Modulation of interhemispheric alpha-band connectivity by transcranial alternating current stimulation

Bettina C. Schwab[1*], Jonas Misselhorn[1], Andreas K. Engel[1]

1 Department of Neurophysiology and Pathophysiology, University Medical Center Hamburg-Eppendorf, Hamburg, Germany

* b.schwab@uke.de

Short title: *Bifocal tACS modulates phase coupling*

Abstract

Long-range functional connectivity in the brain is considered fundamental for cognition and is known to be altered in many neuropsychiatric disorders. To modify such coupling independent of sensory input, non-invasive brain stimulation could be of utmost value. In particular, transcranial alternating current stimulation (tACS) has been proposed to modulate oscillatory coupling, while evidence for frequency- and space-specific modification of connectivity is missing so far. Therefore, we investigated the aftereffects of bifocal high-definition tACS at 10 Hz on alpha-coupling. Healthy participants were stimulated in counter-balanced order (1) in-phase, with identical electric fields in both hemispheres, (2) anti-phase, with phase-reversed electric fields in the two hemispheres, and (3) jittered-phase, generated by subtle frequency shifts continuously changing the relative phase between the two fields. Global pre-post stimulation changes in EEG connectivity were larger for in-phase stimulation than for anti-phase or jittered-phase stimulation. The differences in connectivity change were restricted to the stimulated frequency band and decayed within a time window of 100 s after stimulation offset. Source reconstruction localized the maximum effect between the stimulated occipito-parietal areas. In conclusion, the relative phase of bifocal alpha-tACS

determined alpha-band connectivity shifts between the targeted regions. We thus suggest bifocal high-definition tACS as a tool to specifically modulate long-range cortico-cortical coupling which outlasts the electrical stimulation period.

Author summary

Transcranial alternating current stimulation (tACS) delivers rhythmically varying electric currents to the head with the goal of non-invasively stimulating neural tissue. Recently, tACS was proposed to synchronize neural oscillations locally and, when applied concurrently at multiple sites, between remote brain areas. Dynamic coupling of neural signals is considered essential for normal functioning of the brain, and its modulation may carry great potential for both research and clinical use. Nevertheless, frequency- and space-specific modification of functional connectivity by tACS still remains to be shown. With an optimized stimulation montage and reconstruction of neural sources from EEG signals, we were able to demonstrate effects of tACS on functional connectivity in a time window up to 100 s after stimulation offset. In particular, the effects were specific to the stimulated frequency and focused on the stimulated cortical areas. tACS may therefore help to manipulate functional connectivity in experimental settings to probe claims on brain function. Moreover, abnormal coupling in the diseased brain could in future studies be targeted by repetitive stimulation.

Introduction

Oscillatory activity and its synchronization at various temporal and spatial scales are regarded as crucial for cognition and behavior. In particular, oscillatory coupling has been suggested to functionally link remote brain areas by synchronously opening windows for communication [1, 2]. During sensory stimulation, large-scale connectivity, as measured by EEG or MEG, may thus form task-related networks within cortex [1, 3, 4]. But also ongoing resting-state connectivity, emerging from brain-intrinsic factors rather than external stimuli, was hypothesized to reflect physiological brain function and to bias processing of upcoming stimuli [2].

Due to their prominence in the resting-state EEG, α-oscillations and connectivity in the α-band have been studied extensively. α-oscillations were typically related to functional inhibition [5] and were proposed to route information to task-relevant regions [6]. Accordingly, phase synchronization of α-oscillations may have a crucial role in orchestrating the exchange of information in the brain [7] relevant for attention, memory, and executive functioning [5, 8]. Reduced or increased α-coupling in disease could therefore be related to cognitive impairment [9–11].

Despite the many studies demonstrating correlations between α-coupling and aspects of cognition, it is hard to find evidence for a causal relation based on EEG or MEG measurements alone. In contrast, non-invasive brain stimulation might be able to directly interfere with cortical dynamics. Especially transcranial alternating current stimulation (tACS), applied at different brain sites, was suggested to frequency-specifically modulate coupling between cortical areas [12–14]. tACS could thus be a valuable tool to investigate the role of functional connectivity for cognition, and, in later stages, to re-adapt pathological coupling in patients towards a physiological level.

Yet, up to now it is unclear what the immediate as well as outlasting effects of tACS on cortical electrophysiology are, and, critically, whether they are specific to the stimulated frequency band and cortical area. While α-tACS can lead to an increase in α-power after the offset of stimulation [15–18], much less is known about modulation of coupling. Behavioral changes under tACS have been described and were in some studies related to connectivity modulation [12–14, 18–20]. However, so far, conclusive evidence for connectivity modulation is missing, as the analysis of EEG and MEG data is impeded by a large, hard-to-predict stimulation artifact [21, 22]. Connectivity analyses of stimulation-outlasting effects were often confounded by power effects or were restricted to sensor level analyses limiting spatial resolution.

Hence, we present an in-depth investigation of cortical large-scale synchronization after bifocal α-tACS. To bypass the stimulation artifact, we analyzed resting-state EEG data before and after tACS (Fig. 1A), with conditions differing in the relative phase of the applied field (Fig. 1B). In particular, all tACS conditions involved α-stimulation with comparable power distributions of the electric fields (Fig. 1C), avoiding different aftereffects in α-power or different tactile sensation. We were able to

