Neural mechanisms of training in Brain-Computer Interface: A Biophysical modeling approach

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Abstract

Brain-computer interface (BCI) is a system that translates neural activity into commands, allowing direct communication between the brain and external devices. Despite its clinical application, BCI systems fail to robustly capture subjects' intent due to a limited understanding of the neural mechanisms underlying BCI control. To address this issue, we introduce a biophysical modeling approach that leverages a linear neural mass model to investigate the associated neural mechanisms of motor imagery-based BCI experiments. We tailor this model to simulate both motor imagery task and resting state. We apply this approach to a cohort of 19 healthy subjects trained along four sessions where magnetoencephralography (MEG) and electroencephalography (EEG) signals were simultaneously recorded. The neural synaptic gain and time scale of the modeled excitatory and inhibitory neural mass populations capture changes in neural activity across conditions and sessions. Those changes appear in important areas of the sensorimotor cortex, relevant for motor imagery tasks. We observed these effects in both EEG and MEG modalities. These findings provide insights into the underlying neural mechanisms in a motor imagery task in BCI, paving the way to tailored BCI training protocols.

Keywords: Brain-computer interface, Neural mass model, EEG, MEG, Motor imagery, Training, Biophysical modeling

1 Introduction

Developing a control system based on brain signals is a formidable challenge. Brain-Computer Interfaces (BCIs) necessitate the acquisition of intricate neural data and the conversion of these signals into dependable commands, all within the constraints of an incomplete understanding of brain function (1; 2). Despite these obstacles, BCIs hold significant promise, particularly in clinical applications, where they enable direct brain-to-device communication to compensate for lost motor or sensory functions (3).

Nonetheless, non-invasive methods such as electroencephalography (EEG) and magnetoencephalography (MEG) often provide limited control for many users (4). While training can enhance BCI performance, a critical question persists: why do some users exhibit substantial improvement while others face ongoing difficulties? This issue is particularly evident in motor imagery-based BCIs (MI-BCI), where users operate a device by imagining movements without physically executing them. Numerous studies have attempted to address the various ambiguities in this field, including enhancing brain signal decoders (5), identifying neurophysiological and psychological factors affecting user performance (6; 7; 8), and developing improved

training protocols (9). A critical element believed to be essential for successful BCI control is neuroplasticity, which underlies the brain's capacity to adapt and learn. This has spurred increasing interest in the neurophysiological mechanisms involved in neurofeedback and BCI learning (10). Furthermore, the phenomenon of BCI inefficiency has been examined from multiple perspectives. Learning to use a BCI differs fundamentally from interacting with intuitive, everyday tools; it requires mastering a new, abstract form of control. It is reasonable to infer that specific brain mechanisms are involved when a user learns to perform effectively in a BCI task. However, these underlying mechanisms remain largely unexplored in the current literature, underscoring the need for biophysical models capable of elucidating the brain processes involved in BCI learning and adaptation.

A class of biophysical models that have been particularly useful for these purposes is called the "Neural Mass Models" (NMMs). NMMs model the population-level activity of interactions between excitatory and inhibitory neurons in different brain regions that potentially underlies the macroscopic signals we observe with E/MEG. As a result, they provide biophysical insights into how macroscopic signals may arise in various neurological conditions/diseases. Although these models have been used widely, they have rarely been used in the case of BCI, potentially because of the additional challenge of modeling the tasks appropriately. Indeed, most modeling studies focus on the resting state (RS) with limited attempts to model BCI-related tasks (11; 12). In this study, we developed a novel approach to model MI-BCI leveraging a linearized NMM that accurately captures the spectral features of the neural oscillatory activity estimated from source-localized MEG in healthy individuals in the RS (13; 14). This NMM has also been successful in identifying cortical-level neural impairments in patients with Alzheimer's disease (15) as well as neural modulations in healthy individuals transitioning from wakefulness to light sleep (16).

Using this approach, we hypothesized that key spectral differences between MI and RS result from region-specific, population-level interactions between excitatory and inhibitory neural subpopulations. Furthermore, we posited that such region-specific changes could capture the neural mechanisms occurring during learning. To validate these hypotheses, we utilized source-reconstructed EEG and MEG signals obtained during a four-session MI-BCI training from healthy subjects.

2 Materials & Methods

2.1 Brain-Computer Interface data collection and analysis

In this study, we worked with a previously published data set (17), where a cohort of 19 healthy right-handed subjects (mean age:27.4 ± 4.0 years; 12 men) participated in a BCI study. All participants were BCI-naive, free from medical or psychological disorders, and provided their written informed consent in accordance with the Declaration of Helsinki. The study was approved by the ethical committee CPP-IDF-VI of Paris.

Subjects underwent a two-week motor imagery-based BCI training program consisting of four sessions. At the beginning of each session, the subjects underwent a 3-minute RS with eyes opened. The BCI experiments consisted of a standard one-dimensional, two-target box task (18), where subjects controlled the vertical position of a cursor moving at a constant velocity from left to right to reach a target displayed on the right part of the screen. To reach the up target, the subjects were instructed to perform sustained right-hand motor imagery (MI) by modulating neural activity in the alpha (8–12 Hz) and beta (13–30 Hz) frequency bands. To hit the down target, the subjects had to remain at rest. Each of the six runs consisted of 32 trials, distributed equally and randomly among the targets.

Neural activity was recorded using simultaneous EEG and MEG modalities with a 74-channel EEG system (Ag/AgCl passive sensors, Easycap, Germany) following the 10-10 montage. MEG data were

acquired using 102 magnetometers and 204 gradiometers (Elekta Neuromag TRIUX) in a magnetically shielded environment. The signals were sampled at 1 kHz (0.01–300 Hz bandwidth) and then downsampled to 250 Hz. MEG signals were preprocessed with temporal signal space separation (tSSS) using MaxFilter (19). Eye and cardiac artifacts were removed from EEG/MEG signals via Independent Component Analysis (ICA) with the Infomax approach in FieldTrip (20; 21). EEG electrodes T9 and T10 were excluded from further analysis.Source reconstruction was performed using the Boundary Element Method (BEM) (22; 23) with three-layer MRI-derived surfaces (scalp, inner skull, outer skull; 1922 vertices each). The weighted Minimum Norm Estimate (wMNE) (24; 25; 26) was used to estimate cortical sources via Brainstorm (27), with an identity noise covariance matrix. The dipole orientations were constrained to the cortex. Individual source estimates were projected onto the MNI-ICBM152 template (28) using Shepard's interpolation.

Power spectra were calculated for each subject, session, and condition in the source space using the Welch method (1 s window, 50% overlap). For the rest condition, the estimation was performed directly from the 3-minute recording. For the MI condition, the power spectra were computed from the feedback period (t = 3–6 s) and averaged across trials. To localize the anatomical structures associated with the obtained clusters, we used the Desikan killiany (DK) atlas (29) without restricting the analysis to the motor or sensorimotor regions. The dataset was structured as (19 × 68 × 126) per session and condition, where 19 represents subjects, 68 corresponds to DK atlas-defined brain regions, and 126 represents frequency bins used in spectral analysis.

2.2 Model description

We used a linearized NMM (13; 14; 15) to extract excitatory and inhibitory neuronal subpopulation parameters. The model at the mesoscopic level (regional model), for every region k (k varies from 1 to N and N is the total number of regions) based on the DK parcellation, is modeled as the sum of excitatory signals $x_{\mathrm{e}}(t)$ and inhibitory signals $x_{\mathrm{i}}(t)$. Both excitatory and inhibitory signal dynamics consist of a decay of the individual signals with a fixed neural gain, incoming signals from populations that couple the excitatory and inhibitory signals, and input Gaussian white noise for resting state and colored noise for mentally active state. The model equations for the excitatory and inhibitory signals for every region of the DK atlas that describes the rate of change of signal or firing rate in excitatory and inhibitory neural sub-populations are:

$$\frac{\mathrm{d}x_{\mathrm{e}}(t)}{\mathrm{d}t} = -f_{\mathrm{e}}(t) \star \frac{1}{\tau_{\mathrm{e}}}\big(g_{\mathrm{ee}}x_{\mathrm{e}}(t) - g_{\mathrm{ei}}f_{\mathrm{i}}(t) \star x_{\mathrm{i}}(t)\big) + p(t)\ , \text{ and} \tag{1}$$

$$\frac{\mathrm{d}x_{\mathrm{i}}(t)}{\mathrm{d}t} = -f_{\mathrm{i}}(t) \star \frac{1}{\tau_{\mathrm{i}}}\big(g_{\mathrm{ii}}x_{\mathrm{i}}(t) + g_{\mathrm{ei}}f_{\mathrm{e}}(t) \star x_{\mathrm{e}}(t)\big) + p(t)\ . \tag{2}$$

The symbols used in the equations are as follows: $*$ denotes convolution; parameters g_{ee}, g_{ii}, and g_{ei} represent neural gains for excitatory, inhibitory and for coupling between excitatory and inhibitory populations, respectively; τ_{e} and τ_{i} are the time constants of excitatory and inhibitory populations, respectively; $p(t)$ is input noise; $f_{\mathrm{e}}(t)$ and $f_{\mathrm{i}}(t)$ are average neural impulse response functions. Parameters g_{ei}, g_{ii}, τ_{e}, and τ_{i} were estimated for each region of interest (ROI), while g_{ee} was fixed at 1 for parameter identifiability. The excitatory and inhibitory time constants characterize the duration of neural responses (modeled by a gamma-shaped function) in each neuronal subpopulation. They also determine the rate at which a local signal dissipates in the absence of other inputs. A lower time constant indicates a faster rate of change in signals, whereas a higher time constant corresponds to a slower rate. The excitatory and inhibitory gain parameters represent the strength of self-excitatory and self-inhibitory feedback gain within their respective neuronal subpopulations, and the coupling neural gain represents the strength of interaction between the two populations, as in how one subpopulation influences the activity of other.

We can get a closed-form solution in the Fourier domain since the two equations above are linear, as shown below. The Fourier transform $F()$ is taken at an angular frequency ω, which is equal to $2\pi f$, where f is the frequency in Hz. The Fourier transform of equations 1 and 2 will be the following, where $F(x_{\mathrm{e}}(t)) = X_{\mathrm{e}}(\omega)$ and $F(x_{\mathrm{i}}(t)) = X_{\mathrm{i}}(\omega)$, and j is the imaginary unit.

$$\mathrm{j}\omega X_{\mathrm{e}}(\omega) = -\frac{F_{\mathrm{e}}(\omega)}{\tau_{\mathrm{e}}}(g_{\mathrm{ee}}X_{\mathrm{e}}(\omega) - g_{\mathrm{ei}}F_{\mathrm{i}}(\omega)X_{\mathrm{i}}(\omega)) + P(\omega)\ , \text{ and} \tag{3}$$

$$\mathrm{j}\omega X_{\mathrm{i}}(\omega) = -\frac{F_{\mathrm{i}}(\omega)}{\tau_{\mathrm{i}}}(g_{\mathrm{ii}}X_{\mathrm{i}}(\omega) + g_{\mathrm{ei}}F_{\mathrm{e}}(\omega)X_{\mathrm{e}}(\omega)) + P(\omega)\ . \tag{4}$$

Here, $P(\omega)$ is the Fourier transform of the input noise $p(t)$, which we assume to be identically distributed for both excitatory and inhibitory local populations for each region.

On solving the above equations 3 and 4, $X_{\mathrm{e}}(\omega)$ and $X_{\mathrm{i}}(\omega)$ are:

$$X_{\mathrm{e}}(\omega) = \frac{1 + \frac{F_1}{\tau_{\mathrm{e}}F_2}}{F_3 + \frac{F_1^2}{\tau_{\mathrm{e}}\tau_{\mathrm{i}}F_2}}P(\omega)\ , \text{ and}$$

$$X_{\mathrm{i}}(\omega) = \frac{1 - \frac{F_1}{\tau_{\mathrm{i}}F_3}}{F_2 + \frac{F_1^2}{\tau_{\mathrm{e}}\tau_{\mathrm{i}}F_3}}P(\omega)\ ,$$

where

$$F_1 = g_{\mathrm{ei}}F_{\mathrm{e}}(\omega)F_{\mathrm{i}}(\omega), \quad F_2 = \mathrm{j}\omega + \frac{g_{\mathrm{ii}}}{\tau_{\mathrm{i}}}F_{\mathrm{i}}(\omega), \quad \text{and} \quad F_3 = \mathrm{j}\omega + \frac{g_{\mathrm{ee}}}{\tau_{\mathrm{e}}}F_{\mathrm{e}}(\omega).$$

Then, the transfer functions $H_{\mathrm{e}}(\omega)$ and $H_{\mathrm{i}}(\omega)$ can be separated, and we get:

$$X_{\mathrm{e}}(\omega) = H_{\mathrm{e}}(\omega)P(\omega), \quad \text{and} \quad X_{\mathrm{i}}(\omega) = H_{\mathrm{i}}(\omega)P(\omega).$$

Thus, for the total neural population at each region:

$$X_{\mathrm{local}}(\omega) = (H_{\mathrm{e}}(\omega) + H_{\mathrm{i}}(\omega))P(\omega). \tag{5}$$

To simulate RS condition, we set $P(\omega)$ to be Fourier transform of white noise. Therefore, it was set to be a constant. To simulate MI, we assumed $P(\omega)$ to be Fourier transform of colored noise, proportional to $1/(f^k)$, where k was allowed to vary within the range [0, 1].

2.3 Model parameter estimation

Model parameters were estimated for each region, each subject, and across all sessions, for frequency range of [3-30] Hz. For every subject, each region's spectrum was modeled using equations 3 and 4, and finally the total spectrum obtained from equation 5 was compared to the empirical spectrum. The goodness of fit of the model was evaluated by calculating Pearson's correlation coefficient (r) between the modeled power spectra and the empirical source-localized (E/MEG) spectra. This goodness-of-fit value was used to optimize the model parameters.

Parameter optimization was performed using the basin-hopping global optimization algorithm in Python (30). The initial guess for all parameters and the parameter bounds used for estimating the modeled spectra are provided in Table 1. To ensure model stability, we defined five different bounds for the neural gain

(g_{ei}) (31). Initially, a larger bound was applied to g_{ei}, and if the optimal model parameter crossed the stability boundary, the optimization was repeated with a smaller bound. This process was iterated up to five times to ensure that the final optimal parameters corresponded to stable model solutions. The model parameters and their bounds are specified in Table 1. In addition to these parameters, the color of the input noise was also optimized for in the case of MI, within the range of [0,1] as mentioned in the previous section.

