Identification of the Default Mode Network with Electroencephalography

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Abstract—The Default Mode Network (DMN) is a brain resting-state network that is closely linked to consciousness and neuropsychiatric disorders. The DMN is routinely identified with functional magnetic resonance imaging (fMRI) or positron emission tomography (PET). However, both of these methods impose restrictions on the groups of patients that can be examined. We show that the DMN can also be identified by electroencephalography (EEG). Instructing subjects to alternate between self-referential memory recall and focusing on their breathing induces a spatial pattern of spectral band power modulation in the $\theta$- and $\alpha$-band (4–16 Hz) that is consistent with the DMN pattern observed with PET and fMRI. Since EEG is a portable, safe, and cheap technology, our work enables the characterization of DMN alterations in patient groups that are difficult to study with fMRI or PET.

I. INTRODUCTION

The brain is composed of large-scale cortical networks that are intimately linked to high-level cognition \cite{1, 2}. Among these networks, the default mode network (DMN) \cite{3}, comprising the precuneus/posterior cingulate cortex, medial prefrontal cortex, and the temporoparietal junction, is of particular interest, because it has been linked to neuropsychiatric disorders \cite{4} and to the degree of consciousness \cite{5}.

A variety of neuroimaging techniques have been used to analyse the DMN. Functional Magnetic Resonance Imaging (fMRI) or Positron Emission Tomography (PET) are routinely used to identify the DMN, either by contrasting resting-state to task-induced DMN deactivation levels \cite{3, 6} or by a functional connectivity analysis on resting-state recordings \cite{7}. Several attempts have been also made to recover the DMN from magnetoencephalographic recordings (MEG) \cite{8, 9}. Using resting-state data, De Pasquale et al. identified MEG correspondents of DMN with a topography of interregional band power correlations in the $\theta$- (3.5–7 Hz), $\alpha$- (8–13 Hz), and $\beta$- (14–25 Hz) band \cite{8}. Brookes et al. identified MEG signatures of DMN activity by amplitude envelope correlations in the $\alpha$-band (8–13 Hz) \cite{9}.

However, fMRI, PET and MEG are difficult to perform on certain groups of patients, such as severely paralysed patients in late stages of amyotrophic lateral sclerosis (ALS), that are dependent on artificial ventilation systems and thus cannot be easily put into the MRI, PET, or MEG scanners. In contrast, EEG is a portable, safe (non-invasive), cheap, and widespread technology, that can be used at the patient’s home. EEG-based DMN characterisation would enable the investigation of alterations in DMN activity in a wide range of patients groups that are difficult to examine with other methods. In particular, the connectivity within the DMN is negatively correlated with the degree of consciousness impairment and thus could be used to distinguish the conscious state from the vegetative state in CLIS \cite{5} patients, for whom the degree of consciousness cannot be concluded from behaviour due to the absence of communication.

Previous attempts to identify the DMN with EEG were only partially successful. Knyazev et al. partially reproduced DMN spatial features by first estimating current source density, then applying independent component analysis (ICA), and comparing the resulting resting-state $\alpha$ band (8–12 Hz) activation of the obtained ICs to that during social cognition tasks \cite{10}. However, spatial overlap of the identified EEG-based DMN pattern with that of the fMRI-based DMN was restricted to only one node of the DMN (precuneus/posterior cingulate cortex). This prevented an analysis of the connectivity within the DMN, which is relevant for characterizing the degree of consciousness \cite{5}.

We devised a novel behavioural paradigm that allows us to obtain an EEG-based DMN pattern more similar to the regions identified by fMRI. We instructed healthy subjects to alternate between two experimental conditions, recalling of positive autobiographical memories and focusing on breathing, and then computed dynamic Statistical Parametric Maps (dSPM) \cite{11} from high density EEG recordings in the two conditions. Comparing source level activations, we found $\theta$- and $\alpha$ band power changes in the medial prefrontal cortex, the posterior cingulate cortex, and the temporoparietal junction - a pattern that is highly consistent with the DMN.

II. METHODS

A. Subjects

EEG data was recorded from eleven healthy subjects (eight male and three female, mean age 29.3 ± 8.3 years). All subjects gave informed consent to participate in the study according to guidelines set by the Max Planck Society and received 12 Euro per hour for their participation. The study was approved by an ethics committee of the Max Planck Society.