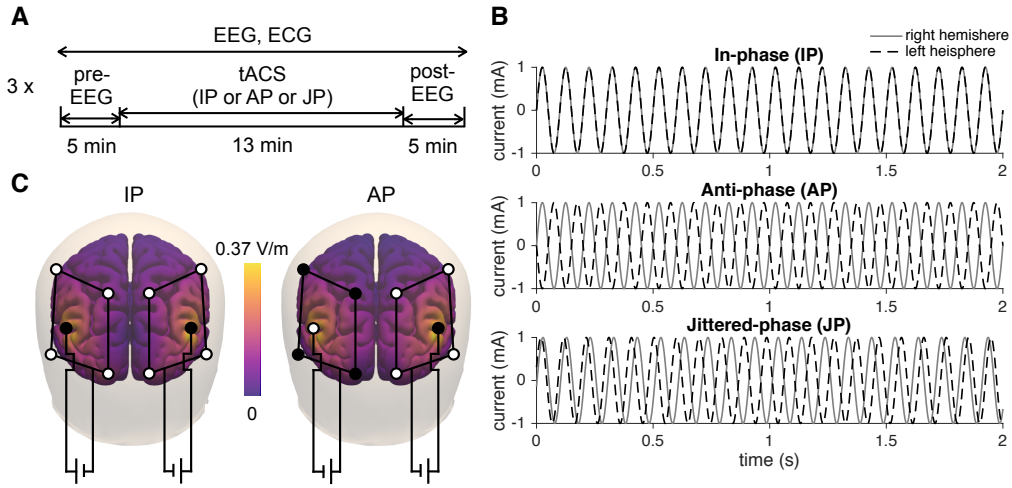


Fig 1. Experimental setup. A: Timeline of resting-state recordings. EEG and ECG were recorded before and after tACS in three sessions. The sequence of tACS conditions was counterbalanced. B: Stimulation current waveforms of the three tACS conditions, applied on the right hemisphere (gray) and left hemisphere (black dashed), shown for 2 s. C: Peak power, for example at time point 25ms, of bifocal occipito-parietal stimulation fields for IP and AP stimulation. Magnitudes of the applied electric fields were similar for IP and AP stimulation. In contrast, the direction of the fields differed between conditions, with IP stimulation generating identical fields within each hemisphere and AP stimulation generating fields of opposite direction within the two hemispheres.

frequency-specifically modulate functional α-coupling between the stimulated regions for up to 100 s after tACS offset, opening the possibility to use tACS as a powerful manipulator of functional connectivity.

Results

α-Coupling is modulated by the phase of bifocal high-definition tACS

We hypothesized that the stimulation conditions, characterized by the relative phase of the two applied fields, differently affected functional connectivity. To assess global connectivity at sensor level, we studied the overall population of change in imaginary coherence between pre- and post-stimulation intervals. Cumulative histograms comprising data from all participants and electrode pairs were analyzed for the stimulated frequency band (9-11 Hz; Fig. 2). Within the first 100 s after tACS offset, global pre-post change in α-connectivity was larger for in-phase (IP) stimulation than for anti-phase (AP) stimulation, while jittered-phase (JP) stimulation led to

intermediate connectivity changes for most pairs (Fig. 2A). Only for high positive connectivity changes, the fraction of pairs was lowest in the JP condition. In later time windows (Fig. 2B,C), condition differences already decayed.

Significance was assessed for grand average condition differences in imaginary coherence change (Fig. 2D) and the Kolmogorov-Smirnov (K-S) distance between the distributions of imaginary coherence change (Fig. 2E). Both measures reached significance (p <0.05 after Holm-correction for 3 multiple stimulation comparisons) for the contrasts IP-AP and IP-JP within the first time window (0-100 s). Power differences between conditions were absent for this time window (inset in Fig. 2E). As the effect lost significance already for the time window 20-120 s, we conclude that changes in connectivity were restricted to the first 100 s after stimulation offset. In sum, we found transient global connectivity changes in the stimulated frequency band dependent on the stimulation condition. The effects were independent of power fluctuations and did not significantly interact with individual physiology (see Fig. S2).

Frequency-specificity of effects

We found robust condition differences in α-coupling within 100 s after stimulation offset - but are these effects restricted to the stimulated frequency band? To test effects across different frequency bands, we computed the grand average change in imaginary coherence (Fig. 3A) as well as the K-S distance between the cumulative histograms of imaginary coherence change of two conditions (Fig. 3B). For both measures, the largest effects in contrasts IP-AP and IP-JP were observed around the stimulated 10 Hz - a frequency band free of power effects (inset in Fig. 3B). All in all, effects were focused on the stimulated α frequency band, although variability and thus confidence intervals were largest in the lower α-band. Also power effects were restricted frequencies around 8 Hz.

Localization of effects in source space

As connectivity is difficult to localize at sensor level [25], we used an eLORETA source projection of the EEG data to study the spatial extent of condition differences within the first 100 s after stimulation offset. On sensor level, the contrast IP-AP showed clear connectivity effects and was not confounded by α-power effects (Fig. 2); hence, we