The algorithm's hyperparameters included the number of iterations (`niter`), temperature (`T`), and step size (`stepsize`), which were set to 5000, 0.1, and 4, respectively. If any parameter hit the specified bounds, the optimization was repeated with a step size of 6 for that specific ROI. Finally, the set of parameters yielding the highest Pearson's correlation coefficient was selected. The cost function for optimization was defined as the negative Pearson's correlation coefficient between the source-localized (E/MEG) spectra in the dB scale and the model power spectral density in the dB scale.

Table 1: Model parameters with initial guesses, and bounds for estimating the spectra

Parameters	**Symbol**	**Initial guess**	**Lower/upper bound for optimization**
Excitatory time constant (s)	τ_{e}	17	[5.0, 30.0] (Same in all 5 bounds)
Inhibitory time constant (s)	τ_{i}	17	[5.0, 30.0] (Same in all 5 bounds)
Alternating population gain	g_{ei}	0.4	[0.001, 0.7], [0.001, 0.5], [0.001, 0.4], [0.001, 0.3], [0.001, 0.25]
Inhibitory neural gain	g_{ii}	1.0	[0.001, 1.9] (Same in all 5 bounds)

As this is the first pilot study, we started with comparing the most contrasting conditions. Therefore, we focused on the resting state recorded before the BCI experiment was started, and the MI performed during the BCI training. Specifically, we did not include the baseline condition in which subjects were instructed to remain at rest during the BCI training – the baseline condition may not correspond to strictly being in the resting state as the subjects can still continue to think about the experiment/their performance. A workflow of this modeling approach is shown in Figure 1.

2.4 Statistical analyses

To examine the potential condition effect as in, significant difference between motor imagery (MI) and resting state (RS) in each session separately, we performed Wilcoxon signed rank test (32) on all four biophysical parameters across 19 subjects. We chose this non-parametric Wilcoxon test as the model derived biophysical parameters were not normally distributed. This analysis was conducted for each region of interest (ROI) to compare MI and RS conditions for all four biophysical parameters. For each session, we obtained 68 p-values (one for each brain region based on the DK atlas) per parameter, thus four sets of 68 p-values per session (one set for each parameter). Thereafter, we applied the false discovery rate (FDR) correction (33) to each of the four sets to account for multiple comparisons. We set a significance threshold at $P_{\mathrm{FDR}} < 0.05$ to identify statistically significant regions for these parameters individually. The brain regions that met this significance criterion were identified for each parameter separately in each session. We used the same procedure both for EEG and MEG.

Next, we analyzed how the session affected MI and RS in both E/MEG. We used the Friedman test (34) on the statistically significant left motor regions for the corresponding parameter from the session 4 condition effect analysis. Then, we corrected the two p-values (as two regions in left motor area in MEG) using the Bonferroni correction method.

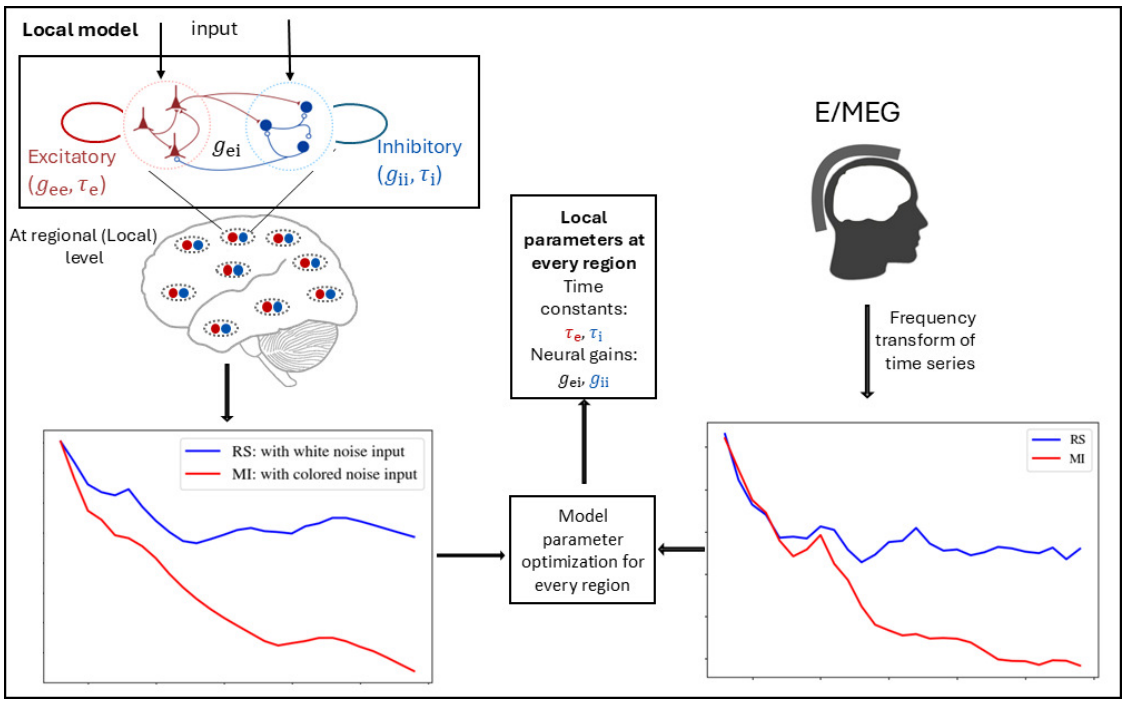

Figure 1: **Workflow for the modeling approach:** On the left, we show that the excitatory and inhibitory neuronal populations are modeled for every brain region using a small set of biophysically meaningful parameters. Both these populations also receive an input that is simulated as white noise for RS (resting state) and colored noise for motor imagery (MI). Upon simulating the model, we directly obtain the power spectral density (PSD) shown at the bottom left. We optimize for the model parameters (shown in the middle) by matching the modeled PSD to the E/MEG PSD for every region, shown in the right most column. After obtaining the optimized parameters, we performed statistical analyses to test for condition and session effect in both E/MEG separately.

3 Results

3.1 Simulating modeled input as colored noise better captures MI power spectra in M/EEG

The model for the excitatory/inhibitory neural population signals incorporates decay of these signals over time, interactions among them, and an external input. We assume that this external input refers to background nonspecific noise in the case of resting state (RS) condition. Therefore, we model them as white noise as was done in previous studies (15; 14). Specifically, we set $P(\omega)$, the input noise parameter in Equation (5), to be a constant, representing Fourier transform of white noise, for every region of interest (ROI) in all sessions and subjects. In contrast, we assume that the external input to the excitatory/inhibitory neural populations should alter to account for performing an active task such as motor imagery (MI). Therefore, we modified $P(\omega)$ to simulate the MI condition differently from the RS condition. Specifically, we simulated $P(\omega)$ to be Fourier transform of colored noise, proportional to $1/(f^k)$, where k was allowed to vary within the range [0, 1] as colored noise for all ROIs, sessions, and subjects, and ω is $2\pi f$.

First, we compared the spectral exponents from the data set itself to demonstrate that they are different in RS versus MI, therefore justifying our approach to model colored noise in the case of MI. To directly calculate the spectral exponents of all ROIs in both RS and MI conditions, we employed FOOOF, a Python package (35) designed to fit to the power spectra from E/MEG recordings. Firstly, the results showed that FOOOF provided a better fit to the median spectra (calculated by taking the median across all performers for each ROI) than to individual spectra in almost all the sessions/modalities that we considered, based on the R2 metric measuring the goodness-of-fit between the actual spectra and the FOOOF-modeled spectra. For MEG, the mean R2 values for RS condition are 0.983 (median spectra) vs. 0.950 (individual spectra), and for MI condition are 0.990 (median spectra) vs. 0.980 (individual spectra). For EEG, the mean R2 values for RS condition are 0.859 (median spectra) vs. 0.865 (individual spectra), and for MI condition are 0.973 (median spectra) vs. and 0.9118 (individual spectra). This is likely because the median spectra was smoother and ,therefore, easier to fit to using FOOOF, rather than some of the individual spectra. Note that we used the median instead of the mean when comparing the spectra, as there were few potential outliers that could skew the results (see Figure S1 in the supplementary).

Figure 2 shows the histogram of the spectral exponents in both RS and MI conditions, for median and individual spectra. Figure 2: A, B demonstrate that the spectral exponents in the RS condition are smaller than those in the MI condition, when evaluating spectral exponents from the median spectra. However, the histograms for the spectral exponents obtained from the individual spectra shown in Figure 2C, D do not indicate any noticeable difference between the RS and MI conditions. Given that FOOOF fit better to the median spectra, we focused on the result obtained from the median spectra which showed a clear difference in the spectral exponents. Note that a larger spectral exponent can be achieved in our model by incorporating colored noise, while a smaller spectral exponent (capturing the RS condition) can be easily modeled with white noise, as was done in previous studies (15; 14). Therefore, we introduced this approach of assuming colored noise as the internal input in the model to simulate the MI condition.

3.2 Model captures M/EEG power spectra accurately in both RS and MI conditions

The comparison between the modeled and actual power spectral density (PSD) across different brain regions and sessions reveals a strong correspondence in the decibel (dB) scale over the frequency range of 3 to 30 Hz. The spectral shape of the modeled PSD closely aligns with that of the actual PSD across all examined regions and sessions, demonstrating the effectiveness of the modeling approach in capturing neural oscillatory patterns measured with M/EEG. To demonstrate this qualitatively, example spectra for a subset of regions

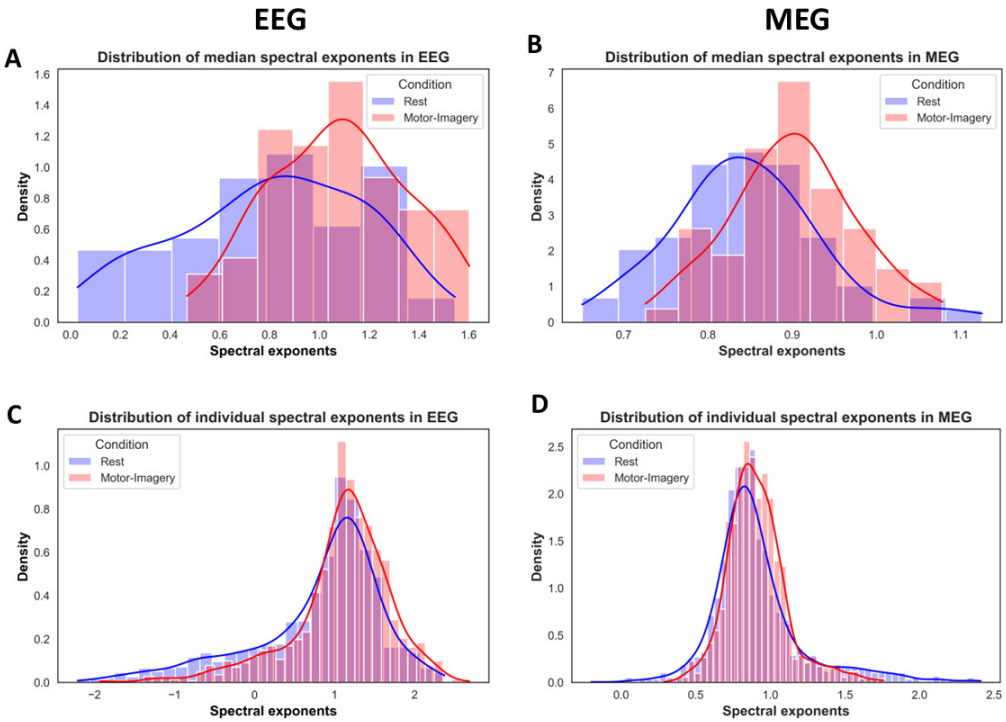

Figure 2: **Distribution of spectral exponents in motor-imagery (MI) and Rest (RS) condition:** A, B exhibit histograms of median spectral exponents in both the condition and show that the mean of total exponents in RS is less than in MI in EEG and MEG, respectively; whereas C and D exhibit histogram distribution of individual spectral exponents in MI and RS, with exponents in both conditions nearly completely overlapping in both EEG and MEG, respectively.

are shown in Figure 3A-D. Here, the regions for RS-modeled EEG, RS-modeled MEG, MI-modeled EEG, and MI-modeled MEG are shown, respectively, demonstrating how the modeled spectral shape aligns with the actual E/MEG spectral shape. A quantitative assessment reinforces these findings, since the Pearson correlation coefficient between the modeled and actual PSD exceeds 0.90 in most cases (see Figure 3 - E, F) in both RS and MI for E/MEG in Session 4. This was also the case for other sessions (see Supplementary Figure S4 for histograms of other sessions). This high correlation indicates that the model successfully replicates key spectral features captured via M/EEG in different sessions and regions.

