS and Sham, no significant changes were observed in either electrode.

When analyzing MRBD effects, there was a significant *Time* X *Stimulation* interaction for FC3 and C3. The results of the post-hoc t-tests showed higher MRBD values (more negative) 15 minutes after 20 Hz HD-tACS in both electrodes. Applying 70-Hz HD-tACS resulted in significant reductions in MRBD values after 15 minutes in FC3 and C3 and after 45 minutes only in FC3. Notably, no significant changes were observed during Sham stimulation. In the subsequent subsections, the implications of these results on beta oscillatory patterns are discussed in greater detail.

4.1. Population Behavioral Scores and Baseline Features

To ensure the representativeness of our participants concerning motor performance, we employed the Desrosiers et al. [54] model to predict the PPT score for each participant based on their age when performing the motor tests during the eligibility session. The PPT quanitfies fingertip dexterity and gross movement of the hand, fingers and arm. The predicted and actual scores of the PPT were similar in all the subtests which suggests that the motor performance of our participants fell within the range of normal scores for the older population.

4.2. Modulation of Resting-State Beta Power

After administering 20 Hz HD-tACS, a delayed reduction in beta power was observed for electrodes C3 and CP3. This modulation appeared only 45 minutes post-stimulation, with no change detected at 15 minutes, suggesting that the impact on beta power is gradual rather than immediate. This finding aligns with previous studies showing no after-effects on beta power within 5 to 20 minutes after 20 Hz tACS [58]. However, prior research demonstrated that tACS could have effects on endogenous EEG power in the range of the stimulation frequency up to 70 min after the stimulation [59], which guided our decision to monitor changes up to 45 minutes post-stimulation. The delayed response may reflect plasticity-like mechanisms emerging over time, consistent with reports of late plasticity-like changes in corticospinal excitability following 20 Hz tACS [60]. Additionally, delayed effects induced by tACS may be attributed to spike-timing-dependent plasticity (**STDP**) [61]. According to STDP principles, synapses in circuits that resonate at frequencies similar to repetitive inputs are strengthened during stimulation. After the stimulation ends, these synaptic modifications continue, leading to increased neural activity at the circuits' resonant frequencies. Prior research has reported that beta tACS can sustain elevated beta oscillations for at least 60 minutes post-stimulation [62].

Our results agree with recent research showing that 20 Hz tACS increased beta power following stimulation of the visual [63] and parietal cortex [64]. However, a study reported no significant effects when applying tACS at 20 Hz, also in M1 during rest, on the beta power in younger participants using the standard double electrode tACS montage [65]. We can identify three possible reasons why our results differ from the aforementioned study. First, the difference in age of our participants, as it has been shown before that older groups demonstrated a decrease in tACS-induced neuroplasticity compared to a younger cohort [66]. We chose an older population, since these tACS frequencies have not been studied before in aging, and we specifically examined its effects on beta oscillations. Secondly, in our study, tACS was applied while participants were performing a handgrip task (online) as opposed to tACS during rest (offline). Differences have been observed across studies comparing tACS-induced changes in online and offline protocols [41]. Additionally, a recent study comparing online and offline HD-tDCS showed that only the online stimulation reduced the power of alpha rhythm during motor skill execution [67]. Thirdly, we used a different electrode montage and it has been shown that a HD-tACS electrode montage delivers a more focal current to M1 than the standard double electrode tACS montage [68,69]. It is important to note that beta-tACS has been shown to have mixed results on other outcomes, such as corticospinal excitability and motor function [70]. Therefore,

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more research should focus on applying tACS at this frequency, while keeping similar parameters to the ones used in studies showing significant effects.

For 70 Hz, there were no changes in resting power, which does not agree with our initial hypothesis. Sugata *et al.* [24] found that his stimulation frequency increased beta power. However, our result does agree with Mastakouri *et al.* [71], who investigated the changes in beta power resulting from gamma-tACS. They showed that these changes only attained statistical significance among a subgroup of subjects who exhibited a behavioural response to the stimulation. This could be explained by recent studies which have also reported that a tACS delivered at a subject-specific frequency has a stronger effect on cortical oscillations and behavior relative to both sham and fixed frequency tACS [61,72–75]. Furthermore, the mechanisms underlying the effects of gamma stimulation on long-term potentiation (**LTP**)-like plasticity are less pronounced in older adults compared to younger adults [76]. Interestingly, gamma-tACS has been found to enhance general motor skill consolidation in older participants, whereas its impact on younger individuals is minimal [76]. Overall, the effects of tACS on brain oscillations are different in pathological groups when compared to healthy participants [77]. These findings suggest that protocols tailored to individual neural characteristics may yield more favourable effects compared to applying standardized parameters across the entire population.

4.3. Modulation of MRBD

We found that after 15 minutes of 20 Hz HD-tACS, MRBD values became more negative in FC3 and C3. Conversely, after 15 minutes of 70 Hz HD-tACS, MRBD values became more positive in both FC3 and C3, and the effect persisted after 45 minutes only in FC3. These results align closely with our initial hypothesis. To the best of our knowledge, there are no other studies investigating the specific effects of tACS on MRBD values in older adults. A recent study demonstrated that tACS at 10 Hz enhanced MRBD during a motor imagery task compared to pseudo-stimulation, indicating the capability of tACS to modify movement-related brain oscillations [78]. Additionally, other forms of NIBS, such as tDCS, have been shown to induce more negative MRBDs during motor imagery after 15 minutes of stimulation.