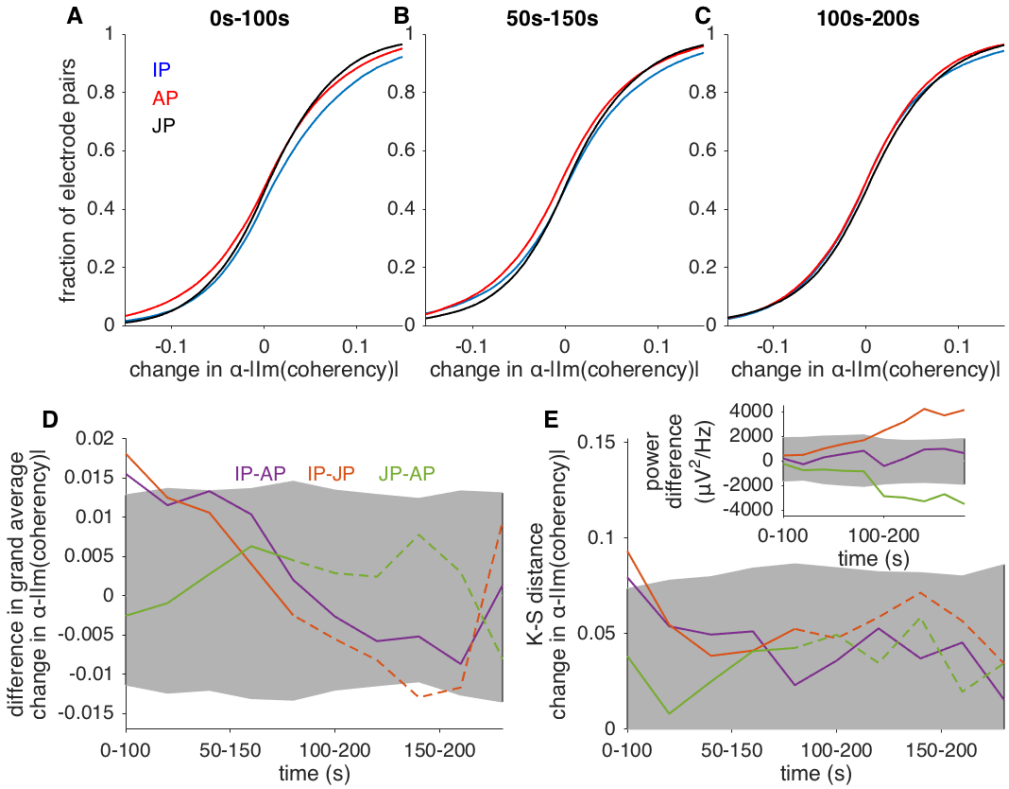


Fig 2. Global α-connectivity differs between tACS conditions within the first 100 s after tACS offset. Histograms (A-C) show the cumulated fraction of α-imaginary coherence change from all participants and electrode pairs. Right-shifted curves thus indicate high global connectivity change compared to left-shifted curves. For the time interval 0-100s (A), IP stimulation increased connectivity compared to AP stimulation, whereas JP stimulation was associated with intermediate connectivity values. The difference decayed with time (B,C). D: Time course of grand average differences in α-imaginary coherence change. 95% confidence intervals Holm-corrected for multiple stimulation condition comparisons are shown in gray. The difference between IP and AP (purple) as well as between IP and JP stimulation (orange) were significant within the first time interval, 0-100 s after stimulation offset. Dashed lines indicate confounding power effects (see inset in E). E: Differences between the distributions, quantified by the K-S distance, decayed similar to grand average changes. Inset: Differences in α-power change between conditions. No significant differences were found for early intervals. For later intervals, JP stimulation decreased α-power compared to IP and AP stimulation.

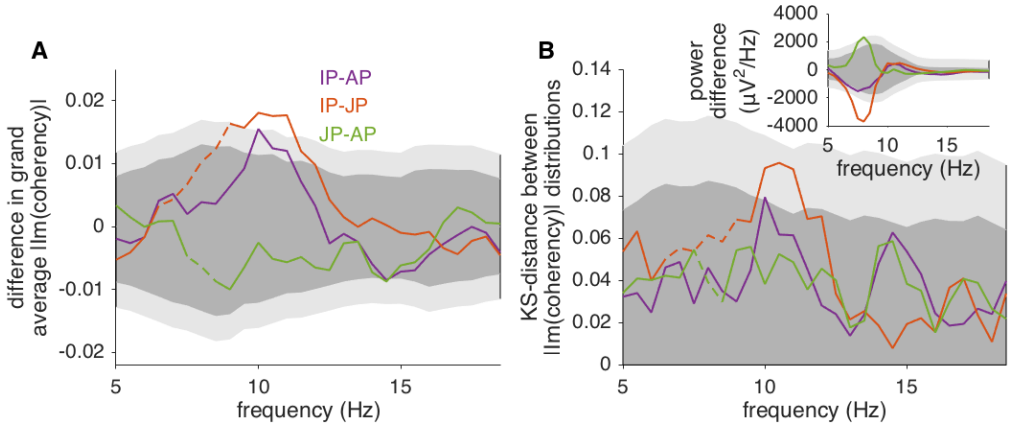


Fig 3. Connectivity effects in the first 100 s after stimulation offset are restricted to the stimulated frequency band around 10 Hz. A: Grand average differences in imaginary coherence for frequency bands of 2 Hz width, shifted in steps of 0.5 Hz. B: K-S distances for the same frequency bands. Dark gray: 95 % confidence intervals Holm-corrected for multiple condition comparisons. Light gray: 95 % confidence intervals additionally corrected for multiple frequency comparisons. Dashed lines indicate confounding power effects. Inset: Differences between power changes for the first 100 s after tACS offset were non-significant in the 10 Hz range.

focused on α-connectivity changes between these two conditions (Fig. 4). Effects were most prominent between the stimulated occipito-parietal regions, marked by dashed lines (Fig. 4A).

Connectivity differences at interhemispheric pairs between the stimulated regions (black dashed box; "stimulated pairs"), were in 34.7% significant (uncorrected $p < 0.05$), while this was the case for only 0.19% of all unstimulated pairs. Average values in coupling differences were about an order of magnitude larger for stimulated pairs (0.026 ± 0.017) than for unstimulated pairs (0.0023 ± 0.011; $p < 0.0001$, two-sample t-test). Differences reached their maximum between left and right superior occipital gyrus. P-values for differences in connectivity change at stimulated pairs were additionally Holm-corrected for 49 region comparisons. After Holm-correction, four region pairs remained significant (corrected $p < 0.05$; white dots in Fig. 4A).

To rule out the possibility that the observed local connectivity changes could be related to local power changes, we also computed local power change between IP and AP stimulation (Fig. 4B). No significant power differences were observed, in particular also not in occipital regions, where interhemispheric coupling changes were largest. Furthermore, power effects might also influence intrahemispheric coupling, but we did not find prominent effects with the right or the left stimulated region. Thus, we do not

expect power differences to drive differences in connectivity.

Looking at the population of stimulated pairs, we again computed cumulative histograms of imaginary coherence for all conditions (Fig. 4C). As in sensor space, IP stimulation increased connectivity compared to AP stimulation, while JP stimulation induced connectivity changes intermediate compared to the ones for IP and AP stimulation. All condition contrasts were significant in a permutation test (Fig. 4D). Taken together, we found differences in connectivity change between IP and AP stimulation to be focused on the stimulated regions. Interhemispheric pairs between the stimulated regions showed similar condition differences as global sensor level data.