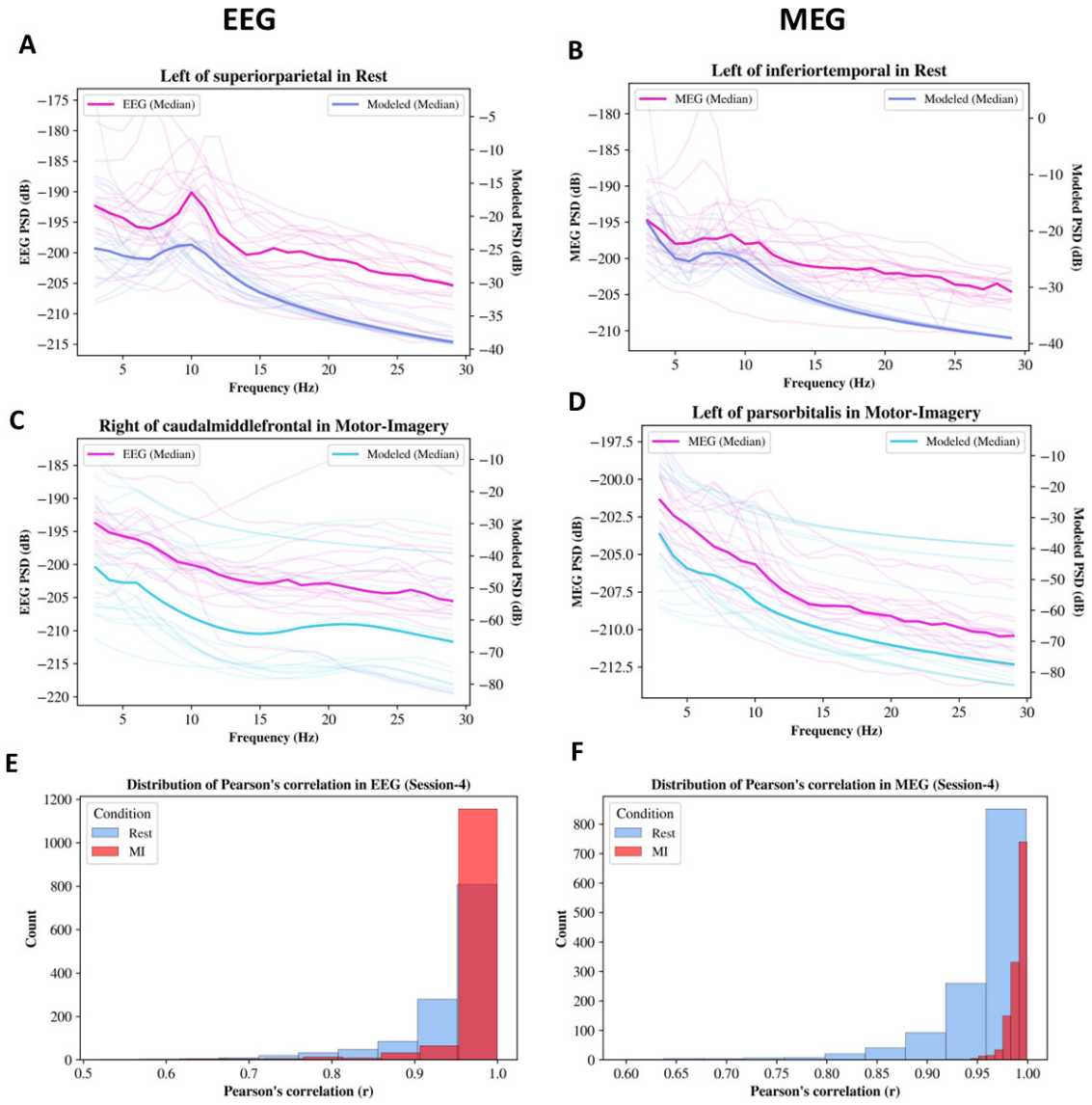

Figure 3: **Matching shape of modeled and actual PSD, with pearson correlation coffecient in both RS and MI**: A, B: Modeled PSD with EEG and MEG PSD shape in dB scale along frequency for Rest in case of Left superior parietal and left inferior temporal region, respectively. It shows how the spectra is changing in 19 subjects, and bold colored spectras (medians of the 19 modeled and actual E/MEG spectra) showing matching of the shape between model and actual median; C, D: Similar to (A, B) these two are for Right Caudal middle frontal and Left pars orbitalis in MI condition, respectively; E, F: Session 4 distribution of Pearson correlation coefficient between modeled and actual spectra for MI (red) and Rest (blue) condition in EEG and MEG, respectively;

3.3 Assessing brain activation changes during BCI learning through region-specific neuronal population dynamics

The biophysical model parameters reflect the underlying excitatory-inhibitory population-level dynamics that shape how and when a region becomes active in response to internal or external inputs in different conditions. Thus, to investigate the possible condition effect for MI versus RS, we conducted Wilcoxon test for each of these parameters separately in each session (see Figure S5, $P_{\text{FDR}} < 0.05$). We first examined the spatial distribution of significant brain regions in MEG - associated with three parameters: g_{ei}, g_{ii}, and τ_{e}, with a focus on consistent activation patterns grouped by lobe (see Figure 5). For an exhaustive presentation of the regions that showed significant effects, please refer to the supplementary table (Table ST1) with the list of significant region of interests (ROIs), session by session for each of these parameters.

As subjects were instructed to perform motor imagery tasks, our primary interest was to investigate the potential involvement of motor areas. For g_{ii}, which reflects self-inhibitory gain within inhibitory population, the *right postcentral gyrus* was found to be significant by session 4. The parameter τ_{e}, which characterizes the duration of excitatory neural response, yielded the most extensive and consistent activation patterns across sessions with the recruitment of the *paracentral*, the *precentral*, and the *postcentral* gyri during the training. In particular, the left *paracentral* gyrus where we observed a consistent condition effect, showed also a significant training effect across time (Friedmann test, $P_{\text{FDR}} < 0.05$). More specifically, such an effect was driven by the MI condition (see Figure 4). This finding is particularly important, as the left paracentral region plays a crucial role in voluntary motor activity of right hand grasping, making it functionally significant in motor control. This change suggests that neural activity in the left paracentral region is dynamically influenced between sessions during motor imagery tasks, potentially reflecting learning effect.

Additionally with non-motor parts, we observed the involvement of non-motor areas throughout the training, beyond those targeted by the BCI system. With g_{ei} parameter that plays key role in stabilizing excitatory signals, effects were exclusively observed in session 1. Such brain regions were located in the frontal lobe (left *pars opercularis*), in the parietal lobe (right *inferior* and *superior parietal cortices*), and in the occipital lobe (left *lateral occipital*). For g_{ii}, the bilateral *transverse temporal* cortex were involved thorough the training. The activations of the other lobes occured only in session 3 before vanishing in session 4. They involved a reduced number of regions of the frontal (left *rostral middle frontal*), the parietal (left *inferior parietal* cortex and the right *posterior cingulate*), and occipital lobes (left *lateral occipital* cortex). The parameter τ_{e} yielded again the most extensive and consistent activation patterns across the lobes and training. In the frontal lobe, the left *pars opercularis* was consistently involved until the last session. The other lobes showed consistent activations in all four sessions: in the parietal lobe (the *supramatrarginal* gyrus bilaterally), the *inferior parietal* cortex, the *superior parietal* gyrus, the *precuneus*, and the *posterior cingulate*); the temporal lobe (*transverse temporal*); and in the occipital lobes (the *cuneus* and the *pericalcarine* cortex). Such findings indicate session-by-session differences in brain activity between MI and RS, with significant activation in key sensory, motor, and cognitive processing areas between different sessions. These results highlight the dynamic nature of brain activations in MI tasks as compared to rest.

We replicated this analysis with EEG data (refer to Figure S5 and Table ST2). No training effect has been observed (Friedman test, $p > 0.05$). Most of the significant conditions effects were observed from session 4. For g_{ei}, such effects involved in particular motor areas (left *precentral gyrus*), parietal lobe (left *inferior parietal* lobule) and occipital areas (bilateral *cuneus*). Similarly, for g_{ii}, the bilateral cuneus showed significant effects. Lastly, for τ_e, the right supramarginal gyrus was significantly affected. These findings suggest session-specific variations in brain activity between MI and RS, mostly starting from session 4 that showed the most pronounced differences.

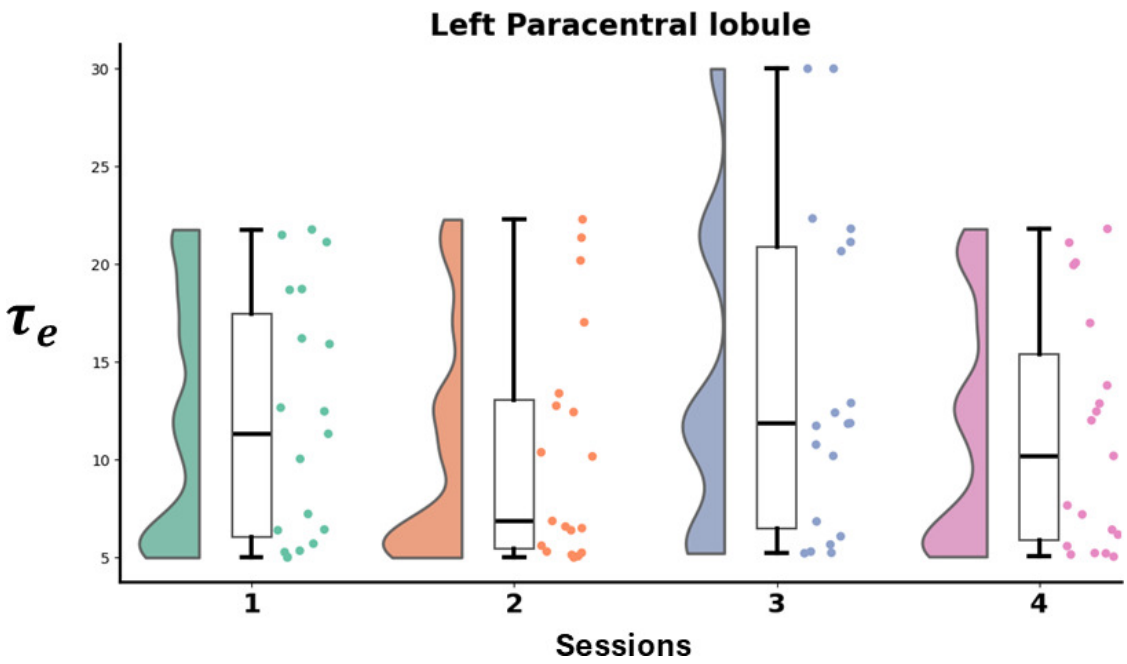

Figure 4: **Variation in τ_e model parameter driven by MI condition:** Distribution of 19 τ_e values in each session show inter-session variation in the left paracentral region during MI in MEG modeled spectra. Substantial variation has been observed due to MI (P = 0.0452; Bonferroni adjusted P = 0.0904), which is a key indicator of session effect driven by motor-imagery on group performance in BCI learning.

4 Discussion

This work represents the first biophysical modeling study in the field of BCI research to use a linear NMM for simulating motor task-related MI data, by modulating the color of the noise in the input signal to the excitatory/inhibitory neuronal populations. Using this novel modeling approach of simulating MI versus rest, we successfully captured E/MEG spectra of individual brain regions during both conditions. Our findings demonstrate that this approach not only effectively models the MI and RS condition but also identifies key brain regions activated in MI across different BCI training sessions. In doing so, we provide new insights into the neural mechanisms underlying BCI training, paving the way for potential improved interpretability and optimization of BCI systems.

4.1 Comparisons with previous models

One important point of discussion is whether using a linear instead of a non-linear model is justified in this case. The key advantage of using a linear model is that we can compute the PSD directly and then compare it with the real data. This reduces the computational load substantially and allows for a wider search of optimal model parameters. Biophysically, it is unclear which model captures the empirical features correctly. However, prior evidence suggests that a linear model may be sufficient in many cases. Most prominently, linear models outperformed non-linear models in predicting resting-state fMRI time series, owing to the linearizing effects of macroscopic neurodynamics and neuroimaging due to spatial and temporal averaging, observation noise, and high dimensionality (36).

It is also unclear whether simulating MI condition by altering the input noise is the most appropriate

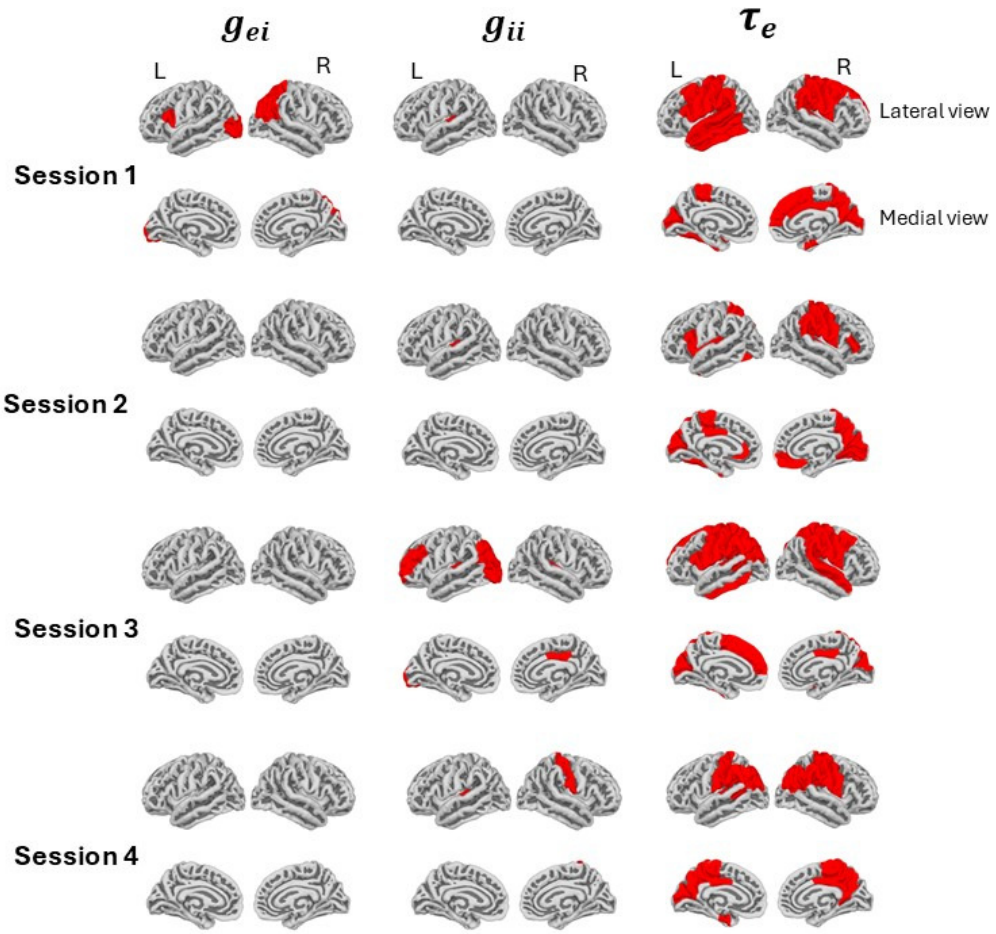

Figure 5: **MEG-based condition effect analysis in BCI learning:** The highlighted red colored regions show statistically significant ($P_{\text{FDR}} < 0.05$) condition effect for MI Vs RS for the three MEG fitted model parameters (g_{ei}, g_{ii}, τ_e) respectively in every session. This show how functional reorganization happening over time in the 19 subjects' group and various cortical regions are getting activated for MI BCI task.

approach. However, to our knowledge, in all the previous studies on non-linear NMMs for motor tasks, the white noise input (that is used to simulate RS) has been modified. For instance, in (12), where the motor task involved grasping a pressure sensor with the right hand to measure grip strength, the mean and standard deviation of the Jansen-Rit model's white noise parameter were adjusted to fit to the alpha-band EEG power spectral density data. Similarly, in (11), where data were recorded during right-hand finger movements and the execution of a working memory task, the Wendling et al. (37) NMM was modified by incorporating white noise with a defined mean and variance. Our study applied a linear NMM in which the noise parameter remains white noise with a constant mean and variance in RS. However, for the MI-BCI task, we assumed the input to be colored noise. This is because unlike non-linear neural mass models, simply changing the mean and variance of the white noise will not change the shape of the PSD – it will only modify the amplitude. Therefore, we needed to modify the color of the noise in our case. Additionally, the spectral exponents evaluated from the M/EEG recordings supported this hypothesis, as shown in Figure 2.