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B. Experimental Data

All data was recorded with 124 active electrodes at 500 Hz sampling frequency using actiCAP active electrodes and a BrainAmp electroencephalograph (both provided by BrainProducts GmbH, Gilching, Germany). Electrodes were placed according to the extended 10-20 system, with the electrode P7P as the initial reference. All recordings were converted to common average reference. The stimuli presentation was realised with the BCI2000 and BCPy2000 toolboxes [12].

First, two resting states (eyes-open and eyes-closed) were recorded. For each resting state, subjects were placed in front of a computer screen at a distance of 1.25 ± 0.2 m and were instructed to relax and let their mind wander for five minutes. In the eyes-open condition, they were additionally asked to fixate on a cross in the middle of the screen. After that, the EEG in the conditions of self-reflective thoughts and breathing was recorded in three blocks with a short break (1-5 minutes) in between. Each block consisted of ten trials for each condition presented in pseudo-randomized order, resulting in 30 trials for each condition. In the beginning of each trial, instructions appeared on the screen, asking subjects either to recall positive autobiographic memories for each condition presented in pseudo-randomized order, or to concentrate on their breathing. Simultaneously, the subjects either to recall positive autobiographic memories or to concentrate on their breathing. Simultaneously, the same instructions were read out by a male voice. After four seconds, the instructions disappeared and subjects performed the announced task while fixating on the white cross. After one minute, the word "Pause" appeared on the screen, indicating the end of the trial. The pause lasted for 5.5±0.5 seconds, then the new trial began.

Due to technical problems, subject 9 had different number of trials per condition (27 and 33 trials). For that subject only the first 27 trials for each condition were used for the analysis.

C. Data analysis

1) Preprocessing: We used the time window 4–30 seconds of each trial and restricted our analysis to a combination of θ and α frequency bands (4–16 Hz, individually adjusted for each subject). The lower θ boundary was set to 4 Hz for all the subjects, while the upper α boundary was determined individually for each subject by determining the intersection of the spectral power of channel Oz between eyes-open and eyes-closed resting states [13]. For subject 5, the eyes-closed recordings were corrupted by noise, so the upper boundary of the individual α band was set to 14 Hz. The data was then bandpass-filtered with a 3rd order Butterworth filter in the θ- and in the α- frequency band, respectively, and downsampled to 50 Hz.

2) Dynamic Statistical Parametric Mapping: To project the sensor activations on the source level, we applied dSPM, a noise-normalized minimum norm estimate, to the preprocessed data. [11]. First, the forward model x[t] = As(t) was computed, with the matrix A specifying the projection of K = 3 × 15028 current dipoles spread over the cortex s[t] ∈ RK on the N = 124 electrodes x[t] ∈ RN. We generated the forward model with the BrainStorm toolbox [14], using standardized electrode locations and a standardized three-shell spherical head model. Then, the activity of each source was estimated from the measurements of the electrical potential on the surface of the scalp at N electrode locations as described in [11]:

\[ \hat{s}(t) = Wx(t), \quad W = \Sigma A^T (\Sigma A^T + C)^{-1}. \]  

Here, \( \Sigma \) is the spatial covariance of the dipole strength vector \( \Sigma = s(t)s(t)^T \), approximated by the identity matrix, and C is the sensor noise covariance matrix, computed individually for each subject from their eyes-open resting-state data. The estimated time series \( \hat{s}(t) \) were then normalized by the noise variance, leading to noise normalized activity estimate \( \tilde{z}_i(t) \) at each time point t and location i [11]:

\[ \tilde{z}_i(t) = \frac{\hat{s}_i(t)}{\sqrt{w_i^TCw_i^T}}, \]

where \( w_i \) is the i\(^{th}\) row of the unmixing matrix W. We then estimated a noise-normalized current dipole power at each time point t and location i [11]. We made no assumptions on dipole orientation and thus averaged three dipoles for each location:

\[ \tilde{q}_i(t) = \frac{\sum_{j \in G_i} \tilde{s}_j^2(t)}{\sum_{j \in G_i} w_j^TCw_j^T}, \]

where \( G_i \) is the set of indices of dipoles located at i. As a last step, we averaged the current dipole power over the 4–30 seconds of each trial.