Effects spatially correlate with stimulating field

Changes in coupling between interhemispheric homologue areas were of particular interest in our study, since these areas received comparable stimulating fields in the IP condition, and comparable fields with opposite directions in the AP condition. Thus, we further examined α-coupling differences between homologue regions (Fig. 5A), power differences associated with these regions (Fig. 5B), as well as vectorial and absolute differences in the stimulating fields between homologue regions (Fig. 5C). Note that we show average properties of regions; peak field differences can be higher and reached 0.64 V/m for vectorial differences (see Appendix S1).

As predicted, large changes in α-coupling between homologue areas were associated with large vectorial differences in field strength (Fig. 5D). In contrast, those coupling changes did not correlate with power changes (Fig. 5E), nor did the power changes correlate with vectorial differences in field strength (Fig. 5F). Therefore, we suggest phase differences in the tACS stimulus to drive changes in α-coupling between stimulated regions, but not α-power.

Discussion

To our knowledge, this is the first study to describe frequency- and space-specific aftereffects of tACS on oscillatory coupling. With an optimized bifocal high-definition tACS montage and stimulation conditions differing in the relative phase of the two fields, we were able to modulate phase coupling at the stimulated frequency between the

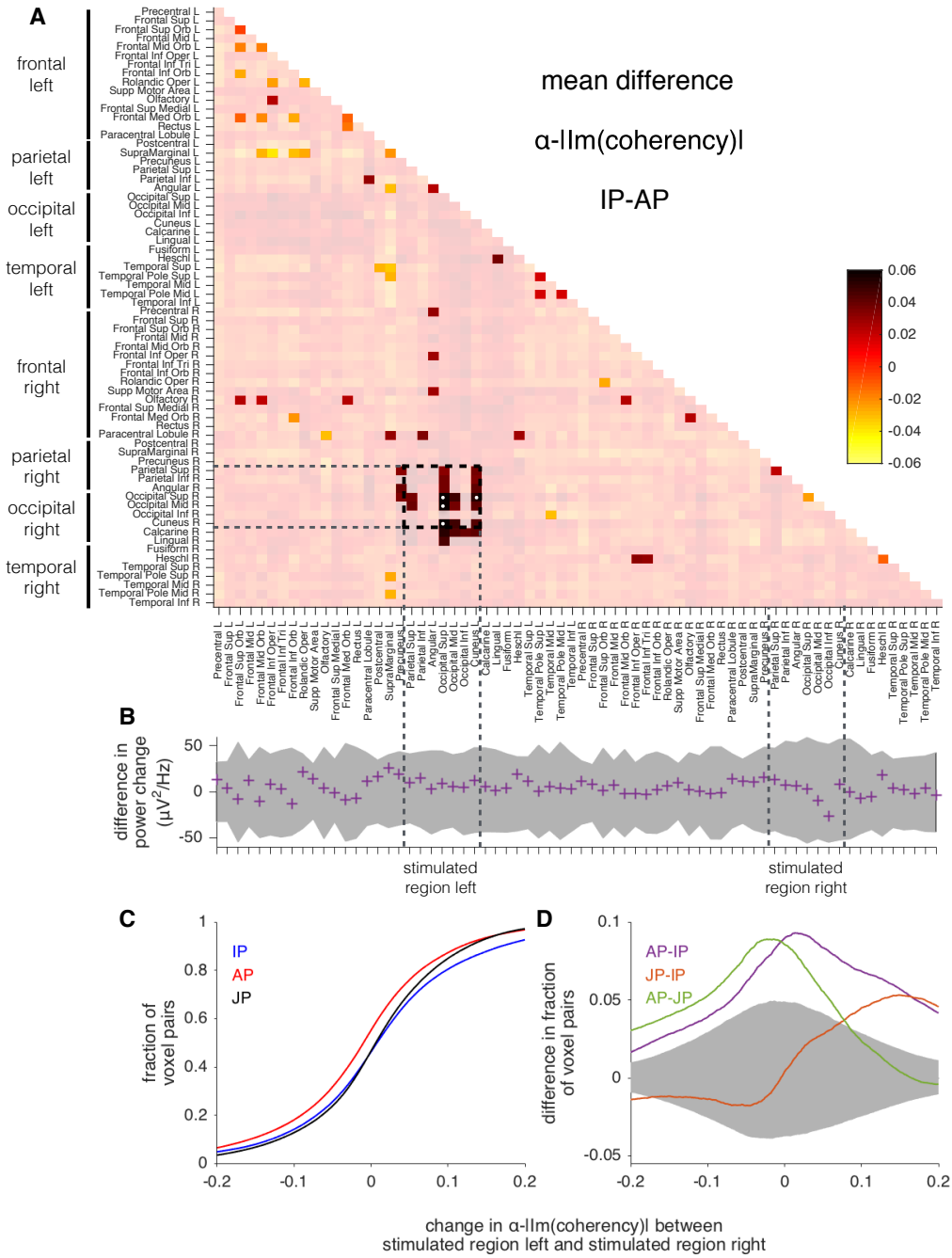


Fig 4. Source level characteristics of average change in α-connectivity within the first 100 s after stimulation offset. A: Connectivity matrix of average α imaginary coherence change, contrasting IP and AP stimulation conditions. Non-significant values (uncorrected $p > 0.05$, permutation test) were shaded. Dashed vertical lines depict the stimulated regions, defined as an average vectorial difference in field strength above 0.08 V/m. P-values for coupling differences between the stimulated regions (black dashed box) were additionally Holm-corrected for 49 multiple regions comparisons. White dots indicate $p < 0.05$ after Holm-correction. B: Differences in α-power change between IP and AP stimulation. No region showed significant power effects. Regions with significant coupling differences were not associated with particularly large power differences. C: Cumulative histograms of change in imaginary coherence for all pairs between the two stimulated regions. D: Difference in cumulative histograms shown in C. All condition differences extended beyond the 95% confidence limit (dark gray, Holm-corrected for three condition contrasts).