4.2 Relevance of model parameters in motor task

The linearized NMM we used here allows us to relate the model's parameters to the neurophysiological functions of the cerebral cortex. This is especially true for the motor cortex, which was recorded by both EEG and MEG during the right-hand grasping motor imagery task. By fine-tuning these model parameter values within a biophysically realistic range, we were able to closely match the PSD shapes of the E/MEG and modeled spectra. This allowed us to further investigate the link between the physiological mechanisms underpinning BCI training and the biophysically meaningful model parameters.

Our findings suggest that with each session, the model parameters are able to capture the functional reorganization occurring in the cortical regions. Furthermore, with progress in training with a specific focus on MI, it appears that subjects modulate brain rhythmic activity in relevant motor regions in the alpha and beta frequency bands, which is captured by the model parameters.

By matching the shape of the E/MEG PSD with the modeled PSD from the NMM, we explore the relationship between model parameters and the associated neural mechanisms of BCI training. The findings indicate that the MEG-fitted modeled parameters evolved from session 1 through session 4. The excitatory time constant (τ_{e}) evolved from randomly capturing cerebral cortex regions in session 1 to precisely capturing specific motor and associated regions in session 4 such as: Motor lobe (right precentral, bilateral paracentral) responsible for motor imagery and motor hand tasks (38); Parietal lobe (bilateral inferior parietal, bilateral postcentral, bilateral precuneus, bilateral supramarginal) involved in sustained attention, detection of salient events, memory retrieval and memory processing (39; 40; 41; 42; 43; 44); Occipital lobe(left cuneus, left pericalcarine) associated with visual mental imagery and detection of patterns (45; 46; 47); Limbic regions (right isthmus cingulate, bilateral posterior cingulate) responsible for spatial memory and navigation (48; 49). Similarly, with inhibitory neural gain parameter (g_{ii}), the postcentral region in session 4 engaged in MI BCI task and showed significant condition effect, whereas excitatory neural gain parameter (g_{ii}) showed no significant effect in later sessions except session 1.

The EEG-fitted modeled parameters also evolved from session 1 through session 4. Initially, no specific brain regions were captured during the early training sessions. However, by the end of the training, in session 4, the parameters successfully identified highly relevant regions for the MI-BCI task. This suggests that although subjects were initially naïve, training enabled them to engage the necessary brain regions for BCI. The excitatory-inhibitory gain parameter (g_{ei}), which showed no significant regional effects in session 1, finely adjusted its values over the course of training. This gradual tuning suggests that the interaction between excitatory and inhibitory neural populations was actively regulated over time, helping certain brain regions become more responsive. Such changes likely reflect a shift toward increased net excitation at the population level, allowing these regions to engage more effectively during task performance. By session 4, it successfully captured critical regions, including: Occipital lobe (bilateral cuneus, left lateral occipital)-involved in visual pattern detection, tracking visual motion, visual memory recognition, and visual mental imagery (45; 46; 47; 50); Parietal lobe (left inferior parietal)-associated with somatosensory processing and spatial awareness (39); Frontal-motor area (left precentral cortex) - responsible for motor control and execution (38). Similarly, with the inhibitory neural gain parameter (g_{ii}), we could observe some important regions (bilateral cuneus, left isthmus cingulate, left lingual) (48) for BCI control that show significant condition effect in session 4. Also, the excitatory time constant (τ_{e}) in session 4 showed significant condition effect for supramarginal gyrus, which involved in somatosensory processing task (43; 44).

These observations align with a previous study (12), where NMM parameters successfully captured the underlying mechanisms of motor tasks in EEG. That study employed a single-channel NMM at the sensor level, specifically the C3 channel, which corresponds to the primary motor area. In contrast, our study utilizes source-reconstructed recordings from all the regions of cerebral cortex, offering very comprehensive and

significantly higher accuracy than sensor-space data. In that study, the effects of average excitatory and inhibitory synaptic gain were analyzed to examine the power spectral density of the model's output signal in the alpha band, while keeping other parameters at typical values. By doing so, they established connections between Jansen's single-channel NMM parameters and the physiological mechanisms of the motor cortex, particularly by analyzing changes in grip force during a motor task. Their findings demonstrated that as grip force increased, the recruitment of motor neurons in the motor cortex also rose. The model effectively captured these changes in EEG signals by adjusting gain parameters, suggesting that increasing grip force activates more motor neurons in the motor cortex and elevates their firing rate.

4.3 Investigating learning effect with linearized NMM

One of the primary challenges in BCI study is addressing inter- and intra-subject variability (4; 51; 52; 53). Understanding the large-scale neural mechanisms underlying plasticity is essential for elucidating the processes of human learning (54; 55; 56; 57). The voluntary modulation of neural activity to operate a BCI is increasingly regarded as an acquired skill. Numerous studies have demonstrated that performance in BCI tasks typically improves with practice (58; 59; 10). Users often report a transition from explicit, effortful cognitive strategies—such as motor imagery—to more automatic, goal-directed control that closely resembles natural motor execution. This behavioral progression is indicative of procedural motor learning mechanisms being engaged in the brain. To investigate the neural dynamics underlying BCI skill acquisition, researchers have employed neuroimaging methods in both non-human primates (58; 60; 61; 62; 63) and humans (64; 65; 66). These studies suggest that, although BCI control typically relies on modulating activity within motor-related brain areas, a broader and dynamically changing network of cortical regions contributes to the early stages of BCI learning. Other groups have investigated the use of power spectra within theta, alpha and gamma bands as potential predictors of BCI scores (6; 67; 68), with some difficulties to replicate such observations (69). Such elements explain why the specific mechanisms supporting BCI learning remain poorly understood. Understanding how neural patterns differ across individuals (inter-subject) and within the same individual across sessions (intra-subject) is crucial for developing robust BCI training programs.

This study aims to enhance specificity at both the individual and regional levels, providing more precise estimates of learning effects in BCI applications and offering deeper insights into the underlying mechanisms. While investigating the effect of MI compared to RS in each session, we analyzed regions of cerebral cortex based on DK atlas independently for each parameter across individuals (inter-subject) in E/MEG. Here we can observe with increase in session the NMM parameters are able to specifically capture various important regions ($P_{\text{FDR}} < 0.05$) that are activated and are responsible for the BCI task. Excitatory-inhibitory coupling neural gain parameter (g_{ei}), reflecting the functional interaction strength between the excitatory and inhibitory populations, showed significant changes in a limited, yet functionally critical areas in session 1. Specifically, significant regions that got activated as a result include the right inferior parietal cortex which is involved in sustaining attention over time (70); left lateral occipital cortex for visual memory recognition, visual object completion (50); left pars opercularis, and right superior parietal cortex essential for the manipulation of information in working memory and supports executive processes required to rearrange and mentally transform information (71). This initial pattern of activation may serve as a baseline or scaffold that enables subsequent learning as we did not observe any significant regions in other sessions. As participants become more experienced with the BCI task, the neural processes underlying sensory integration and motor planning likely become more efficient, possibly reducing the need for further modulation of the g_{ei} parameter. In other words, although g_{ei} is strongly engaged during the initial performance of the task, its influence appears to stabilize with practice, providing a consistent framework upon which other neural adaptations are reflected in parameters such as g_{ii} and τ_{e}.

With inhibitory neural gain parameter (g_{ii}) we did not observe motor related or associated area getting activated at the beginning of the training. However, by session 3, increased functional engagement of task-relevant regions may reflect a shift in local population/circuit dynamics, where refined self-inhibitory feedback within the inhibitory neural population reduces the overall inhibitory output, thus decreases its net suppressive influence on excitatory neurons. This disinhibition potentially gives rise to net excitation with greater excitatory activity and allows specific regions to become more functionally responsive during task performance, enhancing the effectiveness of BCI training. Thus involvement of these areas suggests that as learning progresses, inhibitory gain processes become more distributed across networks including regions involved in sustained attention (*inferior parietal* and *posterior cingulate*) (70; 49); and in visual processing (*lateral occipital*) (50). Such an expansion exhibits the brain's adaptation to more efficiently suppress task-irrelevant inputs (72) and optimize sensorimotor integration during motor imagery. Finally by session 4, the pattern of activation associated with g_{ii} showed very distinct refinement, with significant activity observed in the right postcentral area (73; 41). The involvement of this region underscores the integration of somatosensory information and the fine-tuning of motor learning, while the continued significance of the left transverse temporal area from session 1 till the end reinforces its role in memory processing. This refined pattern in the final session indicates that with sustained practice, inhibitory gain mechanisms evolve to support more specific functions—namely, enhancing voluntary motor control. Additionally, we observed a decrease in the total number of activated regions from session 1 (29 regions) to session 4 (19 regions). This reduction suggests that, with training, subjects were able to selectively engage the most relevant regions responsible for the MI task, thereby improving their control over the BCI. Such observations are in line with previous findings, where event-related desynchronization (ERD) observed as a significant power decrease ($p < 0.025$) in alpha and beta frequency bands. This was observed in cortical regions associated with key cognitive and sensorimotor functions relevant to BCI control, namely (17): sensorimotor integration and motor imagery (*superior parietal lobule*), decision-making and memory consolitation (*middle anterior cingulate gyrus*), behavioral adaptation (*posterior ventral cingulate gyrus*, and *frontomarginal gyrus*), visual tracking (*lingual gyrus*), mental rotation and working memory (*orbital part of the inferior frontal gyrus*), (linked to cognitive control).

With the excitatory synaptic time constant (τ_{e}), which represents the duration of neuronal responses in excitatory neuronal sub-populations, is a critical parameter for determining how long excitatory signals persist in the brain. In our study, τ_e exhibited notable changes in sessions that are indicative of learning-induced neural plasticity during motor imagery. In session 1, τ_{e} related activity is widely distributed in many regions suggesting that at the beginning of BCI training, the duration of excitatory response are widely distributed, reflecting a generalized and random recruitment of neural circuits involved in sensory processing, higher-order cognition, and motor coordination. By session 2, as learning begins, the activation of τ_{e} becomes more focused. Significant effects emerge in regions involved in visual processing and mental imagery (*cuneus*, and *lingual gyrus*) (46; 47; 74; 75), error detection (left *insula*) (76). This indicates an early refinement of the timing of excitatory response, as the brain begins to adapt to the temporal dynamics of the specific motor imagery condition (77). Then, τ_{e} related activity expanded further to include additional areas such as the left inferior parietal and temporal regions, suggesting that, with continued practice, the duration of the excitatory neural population response is being fine-tuned to support more efficient sensorimotor integration and cognitive processing. Finally, in session 4, τ_{e} exhibited the most extensive and robust activation pattern, with significant involvement of regions crucial for motor imagery and motor hand tasks (*precentral, paracentral and postcentral* gyri) (78; 79); for memory processing (*precuneus*, *posterior cingulate*, *entorhinal* cortex) (45; 49; 80); and for sustained attention (*inferior parietal* and *supramarginal* regions) (70; 43). We went one step further and investigated the potential training effect in the left paracentral region for MEG-fitted

modeled τ_{e} values during MI. There is a substantial variation in τ_e over time for this region, which also shows a significant effect of condition. This progressive expansion and increased specialization of τ_{e} between sessions strongly suggest that with repeated practice, the neural circuits underlying the excitatory synaptic time constants become finer tuned. This tuning likely improves the temporal precision of sensorimotor integration, reflecting learning-induced neural plasticity such as functional reorganization as subjects improve their performance during motor imagery tasks. This adaptation is a hallmark of learning, suggesting that repeated motor imagery practice leads to more efficient and targeted excitatory responses in the brain's sensorimotor networks.

4.4 Replicability

In order to further validate our observations, another dataset was brought for analysis. In this dataset, 15 subjects (RH, 8 F) performed a motor imagery task to grasp an object using a robotic arm. They use a combination of gaze to indicate the position to reach and either performed MI of the right hand or resting state to grasp or not the object presented. They actively controlled the robot for 3 runs with 10 trials per condition. The protocols and results of the findings can be found in (81). Among the strategies of control used, was only kept the strategy where the robot moved prior to the motor imagery task as it showed in the past the highest level of desynchronization across subjects. In order to reduce variability of the findings due to the processing pipeline, we performed source reconstruction using wMNE and used parcellation with PCA over dipoles to get the Desikan-Killiany atlas in the different conditions of rest and MI. This set of subjects did not have their MRI scan meaning that the inverse method was done using fsaverage (a MRI template on the average of 40 subjects (82)). This means regions involved are to be interpreted cautiously as some of them could be overlapping over points specific to the different subjects. The MI-specific NMM revealed a good level of power spectra decrease in the post central region (83; 84) although it did not reach significant differences (S7), it contributes to confirm the MI of the right hand taking place in the different subjects. Furthermore, different regions that were brought up in the previous analysis (85) appeared to be significant using the model. Among them, we can pin out the caudal middle frontal region (86; 87) and the paracentral region (88) both used for motor planning to facilitate the execution of goal directed actions and attention allocation which brings up new insights on the robotic involvement in the MI process, the frontal pole associated with high-level cognitive functions like decision-making (with the limitation that it could be artifacted by gaze activity. On another note, the superior parietal lobe was again found to be activated significantly. This region is found to be a central hub of multisensory integration and spatial perception (89), it is also involved in sense of agency and visuomotor coordination. Again, those findings come to reinforce previous results to argue of the impact of the robotic arm over subjects in their integration of the device to their mental workspace. Those results demonstrate by the data the validity of the model as they continue to capture phenomena that were already observed and highlight them even more. Other regions that were previously not listed showed to appear significant and they present neural relevance such as the pericalcarine and cuneus regions involved in visual processing (90); the medial orbitofrontal lobe associated to reward processing (91; 92)(therefore the feedback).

An important point of discussion is the reproducibility of the model findings, especially given that the experimental conditions and the tasks performed in the two BCI trainings are completely different. Despite these differences, the model consistently identified regions of the brain that overlap as statistically significant ($P_{\mathrm{FDR}} < 0.05$) across datasets, suggesting robustness in its ability to capture key neurobiological correlates. Specifically, in the case of the parameter g_{ii} (which reflects the inhibitory gain in inhibitory neural sub-population), two regions emerged as replicable in both data sets: the parahippocampal cortex (responsible for episodic memory formation and memory retrieval) was identified in both data sets as a region

of significance. This recurrent involvement suggests a stable role of this region in the inhibitory-inhibitory circuit dynamics modeled, possibly reflecting task-general mnemonic processing. And the Cuneus region (involved in visual detection of patterns, visual mental imagery and attention modulation) also consistently appeared as significant. This overlap may reflect a shared visual or attentional demand across tasks, even if they differ in cognitive context.