3) Statistical testing: We computed the signed coefficient of determination (signed \( R^2 \)) for every subject and source to evaluate condition-induced differences between band power, averaged across the θ- and the α-band (4–16 Hz). To test the null-hypothesis \( H_0: R^2 = 0 \) that there is no difference between the two conditions on the group level we first estimated p-values for each subject. For that we randomly permuted the condition labels of the trials 10^3 times. We then counted the instances in which the resulting \( |R^2_{HO}| \) exceeded \( |R^2| \) and estimated the probability that \( |R^2_{HO}| > |R^2| \). We thereby obtained a p-value for each of the \( K = 15028 \) sources and \( M = 11 \) subjects. Then, we computed the empirical cumulative distribution function (CDF) of these p-values across subjects for each source and quantified its deviations from a uniform CDF with support from zero to one by integrating the differences between the two CDFs. We drew samples from the uniform CDF 10^3 times. This enabled us to estimate the probability of observing the obtained p-values under \( H_0 \), because by construction p-values from a null-distribution are uniformly distributed between zero and one. As a last step, we corrected the significance level using a false discovery rate (FDR) of \( \alpha_{FDR} = 0.05 \) [15] to compensate for the multiple comparisons for each of the \( K \) cortical sources.

To plot the results, we averaged the signed \( R^2 \) across subjects and set the signed \( R^2 = 0 \) for the sources for which we could not reject the null-hypothesis.
III. RESULTS

Figure 1 displays the sources that we found to show a statistically significant modulation on the group-level. We find the most prominent modulation of band power in the posterior cingulate cortex, which constitutes a hub of the DMN [16]. In addition, we observe band power modulation in the medial prefrontal cortex and in the left temporo-parietal junction. With the exception of the right temporo-parietal junction, our method thus identifies the core areas of the DMN [3].

IV. DISCUSSION

Using the cognitive strategy of alternation between autobiographical memories and focusing on breathing, we identified a pattern of EEG band power modulation that is highly consistent with the DMN as characterized by PET [6], [18] and fMRI [3], [7], [17]. This is of particular significance for two reasons. First, this EEG-based identification of the DMN enables us to study oscillatory properties of the DMN that are not accessible by PET or fMRI. And second, our work makes it feasible to study DMN alterations in patient
groups that are difficult to study by with PET, fMRI and MEG, such as severely paralysed patients in late stages of ALS.

We note, however, that the EEG-based DMN nodes are smaller than those found by fMRI, despite the fact that fMRI has a higher spatial resolution. One potential reason for this observation is the low spatial resolution of EEG source localization methods. Due to inter-subjects differences in head shape and cortex folding, the spatial overlap between individual DMN patterns may be smaller than each individual DMN pattern on its own. On the group level, this may lead to a spatial underestimation of the DMN. This problem could be addressed by using individualised EEG forward models derived from structural MRI scans.

The difference between the two conditions arises from the properties of the DMN: self-referential thoughts, such as autobiographical memories, activate the DMN [6], [18], while task-related activity free from self-referential thoughts and memories, such as focusing on the breathing, induces DMN deactivation [3]. To capture the effect of self-referential processing, we restricted our analysis to the θ and α frequency bands (4-16 Hz), following the previous work of Mu et al. who correlated self-referential processing with spectral power in the θ and in the α range [19]. We combined the two bands, since PET and fMRI analysis are not frequency-specific and the DMN pattern obtained with those methods is likely to consist of neurons oscillating in different frequency ranges. Our choice of tasks, however, could also lead to underestimation of the extent of the DMN. While focusing on breathing has been shown to deactivate the DMN, it also increases synchronization within the DMN [20]. Because EEG is sensitive to synchronized oscillations, the increase in synchronisation may lead to an increase in the spectral power that is not distinguishable from changes in activation levels of the involved cortical areas. Further refining the behavioural paradigm may alleviate this problem.

Future studies need to investigate if EEG-based connectivity measures between nodes of the DMN correlate with levels consciousness in similar ways to fMRI-based connectivity patterns [5]. The EEG-based identification of the DMN brings a new exciting possibility of DMN characterisation to groups of patients that are difficult to study with other methods.

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REFERENCES