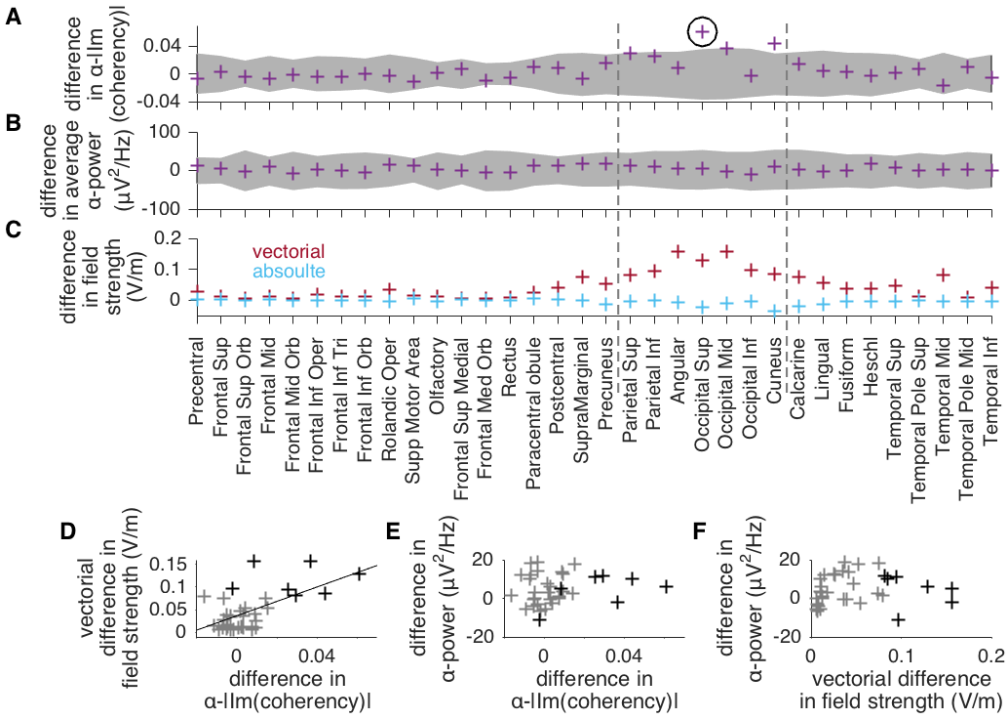


Fig 5. Source level change in α-connectivity between interhemispheric homologue areas spatially correlates with vectorial field differences. All data shown was restricted to the first 100 s after tACS offset and the contrast IP-AP. A: Changes in imaginary coherence differences between interhemispheric homologue areas for the α-band (9-11 Hz). Uncorrected 95% confidence intervals are shown in dark gray. Coupling between the superior occipital gyri remained significant after Holm-correction for 49 multiple region comparisons (black circle; see also Fig. 4). B: Differences in power change, averaged over interhemispheric homologue areas. C: Differences in the applied electric field between stimulation conditions IP and AP, shown as interhemispheric differences. The montage was optimized such that differences in absolute field stength (blue) became small, while vectorial field differences were large for parietal and occipital regions (red). The stimulated region is marked by dashed vertical lines. D: The vectorial difference in field strength correlated with change in imaginary coherence (p <0.001). Points representing stimulated regions are shown in black. E: In contrast, power differences did not correlate with changes in coupling. F: Power differences neither correlated with vectorial field strength, indicating the missing influence of tACS phase on α-power.

stimulated regions. Both for neurophysiological research, testing causal relations between coupling and behavior, as well as for clinical research, targeting pathological coupling, our results are of importance, since specificity in frequency and space is essential for all of those applications.

Several preceding studies already suggested tACS as a tool to modify coupling in the brain [12–14, 18–20, 23, 24], but interpretation of connectivity in the stimulated frequency band has remained problematic. Online recordings during application of tACS [14] are subject to nonlinear stimulation artifacts which are difficult to estimate [21, 22]. While work combining tACS with fMRI circumvents those large artifacts and revealed tACS to modify coupling within the motor network, its investigation is only possible at a time scale of seconds or slower [23, 24]. Offline analysis of EEG aftereffects [12–14, 18–20] was so far restricted to a small number of electrodes in sensor space, where connectivity changes cannot reliably be located in space [25]. Importantly, none of the previous studies used conditions with stimulation differences restricted to the phases of the applied fields [26, 27].

Here, we optimized our study design to minimize those restrictions. We analyzed pre-post stimulation connectivity change to avoid analysis of electric artifacts during stimulation and used detailed source reconstruction to localize effects with a centimeter resolution. Furthermore, our tACS montage yielded stimulation conditions with comparable power distributions of the electric fields, differing in their orientation. This design led to the absence of power differences between conditions at the stimulated frequency within the first 100 s after stimulation offset. The contrast IP-AP, which involved purely sinusoidal stimulation waveforms, did not induce any α-power differences and thus allowed for analysis of connectivity during the complete time course.

Nevertheless, some limitations also apply to our study. First, while we demonstrate robust changes in functional connectivity, the relation of these changes to behavior was not investigated here. The physiological relevance of the observed changes in coupling is thus left to future studies combining tACS with behavioral paradigms. Second, stimulation of peripheral nerves in the skin could contribute to the obtained effects. However, a large influence of tactile sensation seems unlikely as no significant increases in coupling were observed between the postcentral gyri and other regions (see Fig. 4A). Moreover, participants indicated no subjective difference in tactile sensation between

the stimulation conditions. Third, as we did not obtain electrophysiological recordings during stimulation, our results cannot be generalized to online effects of tACS which might be of a different nature compared to aftereffects.