It is worth highlighting that this consistency between datasets, particularly in the absence of identical experimental conditions, demonstrates potential generalizability of the model and reinforces confidence in its biological validity. However, not all parameters showed such reproducibility. For instance, g_{ei} and τ_e did not yield overlapping regions between datasets, possibly indicating that parameter-specific effects may be more sensitive to task or dataset characteristics.

4.5 Limitations

One limitation of the model is that we only tried to match the modeled spectra to the shape of the power spectra, not the magnitude. This is because the most prominent change from resting state to MI is in the shape of the spectra, and therefore it is reasonable to focus only on capturing the shape by using the Pearson correlation coefficient as the cost function in the optimization. This also eases the computational burden since we do not need to estimate the optimal magnitude as an additional parameter.

An interesting observation from our analysis of EEG and MEG data is the behavior of the fitted model parameter τ_{e}. In EEG, τ_{e} does not show any condition effect or capture any region except one in session 4, making it difficult to pinpoint specific areas of activity for this parameter. However, in MEG, τ_{e} reliably highlights regions within the cerebral cortex during all four sessions, suggesting that MEG provides a more stable and regionally specific representation of neural dynamics compared to EEG. Similar to the findings in our previous study (17), a greater number of regions of interest exhibited significant condition effects in MEG compared to EEG. This discrepancy can be attributed to the distinct biophysical properties of the two modalities. EEG signals are attenuated by the varying conductivities of intervening tissues, such as the scalp, skull, and cerebrospinal fluid (93), and are sensitive to both tangential and radial components of cortical dipoles. In contrast, MEG offers superior spatial resolution and is primarily sensitive to the tangential components of neural sources (94). Consequently, EEG and MEG do not necessarily capture the same aspects of brain activity. Furthermore, participants were instructed to lean their heads back within the MEG helmet during data acquisition. This positioning brought the MEG sensors closer to the parietal and occipital regions, potentially enhancing the sensitivity to posterior brain activity relative to signals arising from frontal areas.

5 Conclusions

Despite the fact that BCI is no longer an emerging field, mechanisms involved in training to master the control are too often examined solely through the lens of performance, without adequately addressing the brain's intrinsic dynamics. One key aspect in the adoption of BCIs–particularly in motor imagery paradigms–is understanding how excitatory-inhibitory (E/I) coupling mechanisms unfold. Models previously described E/I coupling have been mostly applied to resting-state data where the input to the E/I neuronal populations are assumed to be background non-specific noise and therefore approximated by Gaussian noise. BCI tasks rely on active engagement–such as motor imagery. Our work addresses this gap by extending existing modeling approaches to account for task-driven activity. Our results show that, once fine-tuned, the model is capable of replicating task dynamics across sessions in a way that closely mirrors real data. Its ability to track session-to-session variations provides compelling evidence that E/I coupling may play a role as an underlying

brain mechanism during BCI learning. Furthermore, the model successfully captures differences between tasks across modalities (EEG, MEG) and experimental conditions–whether the BCI feedback was provided via a screen (visual feedback) or through a robotic arm (embodied feedback). These additional findings strengthen the robustness of the observed effects and lend credibility to the overall approach.

Looking ahead, these insights open the door to integrating E/I coupling into the BCI framework, as it emerges as a relevant feature in shaping and informing the training process. More broadly, embedding these mechanisms into BCI design could contribute to more refined and neurobiologically grounded models, ultimately enriching both the theoretical and practical aspects of BCI development.

Availability of materials

All the codes related to the paper can be found on this Github page: https://github.com/ApurbaApd/sgm_bci_modeling

Funding

The authors acknowledge support from Transatlantic Research Partnership from the FACE Foundation and Alzheimer's Association grant AARFD-22-923931.

Supplementary material

Please see the supplementary PDF for the model equations and supplementary figures.

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Supplementary Document

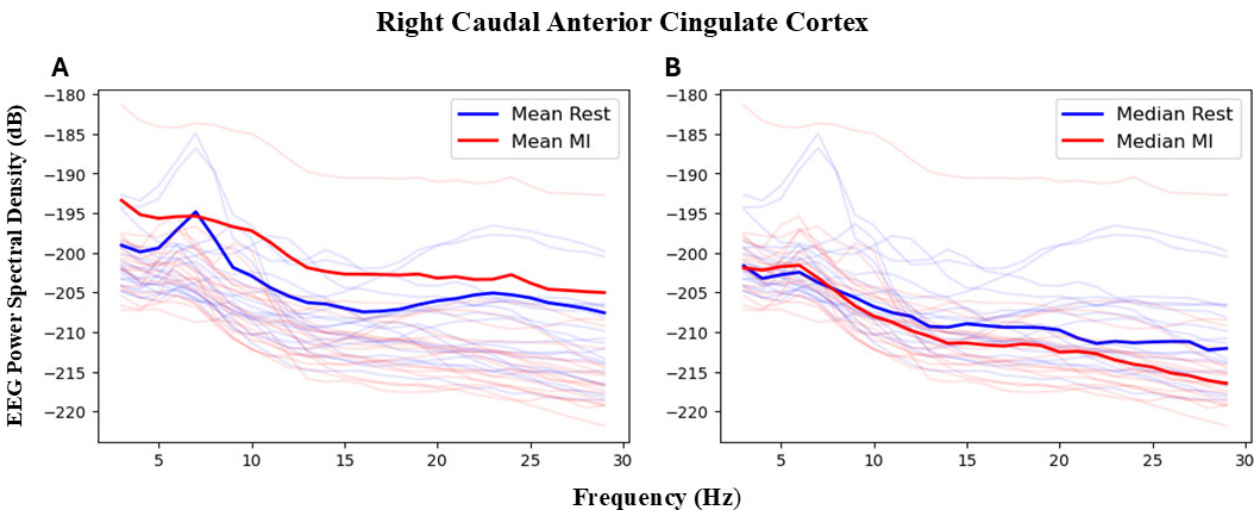

Figure S1: **Mean Vs. Median spectra:** From the figure A and B, it's very clear that the median spectra in B gives better representation of 19 individual spectra than mean spectra in A in both motor-imagery (MI) and Rest condition, as few individual spectra skewing the overall representation in case of mean.

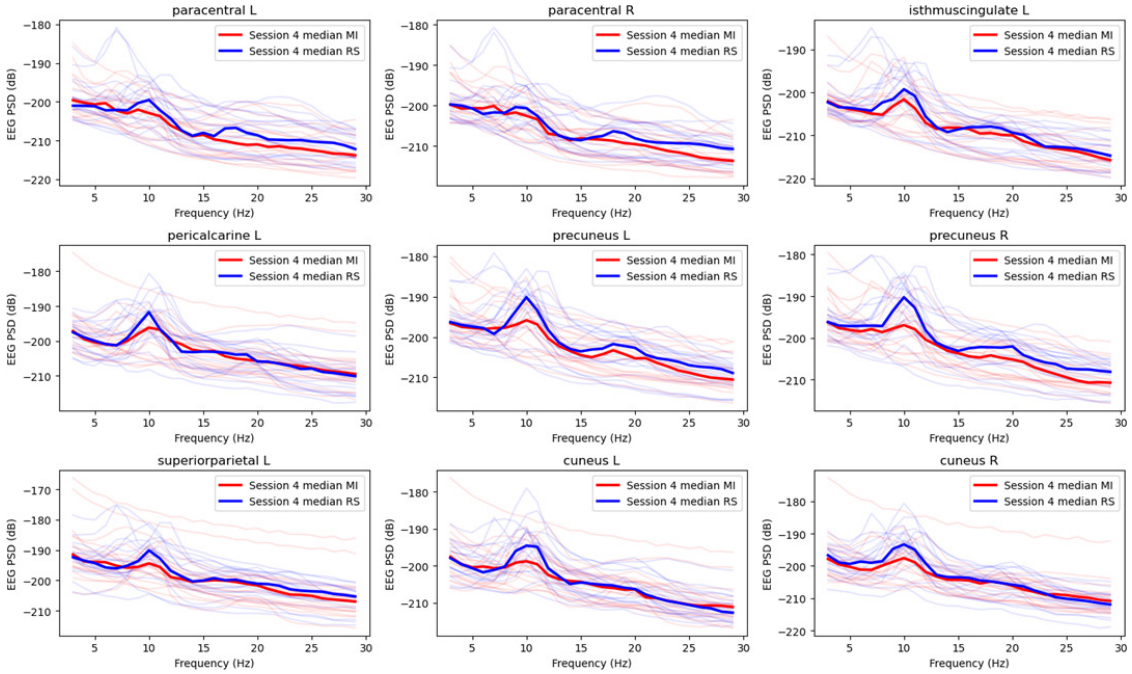

Figure S2: Representation of changes in power spectral density (PSD) in session 4 EEG along the frequency range (3-30 Hz) in a few regions.

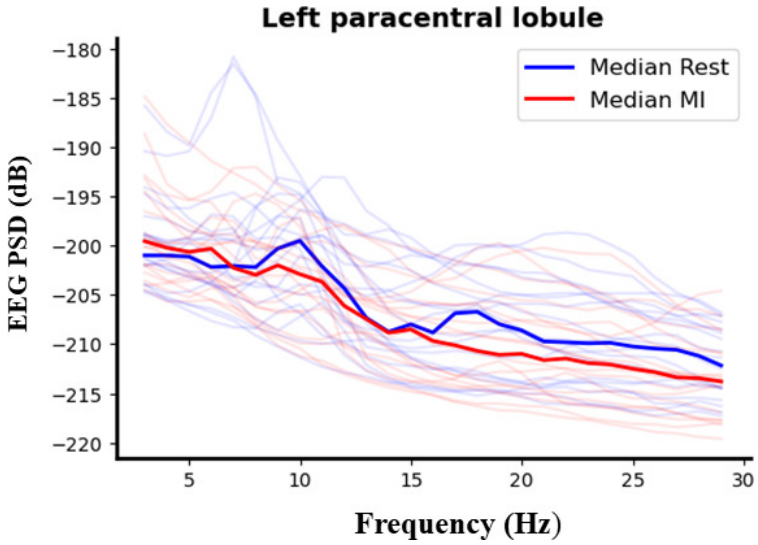

Figure S3: **Changes in EEG-Power Spectral Density (PSD) in Left paracentral lobule:** It shows that the changes in PSD with frequency in case of Rest and MI condition, with median of the distribution of 19 spectra in both the condition. The alpha and beta peak can be observed in both the condition.

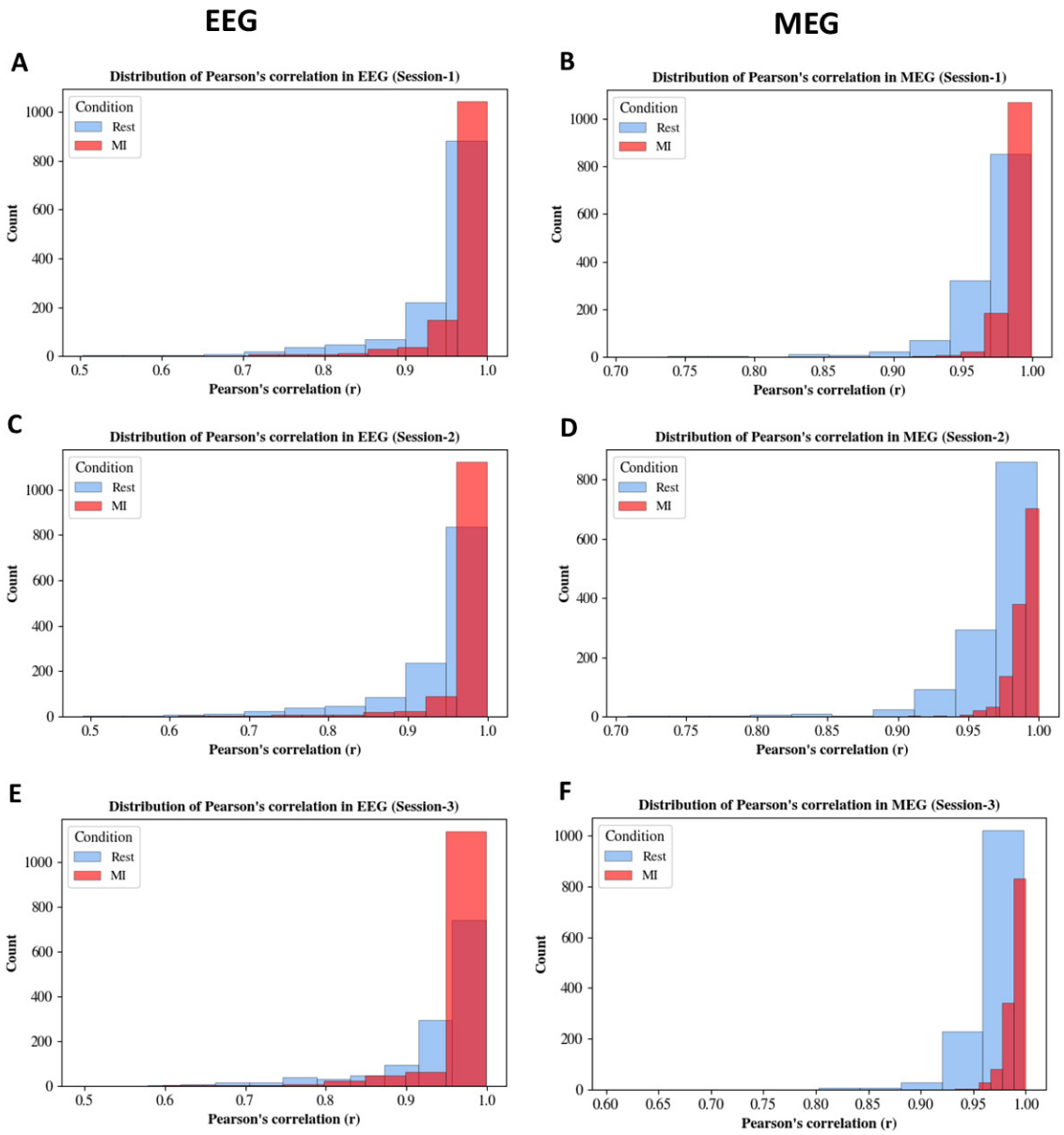

Figure S4: **Pearson correlation coffecient between modeled and actual PSD in both Rest (RS) and MI**: A, B: Shows the distribution of the coeffecient to be more than 0.90 in most cases in both RS and MI condition in session 1 EEG and MEG, respectively; C,D: Similar to the above fig, here it shows in session 2 EEG, MEG respectively; E, F: it Shows the distribution in session 3 EEG, MEG respectively