Xifra-Porxas *et al.* [13] previously mentioned that the motor performance decline observed in healthy aging may not be due to an impairment in the capacity to modulate beta oscillations. In fact, they observed a larger modulation in older compared to younger adults. On the other hand, beta oscillations at rest are greater in older adults [17], which suggests that increased desynchronization is needed to reach a threshold to initiate a movement. This would mean that modulating this desynchronization could later translate into a change in motor performance. Nonetheless, this is only an hypothesis and future research that looks both at MRBD and motor performance modulation should validate it.

Finally, the fact that electrode C3, the anode of our HD-tACS montage, showed significant changes in both resting-state beta power and MRBD values, suggests a focalization of the current right on the electrode that delivers the current. This focalization possibly results from a reduced distribution of the electrical field compared to the conventional tACS, which utilizes two distant patch electrodes [36].

Taken together, the results of this study were mostly aligned with our initial hypothesis. In terms of beta power, only 20 Hz tACS showed the expected increase in beta power, however, only 45 minutes after end of stimulation, while 70 Hz tACS didn't show any significant changes, contrary to our hypothesis that it would decrase beta power. Regarding MRBD, both 20 Hz and 70 Hz showed the expected results, the former resulted in a more negative MRBD and the latter in a more positive. However, these changes in beta oscillations may not necessarily translate to improvements in motor performance. Future research should focus on assessing the impact of tACS on motor performance using tasks that are more complex and reflective of real-world daily activities, particularly for older adults.

There are certain limitations in our study that may explain some of variability in our tACS outcomes and previously reported outcomes, such as interindividual differences, including skull thickness and the actual amount of current that reaches the cortex [79,80]. Because of these differences, individualized stimulation frequencies and current amplitudes, validated by studies such as Yamaguchi *et al.* [81], emphasize the parameter-dependent nature of tACS effects. Besides, our limited stimulation duration (10 minutes) contrasts with longer-lasting effects seen in extended gamma-tACS in mice, suggesting prolonged sessions or multiple-day approaches akin to tDCS studies for in-depth exploration [44,82]. Finally, the analysis of ongoing brain signals during concurrent stimulation is of primary interest and would also shed light on the immediate effects and mechanisms of tACS. Recent advances in artifact-removal algorithms will enable this type of analysis in our data [83].

5. Conclusions

To our knowledge, this is the first HD-tACS study that looks at the beta oscillation effects on healthy older adults. Future research should focus on replicating protocols that have been shown to have an effect on the desired outcomes on a bigger cohort to establish a more robust effect. In summary, our study reveals that HD-tACS has a modulating effect on beta oscillations during movement. Notably, different HD-tACS frequencies led to specific alterations in MRBD values, indicating frequency-specific effects on movement-related brain oscillations. The focal impact observed at electrode C3, the site of HD-tACS anodal stimulation, underscores the technique's precision over brain regions related to motor control. Future studies should explore personalized protocols tailored to individual neural characteristics, potentially employing advanced tACS methods such as HD-tACS, phase-shifted tACS, amplitude-modulated tACS, temporally interfering, and intersectional short pulse techniques. Addressing these intricacies will enhance our understanding of tACS efficiency, guiding its optimized application in clinical settings. This research not only contributes to the ongoing discourse on brain stimulation but also holds promise for therapeutic interventions, making tACS a promising avenue for exploration in diverse clinical populations such as stroke and PD patients. Further investigations in these domains will unveil the full potential of HD-tACS as a targeted therapeutic tool. Exploring HD-tACS effects in healthy older adults is crucial in providing insights into designing targeted interventions, addressing the complex interplay between neural oscillations, brain stimulation, and motor performance in the aging population.

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Data Availability Statement: The raw data supporting the conclusions of this article will be made available by the authors on request. The code is publicly available in the following repository: https://github.com/kenyamelissamf/tACS_Aging.

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Abbreviations

The following abbreviations are used in this manuscript:

BBT	Box and Block Test
CMC	Corticomuscular Coherence
ECR	Extensor Carpi Radialis
EEG	Electroencephalography
EMG	Electromyography
FDI	First Dorsal Interosseous
GABA	Gamma-aminobutyric Acid
HD-tACS	High Definition Transcranial Alternating Current Stimulation
HGS	Handgrip Strength Test
iAPF	Individual Alpha Peak Frequency
ICA	Independent Component Analysis
ISP	Intersectional Short Pulse
LTP	Long-term Potentiation
M1	Primary Motor Cortex
MEG	Magnetoencephalography
MEP	Motor-Evoked Potential
MPS	Mild Parkinsonian Signs
MRBD	Movement-Related Beta Desynchronization
MVC	Maximum Voluntary Contraction
NREM	Non-rapid Eye Movement
NIBS	Non-Invasive Brain Stimulation
PD	Parkinson's Disease
PPT	Purdue Pegboard Test
rmANOVA	Repeated-Measures Analysis of Variances
SICI	Short-Interval Intracortical Inhibition
tACS	Transcranial Alternating Current Stimulation
tDCS	Transcranial Direct Current Stimulation
TI	Temporally Interfering
TMS	Transcranial Magnetic Stimulation

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