Without knowledge on immediate effects of tACS, how can stimulation outlasting changes in functional connectivity be explained? While entrainment echoes might decay within few cycles, plasticity could be crucial for longer-lasting effects. Discrimination between entrainment and plasticity was discussed recently for aftereffects in α-power [16]. In contrast to connectivity, tACS aftereffects in power can robustly be studied at sensor level. Furthermore, optimization of high-definition tACS montages, as used here to minimize power differences between conditions, is not required to investigate aftereffects in power, although high-definition montages can in general help to spatially steer the stimulating fields. Stimulation outlasting power effects have been described in several recent studies [15–18] and may last up to 70 min or even longer [17]. Vossen et al. [16] argued that characteristics of α-power aftereffects did not fit with the hypothesis of entrainment echoes, as, for example, aftereffects did not depend on phase-continuity of the tACS stimulus. Rather, mechanisms such as spike-timing-dependent plasticity (STDP) could play a role, a suggestion which was supported by Wischnewski et al. [28] via application of the N-methyl-D-aspartate receptor (NMDAR) antagonist dextromethorphan. The NMDAR antagonist, limiting excitatory STPD, abolished tACS aftereffects on β-power that were seen in a placebo control [28]. Similarly, Zähle et al. [15] explained their α-power aftereffects by STDP.

In our study, it seems hard to delineate a mechanism explaining connectivity effects lasting for tens of seconds after tACS offset. Similar to Ahn et al. [18], we did not find a correlation between the match of stimulation frequency to the individual alpha frequency (IAF) and effect sizes (Fig. S2). As we neither detected an interaction of effects with baseline power or connectivity, we could not find any indication for ongoing activity strongly influencing the efficacy of tACS. Instead, we speculate that tACS might slightly increase or decrease the probability for spikes in a certain phase of the 10 Hz cycle, potentially without strong dependence on average ongoing activity. The relative phase between the stimulating fields would then bias the probability of simultaneous spikes, directly affecting STDP of synapses. This interpretation is consistent with our finding that JP stimulation, constantly varying the relative phase of

fields, showed intermediate or low connectivity changes. To directly test for the proposed mechanism, NMDAR antagonists as used by Wischnewski et al. [28] might be helpful in future studies.

In conclusion, we were able to modulate phase coupling, outlasting the stimulation period, between distant brain regions by transcranial electric stimulation. Our results lend strong support to the efficacy of tACS for frequency- and space-specific modulation of neural signals. Based on these findings, we suggest the use of bifocal high-definition tACS to manipulate large-scale cortico-cortical coupling in experimental settings. In addition, clinical studies with the aim of modulating pathological or maladaptive coupling could benefit from both specificity as well as non-invasive and easy applicability of tACS.

Materials and methods

Experimental setup

Participants

28 participants entered the study and received financial compensation. One participant was excluded as she closed her eyes during parts of the recording, and technical problems led to the exclusion of another participant. Since two other participants reported discomfort during test stimulation, no recordings were obtained from them. Hence, we were able to complete data collection from 24 participants in a counterbalanced sequence of three stimulation conditions. These final participants (12 male, 12 female) were on average 26 $\pm$ 4 years old. Vision was normal or corrected to normal and none of them had a history of neurological or psychiatric disorders. Participants gave written consent after introduction to the experiment. The ethics committee of the Medical Association Hamburg approved the study.

Electrophysiological recordings

Resting-state recordings included electroencephalogram (EEG) and electrocardiogram (ECG). Participants were seated comfortably in a dimly lit, electrically shielded room. 64 Ag/AgCl EEG electrodes mounted on an elastic cap (Easycap) were prepared with

an abrasive conducting gel (Abralyt 2000, Easycap) to keep impedances below 20 $k\Omega$. EEG signals were referenced to the nose tip and recorded using BrainAmp amplifiers (Brain Products GmbH). ECG was measured as the voltage between right infraclavicular fossa and left flank. All participants took part in three sessions on one day, differing only in the applied tACS condition. Each session was 23 min long: After a period of 5 min EEG resting-state recording, tACS was applied for 13 minutes, followed by 5 min post-tACS EEG recording (Fig. 1A). The beginning of each of these three periods was marked by a trigger signal synchronized to the EEG. Between the sessions, a break of at least 5 min was held.

Transcranial stimulation

Ten additional Ag/AgCl electrodes with 12 mm diameter were mounted between EEG electrodes for application of tACS. After preparation with Signa electrolyte gel (Parker Laboratories Inc), impedances of each outer electrode to the middle electrode ranged between 5 and 120 kΩ. For each montage, similar impedances were attempted, leading to a total impedance below 30 kΩ. Participants were made familiar with tACS by slowly increasing the amplitude of a test stimulus. During each session, α-tACS was applied with a peak-to-peak amplitude of 2 mA for 13 min, including a linear ramp-up within the first 10 s.

The three α-tACS conditions were in-phase (IP) stimulation, anti-phase (AP) stimulation, and jittered-phase (JP) stimulation. IP and AP stimulation consisted of two 10 Hz sinusoidal currents with zero and 180° phase offset, respectively, applied at the two hemispheres. For JP stimulation, the two stimulating currents independently changed their frequency between 9.5 and 10.5 Hz at constant amplitude (Fig. 1B). While power of all stimulating currents was focused around 10 Hz, coherence between the two signals was low in the JP condition (see Fig. S1). Hence, all three conditions involved α-stimulation, but differed in the relative phase between the stimulating currents. The sequence of tACS conditions was counterbalanced between participants.

tACS electrode montage and simulation of the applied electric field

Spatial application of tACS was chosen to target generators of alpha activity in parieto-occipital regions. We used two 4-in-1 montages similar to Helfrich et al. [14]. At

each hemisphere, one montage was applied and connected to an alternating current source (DC-Stimulator, NeuroConn), as shown in Fig. 1C. The use of separate stimulators for each montage ensured a focal field underneath the stimulation electrodes.

In order to quantify and compare field distributions for the different stimulation conditions, we simulated the electric field induced by tACS. The leadfield matrix L was computed with the boundary element method volume conduction model by Oostenveld et al. [29] and segmented using the AAL atlas, equally to source reconstruction of EEG signals. Distances between grid points were 1 cm in all three spatial dimensions. The electric field $\vec{E}$ at location $\vec{x}$ could then be calculated by linear superposition of the evoked fields of all injected currents α_i at stimulation electrodes $i = \{1, 2, ..., 10\}$ as

$$\vec{E}(\vec{x}) = \sum_i (\vec{L}_i(\vec{x})\alpha_i),$$

with the field strength $\sqrt{[\vec{E}(\vec{x})]^2}$. Contrasts in field strength were either computed as vectorial differences ($\sqrt{[\vec{E}_1(\vec{x}) - \vec{E}_2(\vec{x})]^2}$; capturing differences in field direction as well as absolute strength) or absolute differences ($\sqrt{[\vec{E}_1(\vec{x})]^2} - \sqrt{[\vec{E}_2(\vec{x})]^2}$; capturing differences only in absolute field strength).