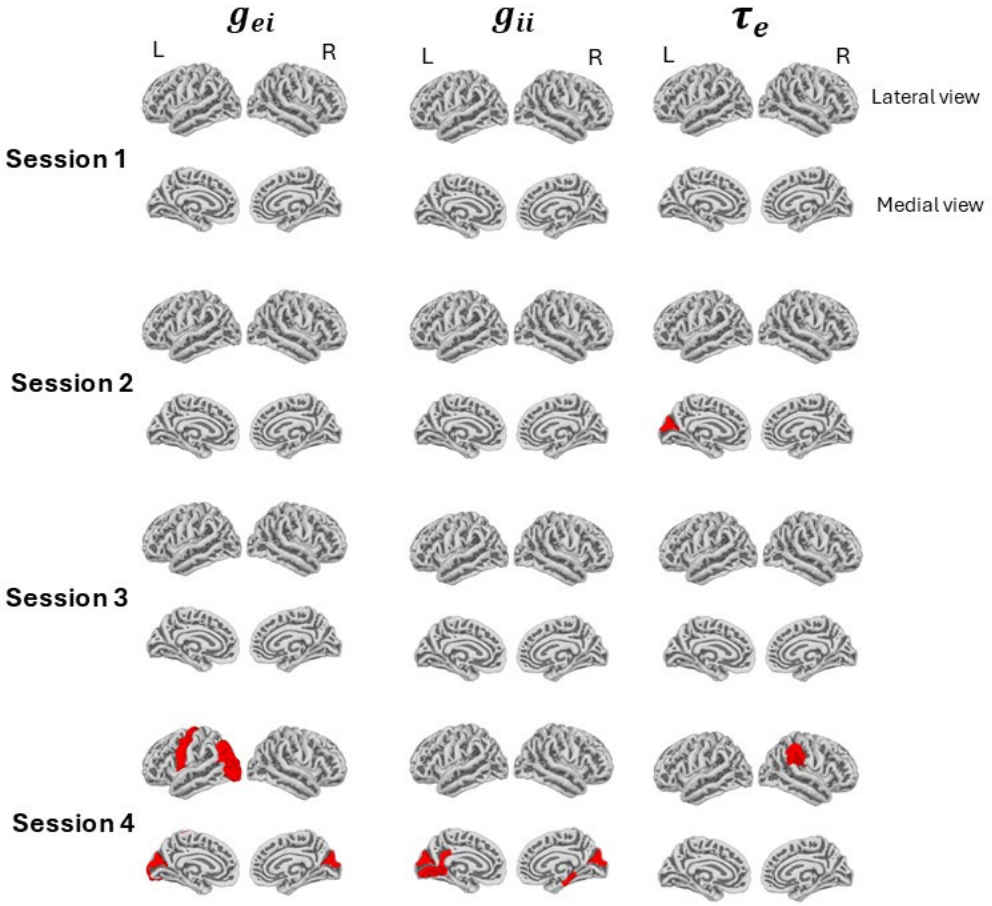

Figure S5: **EEG-based condition effect analysis in BCI learning:** The highlighted red colored regions show statistically significant ($P_{\text{FDR}} < 0.05$) condition effect for MI Vs RS for the three EEG fitted model parameters (g_{ei}, g_{ii}, τ_e) respectively in every session. This show how functional reorganization happening over time in the 19 subjects' group and various cortical regions are getting activated for MI BCI task.

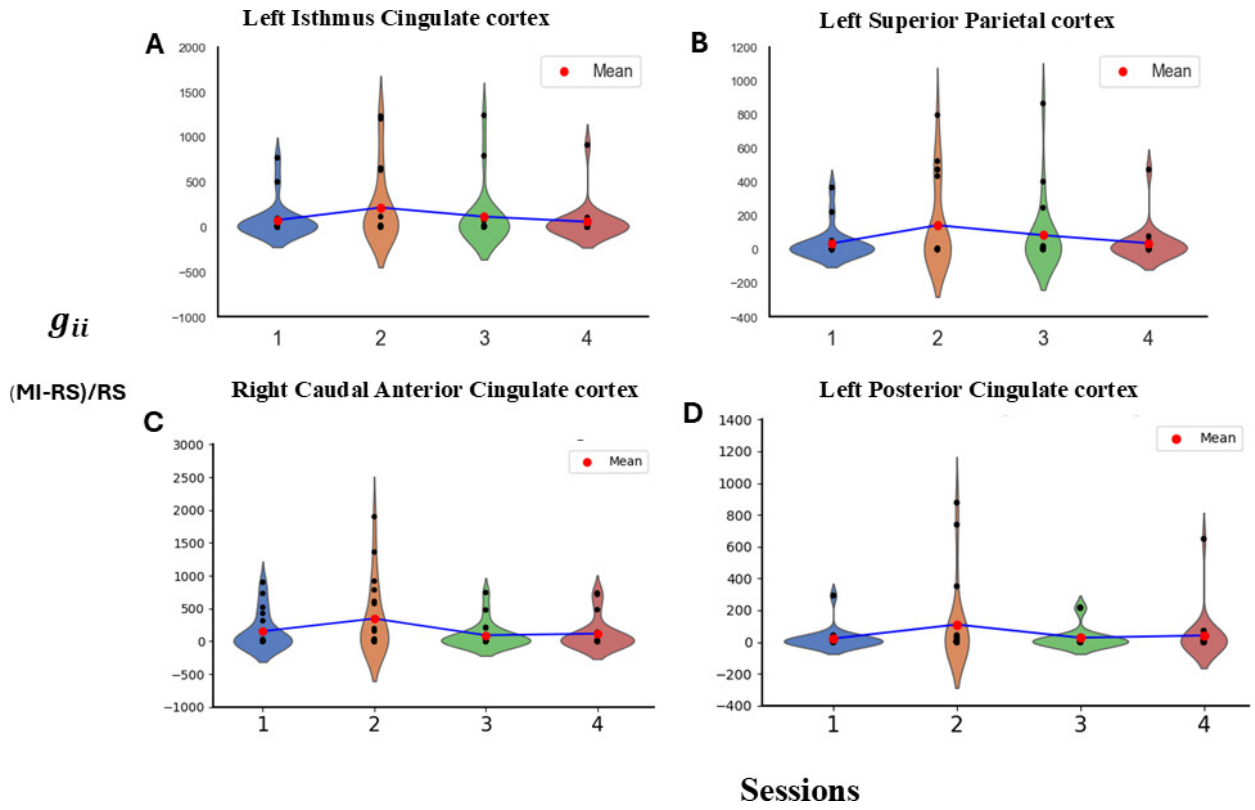

Figure S6: **Substantial changes in relative values ($\frac{\text{MI}}{\text{RS}}$-1) of g_{ii} over time in EEG– session after session:** The left isthmus cingulate (A), superior parietal (B), posterior cingulate (D) and Right caudal anterior cingulate cortex–In each of these regions we see there is a notable change in the mean value of the distribution of 19 relative g_{ii} values over time, which tells the effects of sessions in these regions for BCI learning.

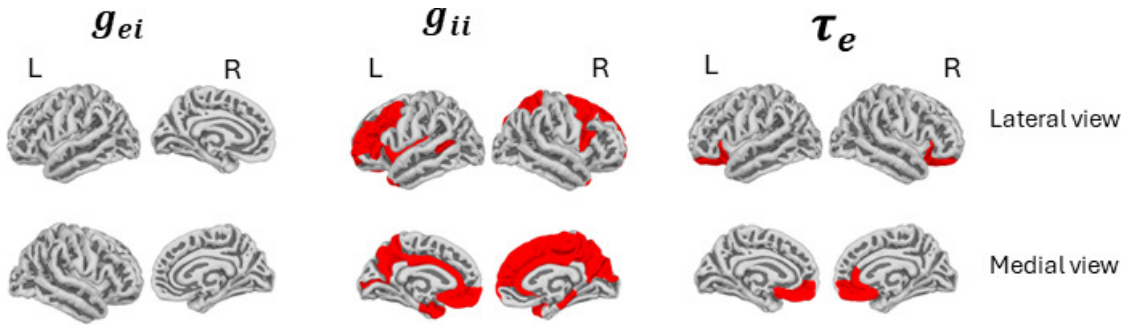

Figure S7: **Condition effect analysis for Independent data set:**The highlighted red colored regions show statistically significant ($P_{\text{FDR}} < 0.05$) condition effect between MI and RS for the three EEG fitted model parameters (g_{ei}, g_{ii}, τ_e) respectively. This show how various cortical regions are getting activated for MI BCI task.

Table ST1: Significant ROIs ($P_{\text{FDR}} < 0.05$) across MEG sessions, grouped by lobe

Session	g_{ei}	g_{ii}	τ_{e}
Session 1	Parietal (Right inferior and superior Parietal), Occipital (Left lateral Occipital), Frontal (Left pars Opercularis)	Temporal (Left transverse Temporal)	Temporal (Left banks of superior temporal sulcus (STS), Left Fusiform, Left inferior and middle Temporal, Left pars Opercularis, Left and right transverse Temporal), Occipital (Left and right Cuneus, Left and right Pericalcarine), Frontal (Left and right caudal middle Frontal, Left and right Precentral, Right superior Frontal), Parietal (Left Paracentral, Left and right Postcentral, Right Precuneus, Left and right Supramarginal), Limbic (Right Entorhinal, Right posterior Cingulate)
Session 2	Nil	Temporal (Left transverse Temporal)	Occipital (Left and right Cuneus, Left and right Pericalcarine), Temporal (Left Fusiform, Left Insula, Left pars Opercularis, Right pars Triangularis, Left transverse Temporal, Right transverse Temporal), Frontal (Right medial Orbitofrontal, Right Precentral), Parietal (Left Paracentral, Right Postcentral, Right Precuneus, Left superior Parietal, Right Supramarginal), Limbic (Left posterior Cingulate, Left rostral anterior Cingulate)
Session 3	Nil	Parietal (Left inferior parietal, Left lateral Occipital), Frontal (Left rostral middle frontal), Temporal (Left and right transverse temporal), Limbic (Right posterior Cingulate)	Temporal (Left banks STS, Left Inferior temporal, Left pars Opercularis, Left and right transverse temporal), Occipital (Left and right Cuneus, Left pericalcarine), Frontal (Right caudal middle Frontal, Left and right Precentral, Left superior Frontal), Parietal (Left inferior Parietal, Left and right Postcentral, Left and right superior Parietal, Right superior temporal, Left and right Supramarginal), Limbic (Right posterior Cingulate)
Session 4	Nil	Parietal (Right postcentral), Temporal (Left transverse temporal)	Temporal (Left banks STS, Left transverse temporal), Occipital (Left cuneus, Left pericalcarine), Frontal (Right precentral), Parietal (Left and right inferior parietal, Left and right paracentral, Left and right postcentral, Left and right supramarginal), Limbic (Left entorhinal, Right isthmus cingulate, Left and right posterior cingulate, Left and right precuneus)

Table ST2: Significant ROIs ($P_{\text{FDR}} < 0.05$) across EEG sessions

Session	g_{ei}	g_{ii}	τ_{e}
Session 1	Nil	Nil	Nil
Session 2	Nil	Nil	Left Cuneus
Session 3	Nil	Nil	Nil
Session 4	Left and right Cuneus, Left inferior Parietal, Left lateral Occipital, Left Precentral	Left and right Cuneus, Left isthmus Cingulate, Left Lingual, Right Parahippocampal	Right Supramarginal

Neural mechanisms of training in Brain-Computer Interface: A Biophysical modeling approach

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[3]Department of Chemical Engineering, Indian Institute of Technology Madras, Chennai, India
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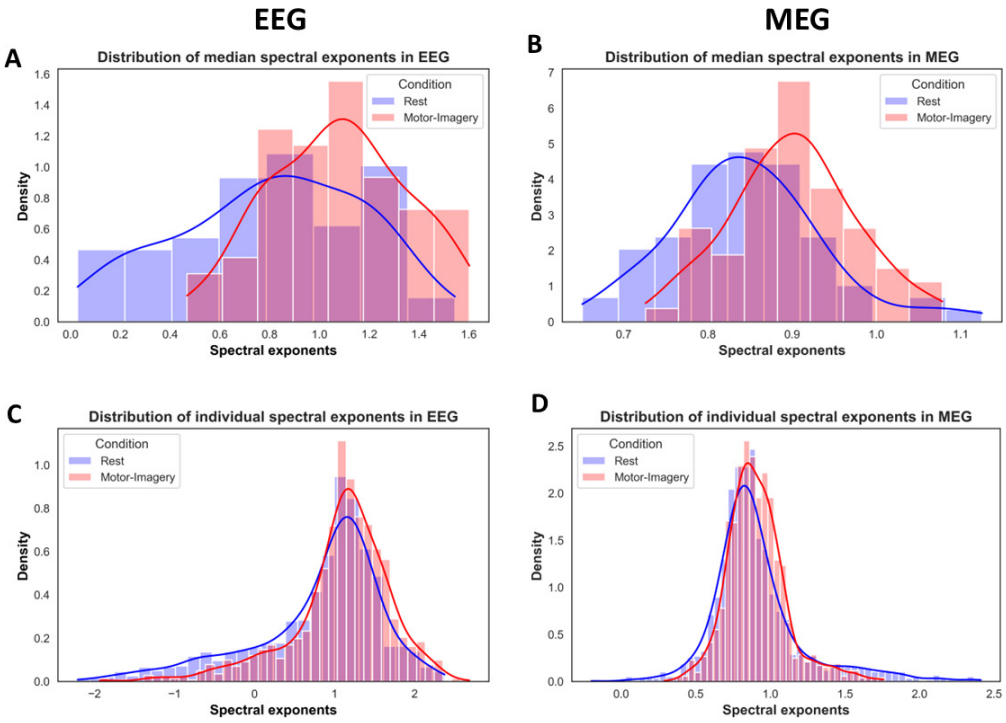


Figure 1: **Distribution of spectral exponents in motor-imagery (MI) and Rest (RS) condition:** A, B exhibit histograms of median spectral exponents in both the condition and show that the mean of total exponents in RS is less than in MI in EEG and MEG, respectively; whereas C and D exhibit histogram distribution of individual spectral exponents in MI and RS, with exponents in both conditions nearly completely overlapping in both EEG and MEG, respectively.