To additionally simulate the stimulation field with a higher spatial precision, the leadfield matrix was also computed with the boundary element method on a volume conduction three-shell model which was reconstructed from the MNI template brain (see Fig. 1C). Here, spatial resolution was 5 mm in all three dimensions and allowed for detailed description of the stimulating fields (Appendix S1).

Data processing

EEG pre-processing

64-channel EEG data, recorded at 5 kHz, was pre-processed in MATLAB 2015b (The MathWorks Inc) and FieldTrip [30]. Time series were segmented into 300 s trials before ("pre recording") and after tACS ("post recording") prior to preprocessing to avoid leakage of the tACS artifact into the pre and post recordings. The first 500 ms of each trial were discarded as post recordings may contain remaining tACS artifacts related to capacitive effects within this period. Trials were re-referenced to common average,

two-pass high-pass (1 Hz) as well as low-pass (25 Hz) filtered with a second order Butterworth filter, and down-sampled to 100 Hz.

For the six trials of each participant, an independent component analysis (ICA) using the infomax ICA algorithm [31] was computed over all channels. Pearson's linear correlation coefficient [32] between the largest 30 independent components (ICs) and the ECG as well as the two electrooculography (EOG) traces were computed. ICs that showed a correlation coefficient larger than 0.2 to the ECG or one of the EOGs were removed in order to minimize cardiac and eye-blink artifacts. After back-projection to sensor space, channels were screened for high noise levels. If the standard deviation of a channel was higher than three times the median standard deviation of all channels, the channel was excluded. On average, 4.7 ± 1.5 components and 1.1 ± 1.5 channels per participant were removed. The whole pre-processing pipeline was thus automated and did not rely on subjective decisions of the investigator.

Source reconstruction

Exact low resolution brain electromagnetic tomography (eLORETA) [33, 34] - a discrete, three-dimensional distributed, linear, weighted minimum norm inverse solution - was used to estimate neural activity at source level. Before source projection, we band-pass filtered sensor data with a 2nd order butterworth filter between 7 and 13 Hz. eLORETA was then applied with 1% regularization, using the boundary element method volume conduction model by Oostenveld et al. [29]. Three-dimensional time series of dipoles were reconstructed at a linearly spaced grid of 1074 cortical and hippocampal voxels with distance 1 cm in all three spatial directions, and regions were assigned following the AAL atlas parcellation. To reduce the spatial dimension of the resulting time series, we used spatio-spectral decomposition (SSD) [35], maximizing activity between 9-11 Hz while suppressing activity in the flanking bands 8-9 Hz and 11-12 Hz. SSD was applied to time series of the three spatial dimensions at each voxel. Only the largest component was considered for further analysis [25].

Time-frequency analysis

EEG was analyzed in sensor as well as in source space. At each electrode in sensor space and each voxel in source space, power in frequency bands of 2 Hz width was

computed within 1 s segments without overlap and subsequently averaged over the 100 s window of interest. Likewise, for all electrode pairs in sensor space and all voxel pairs in source space, the absolute value of the imaginary part of coherency (|Im(coherency)|, also called imaginary coherence) [36] was computed in 1 s segments without overlap and averaged over the 100 s window of interest. In general, we looked at differences between pre- and post-recordings. To keep the distance between pre- and post-recordings constant for comparisons involving a time course, 100 s time segments were cut identically from pre- and post-recordings, leading to a constant time difference of 18 min between the onsets of each time window.

Connectivity metrics

Global connectivity shifts were assessed via the distribution of connectivity values between electrode or voxel pairs. We computed cumulative histograms as the fraction of pairs with a certain minimum connectivity change. In sensor space, all electrode pairs were taken into account. In source space, all interhemispheric pairs of voxels between the stimulated area in the right hemisphere ("right stimulated region") and the stimulated area in the left hemisphere ("left stimulated region") were included. Stimulated regions were defined as regions with average differences in vectorial field strength between IP and AP stimulation above 0.08 V/m. Averaging over many pairs further minimizes spurious local functional connectivity [36], in particular in sensor space. In source space, we additionally investigated the topology of α-coupling changes. Differences in imaginary coherence were computed for all pairs of voxels and averaged within regions of interest.

Permutation statistics

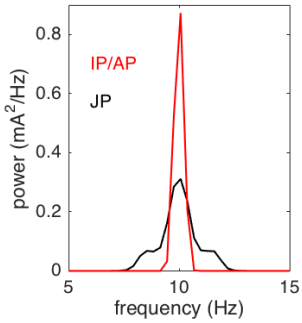
To test for significance of condition differences between IP, AP, and JP stimulation, we performed permutation statistics. 100 s epochs of both pre- and post-recording were cut into trials of 1 s each. For each trial, conditions were shuffled to observe control time series, while pre-post pairings remained intact. As conditions were randomly mixed in these control time series, confidence intervals for connectivity differences between the conditions could be estimated. We computed 100 control time series (200 control time series in sensor space) and control connectivity measures. The 4950 (19900 for sensor

space) difference values between the 100 control connectivity measures served as control distributions for condition differences: The 2.5th and 97.5th percentile of these distributions defined the uncorrected lower and upper 95% confidence limits, respectively.