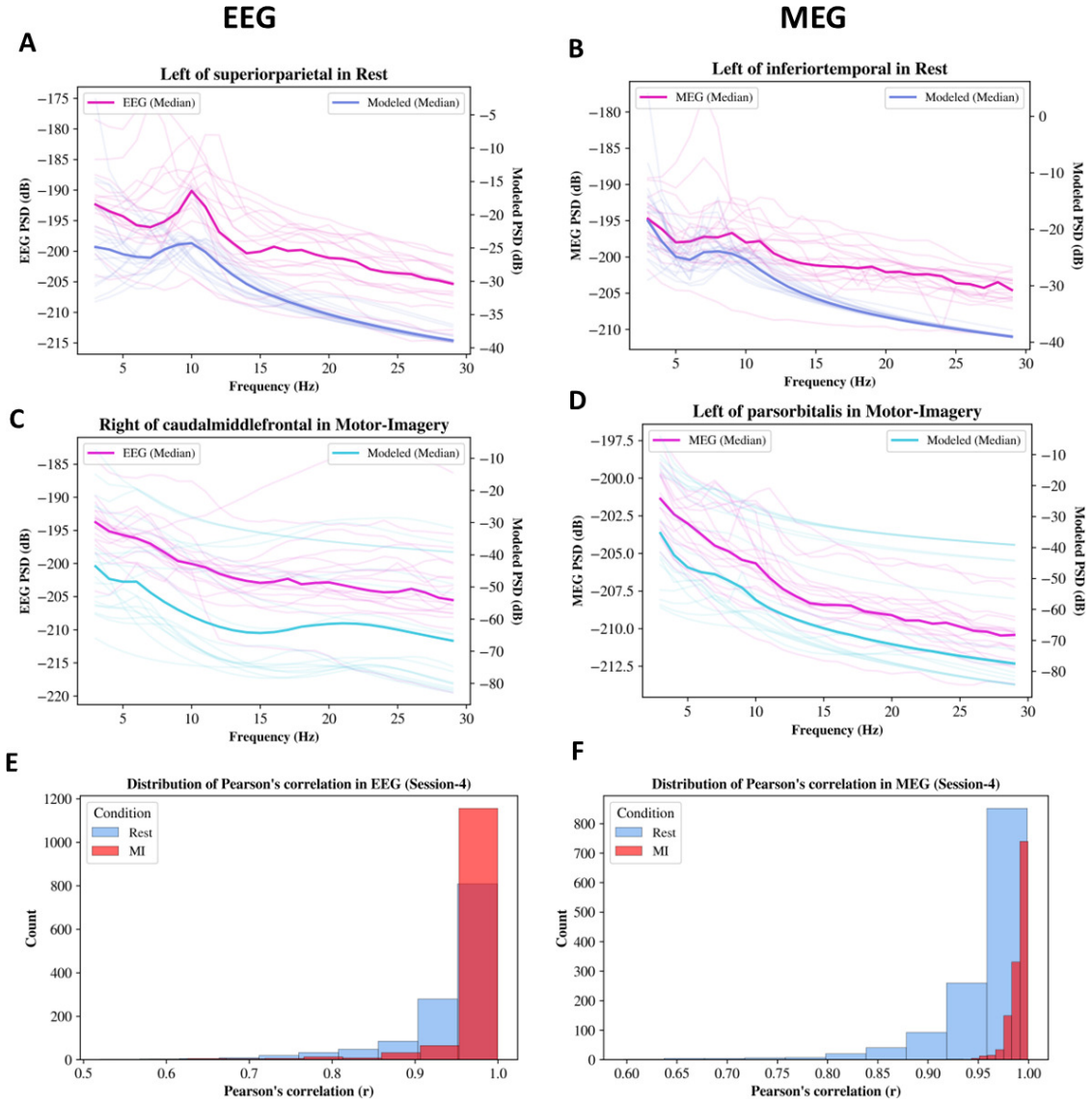


Figure 2: **Matching shape of modeled and actual PSD, with pearson correlation coffecient in both RS and MI**: A, B: Modeled PSD with EEG and MEG PSD shape in dB scale along frequency for Rest in case of Left superior parietal and left inferior temporal region, respectively. It shows how the spectra is changing in 19 subjects, and bold colored spectras (medians of the 19 modeled and actual E/MEG spectra) showing matching of the shape between model and actual median; C, D: Similar to (A, B) these two are for Right Caudal middle frontal and Left pars orbitalis in MI condition, respectively; E, F: Session 4 distribution of Pearson correlation coefficient between modeled and actual spectra for MI (red) and Rest (blue) condition in EEG and MEG, respectively;

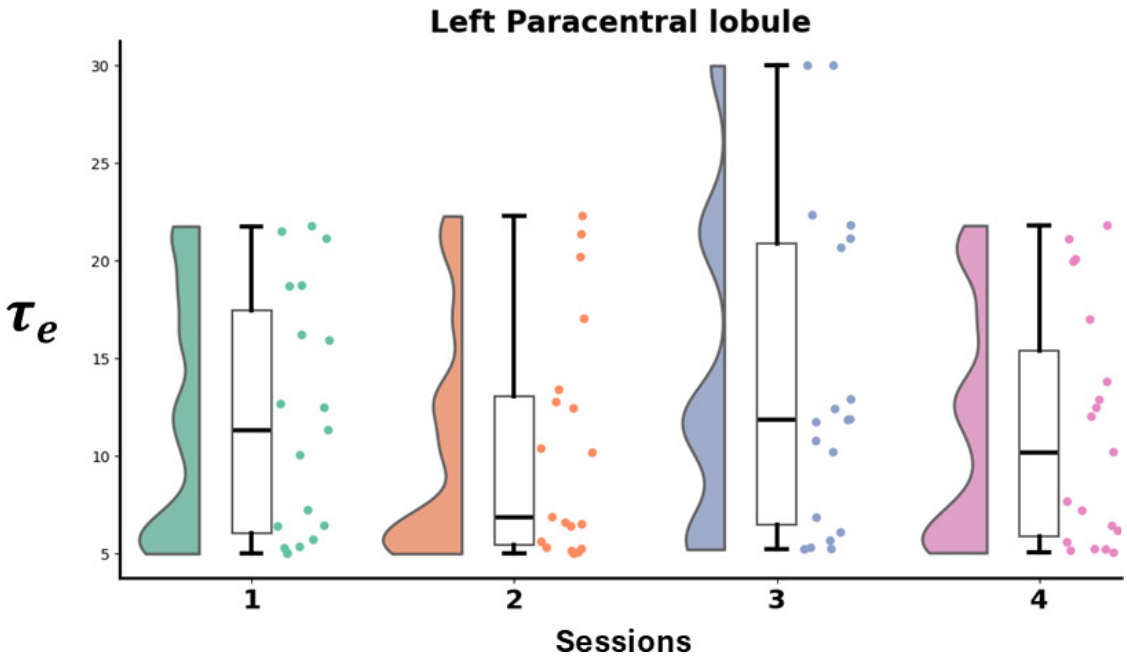
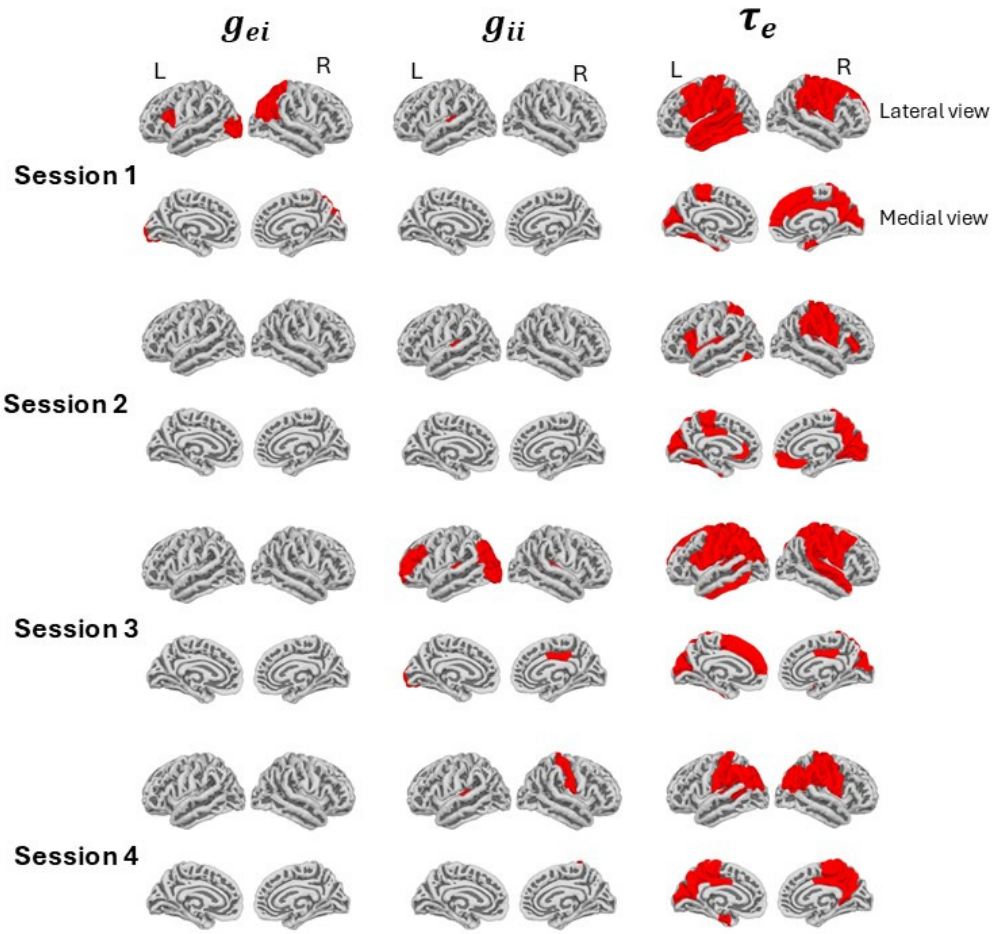


Figure 3: **Variation in τ_e model parameter driven by MI condition:** Distribution of 19 τ_e values in each session show inter-session variation in the left paracentral region during MI in MEG modeled spectra. Substantial variation has been observed due to MI (P = 0.0452; Bonferroni adjusted P = 0.0904), which is a key indicator of session effect driven by motor-imagery on group performance in BCI learning.



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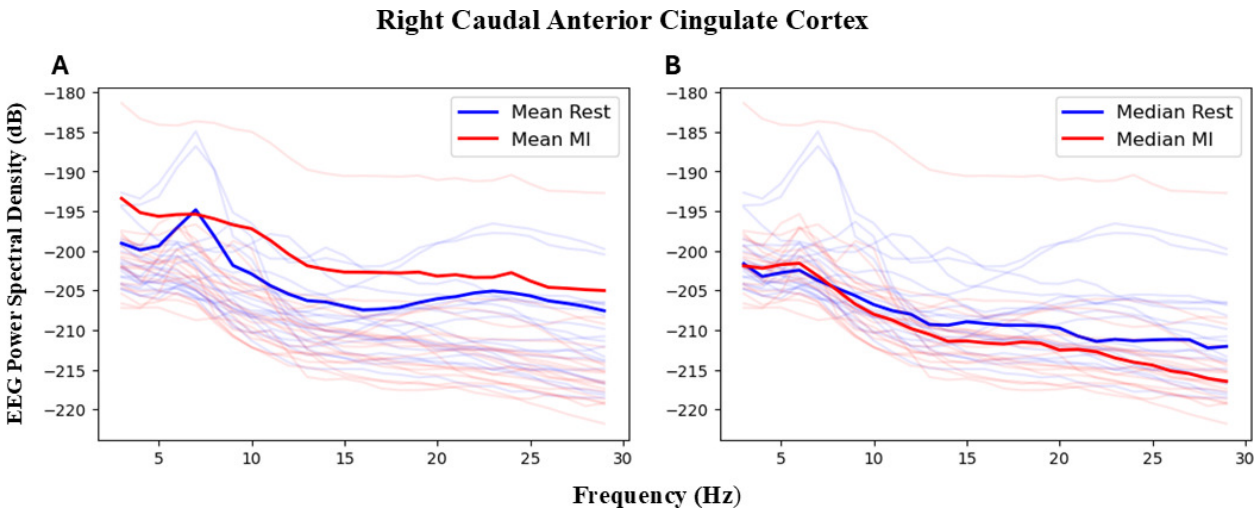

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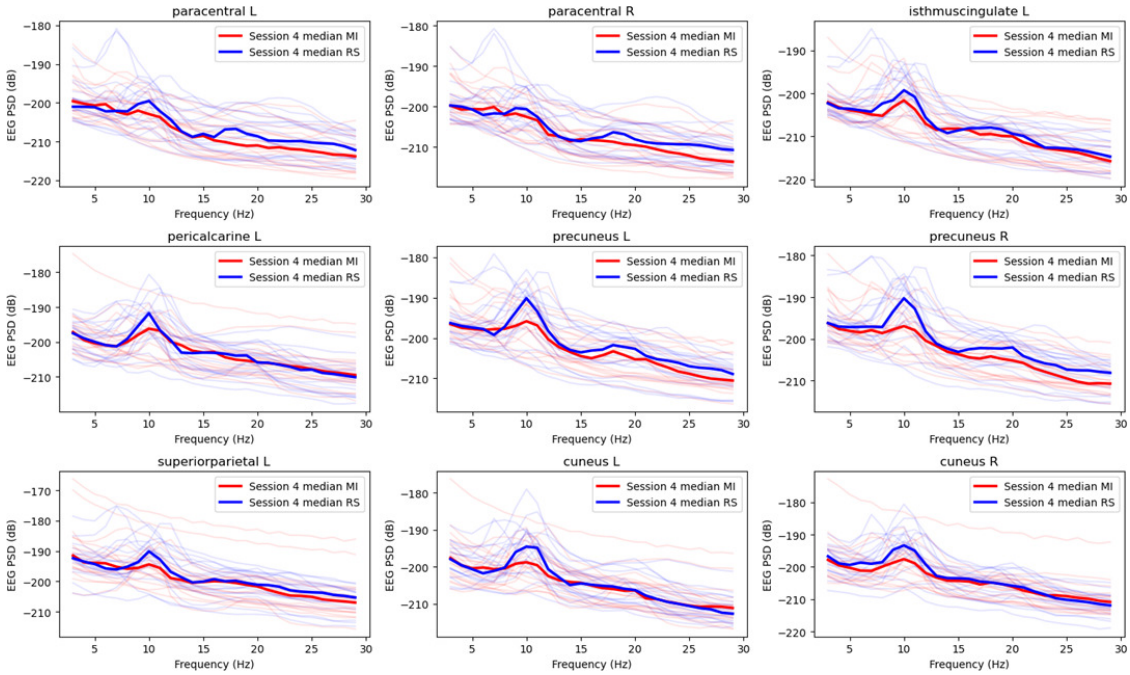

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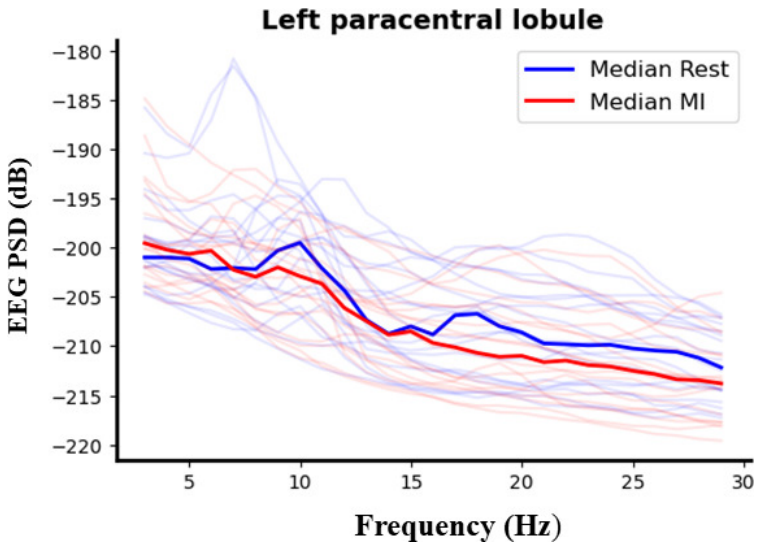

Figure 7: **Changes in EEG-Power Spectral Density (PSD) in Left paracentral lobule:** It shows that the changes in PSD with frequency in case of Rest and MI condition, with median of the distribution of 19 spectra in both the condition. The alpha and beta peak can be observed in both the condition.