When three stimulation condition differences were compared, we applied Holm-correction to the confidence limits: If the contrast with the lowest p-value was significant after Bonferroni-correction, subsequent comparisons were adjusted for a lower number of comparisons. Figures show the confidence interval after the last comparison yielding significant results. For analyses on multiple time windows, the first time window (0-100 s after tACS offset) was relevant for Holm-correction. To additionally account for multiple comparisons of frequency bands, the confidence intervals were further shifted by a constant. This constant was chosen such that in the sum of all multiple comparisons, only 5% (or a lower fraction after Holm-correction for multiple stimulation conditions) of control time series yielded significant results. Hence, these corrected confidence intervals, shown in light gray, indicate a 5% probability of false significant results in the complete analysis.

Supporting information

S1 Fig. Description of tACS stimuli. While IP as well as AP stimulation consisted of pure sinusoidal signals, frequency "sweeps" were included in the stimulating current for the JP condition (see Fig. 1B). The sweeps included several linear shifts in frequency from 9.5 to 10.5 Hz and back. Consequently, power of the stimulating current peaked at 10 Hz, but also extended to the neighboring frequencies. In contrast, sinusoidal stimulation used for IP and AP stimulation showed a sharp peak in power at 10 Hz. Coherence between JP stimulation currents was below 0.02 for all frequencies.



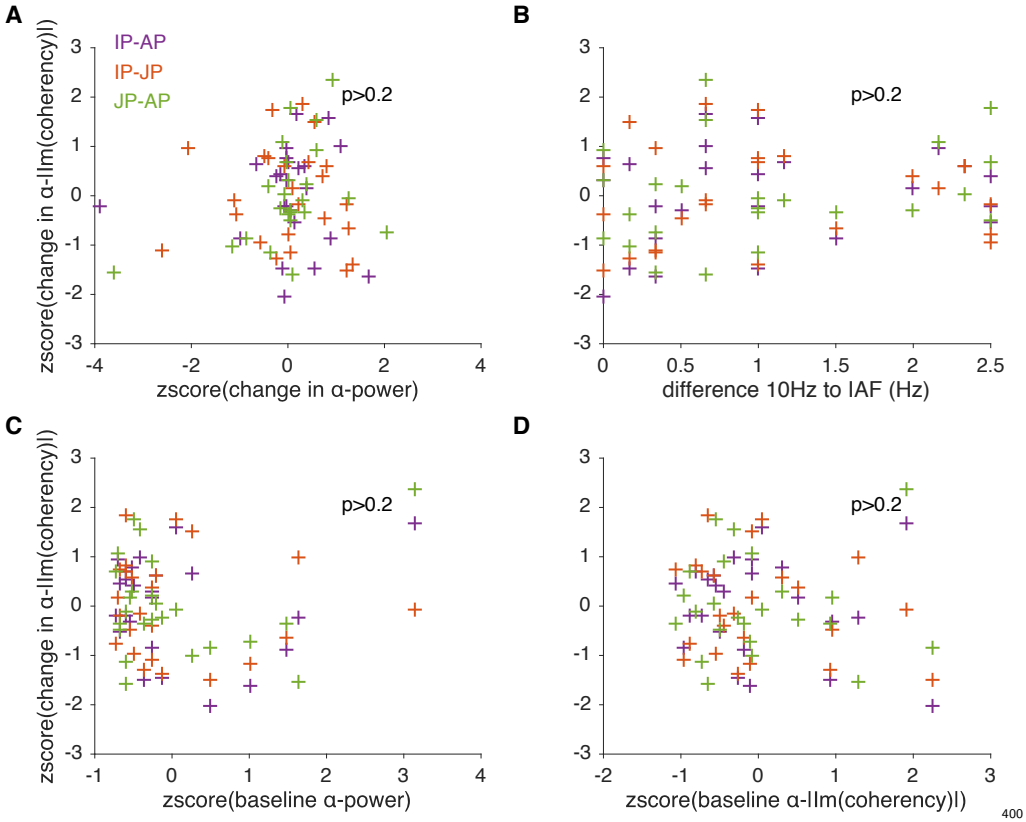
S1 Appendix Properties of stimulating fields at 5 mm resolution. We simulated the stimulating cortical electric field dependent on the stimulation condition. Fig. 1C depicts the absolute field strength distribution for IP and AP stimulation at time point 25 ms, where current flow reaches the maximum value of 1 mA in each montage. Within one stimulation cycle, all polarities reversed for IP and AP stimulation. For JP stimulation, the relative phase of the stimulating currents constantly varied, and thus the stimulating electric field changed between the ones computed for IP and AP stimulation. As stimulating fields were maximally different between IP and AP stimulation, we restricted ourselves to reporting differences between these two.

Ideally, stimulation conditions should differ only in the direction or temporal phase of the applied field and not in their absolute strength or focality [26]. Our stimulation configuration was chosen to optimally meet these criteria. At each voxel, the absolute field strength difference between IP and AP stimulation was below 70 mV/m; the average difference was -1.3 mV/m. Compared to peak field strength values of 356.1 mV/m (IP) and 369.0 mV/m (AP; 3.6 % more than IP), condition differences were thus low. In contrast, vectorial differences peaked with 637.7 mV/m and had average values of 39.6 mV/m.

S2 Fig. Connectivity effects correlate neither with power effects nor with individual physiology. Pearson correlations coefficients over participants between the main effect - the change in grand average α-imaginary coherence between tACS conditions within the time interval 0-100 s - and other metrics, namely: A: change in grand average α-power between tACS conditions, within the same time interval; B: difference between stimulation frequency (10 Hz) and individual alpha frequency (IAF); C: baseline α-power of all pre-recordings; D: baseline α-imaginary coherence of all pre-recordings. None of the correlations reached significance (uncorrected $p > 0.2$). Imaginary coherence and power values were z-scored. α denotes the stimulated frequency band (9-11 Hz).

The absence of a correlation between connectivity and power effects over participants supports the robustness of our connectivity effects. Furthermore, we did not find a correlation of connectivity effects with the match of the stimulated frequency to IAF, indicating no potential benefits of stimulation at the IAF compared to 10 Hz

stimulation. Finally, both baseline α-power and -connectivity of participants did not significantly affect connectivity differences between conditions. Hence, we did not find evidence for individual physiology to interact with tACS efficacy.



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