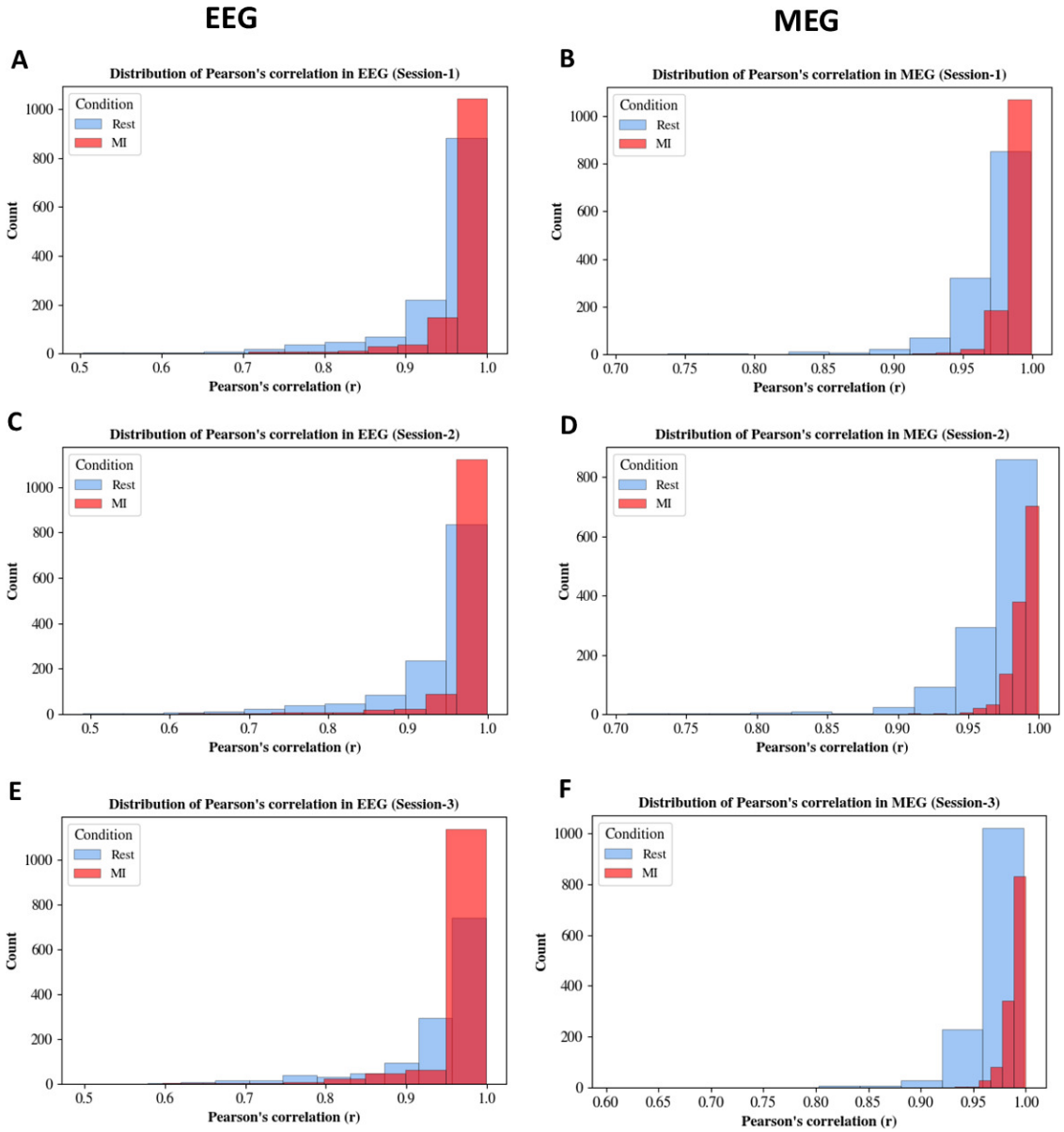

Figure 8: **Pearson correlation coffecient between modeled and actual PSD in both Rest (RS) and MI**: A, B: Shows the distribution of the coeffecient to be more than 0.90 in most cases in both RS and MI condition in session 1 EEG and MEG, respectively; C,D: Similar to the above fig, here it shows in session 2 EEG, MEG respectively; E, F: it Shows the distribution in session 3 EEG, MEG respectively

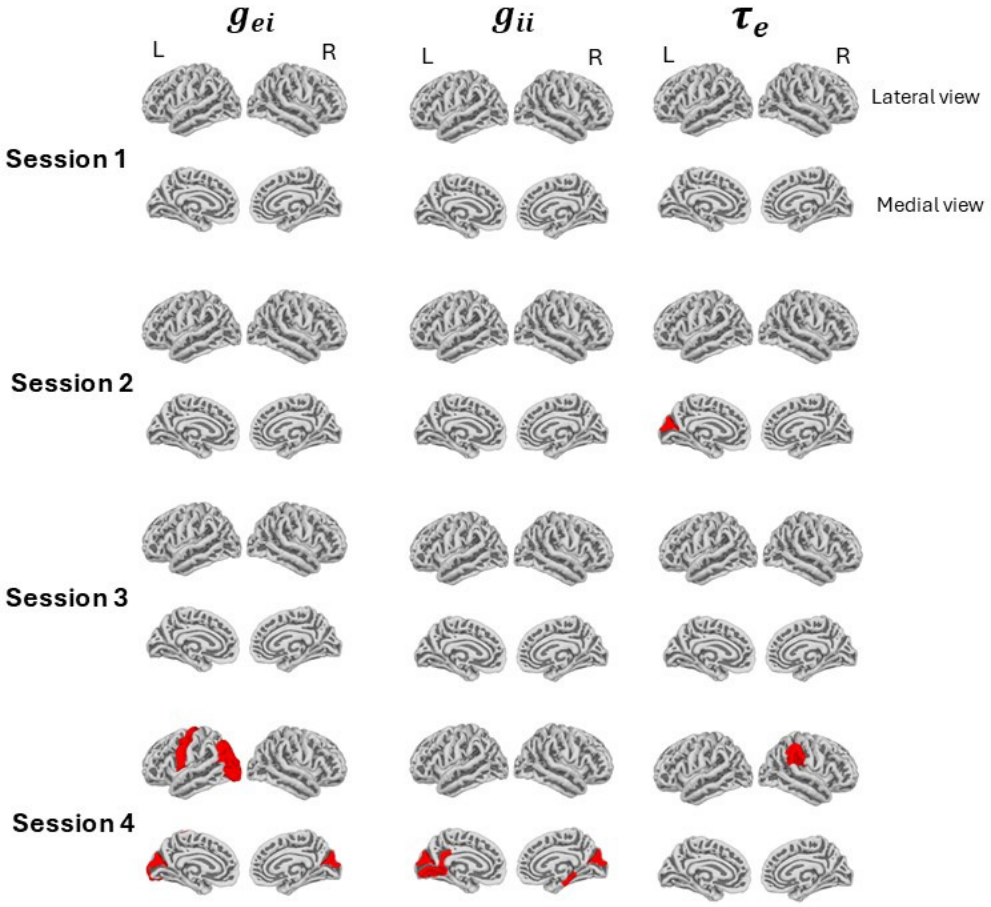

Figure 9: **EEG-based condition effect analysis in BCI learning:** The highlighted red colored regions show statistically significant ($P_{\text{FDR}} < 0.05$) condition effect for MI Vs RS for the three EEG fitted model parameters (g_{ei}, g_{ii}, τ_e) respectively in every session. This show how functional reorganization happening over time in the 19 subjects' group and various cortical regions are getting activated for MI BCI task.

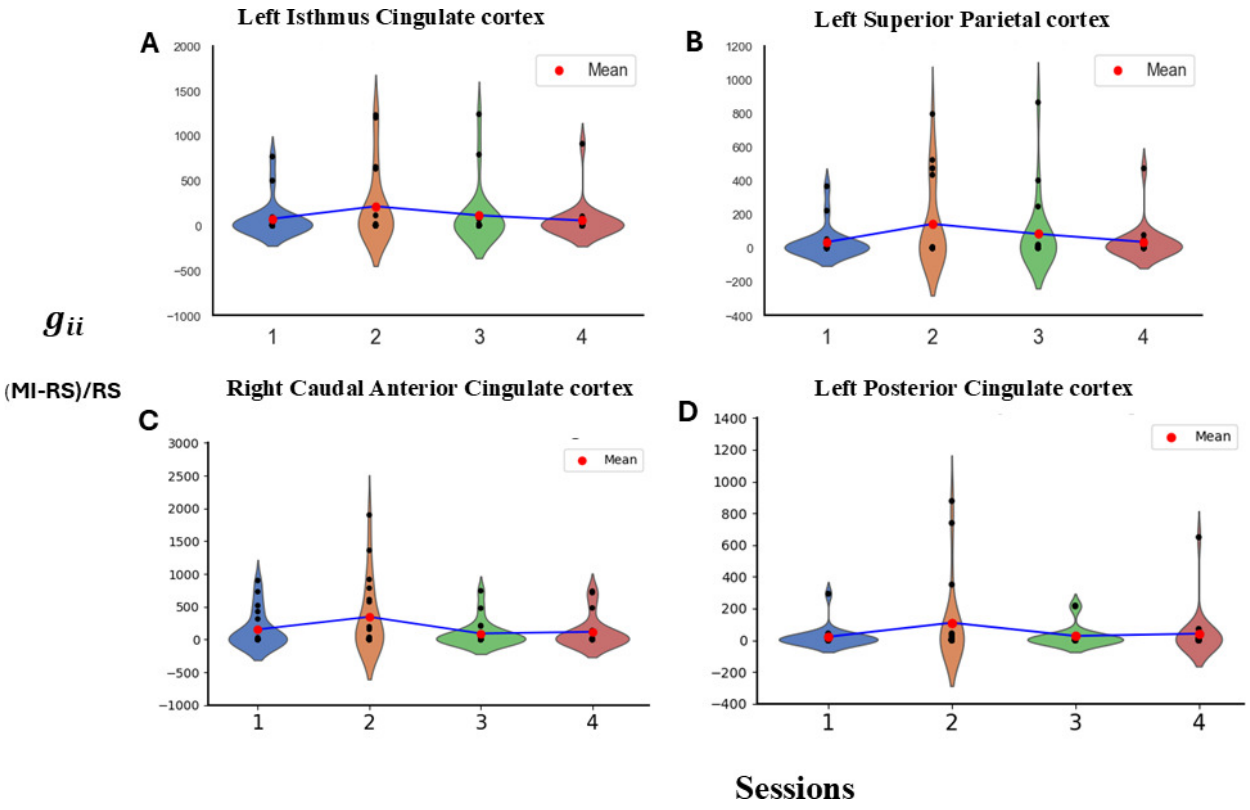

Figure 10: **Substantial changes in relative values ($\frac{\text{MI}}{\text{RS}}$-1) of g_{ii} over time in EEG– session after session:** The left isthmus cingulate (A), superior parietal (B), posterior cingulate (D) and Right caudal anterior cingulate cortex–In each of these regions we see there is a notable change in the mean value of the distribution of 19 relative g_{ii} values over time, which tells the effects of sessions in these regions for BCI learning.

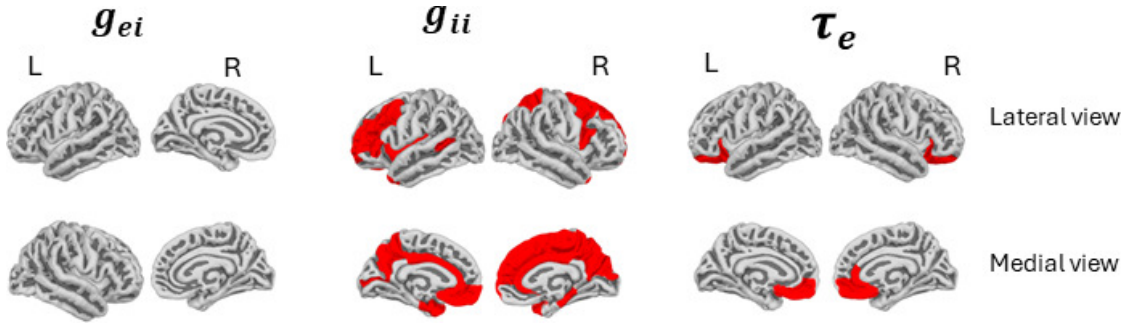

Figure 11: **Condition effect analysis for Independent data set:**The highlighted red colored regions show statistically significant ($P_{\text{FDR}} < 0.05$) condition effect between MI and RS for the three EEG fitted model parameters (g_{ei}, g_{ii}, τ_e) respectively. This show how various cortical regions are getting activated for MI BCI task.