Scalp EEG Continuous Space ERD/ERS Quantification

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Abstract. In the scope of EEG applications such as Brain Computer Interfaces (BCI) or the evaluation of epileptic activity, the detection of event-related potentials (ERP) and associated event-related desynchronization / synchronization (ERD/ERS) are common goals. The most commonly used method for assessing ERD/ERS consists on the evaluation of EEG power changes upon the event onset in relation to the baseline, reflecting increased/decreased synchronization. Phase synchronization measures have also been used for this purpose, both across event trials as well as between pairs of electrodes. Here, we propose a 2D spatially continuous extension of the *Phase Locking Factor* (PLF) metric for ERD/ERS quantification, called *PLF Field* (PLFF), based on measuring spatial variations of the EEG phase. A continuous phase map is estimated from the discrete set of EEG traces by using the Hilbert transform in the analytical signals framework. The synchronization at each arbitrary spatial location on the scalp space is then computed from the magnitude of the phase gradient at that location. The method is illustrated with EEG data obtained in two motor tasks paradigms. Our results indicate that the proposed approach is largely consistent with the conventional power-based ERD/ERS method, but may provide additional information in studies of neuronal synchronization using EEG.

Keywords: EEG, ERD/ERS, Phase locking factor, Hilbert transform.

1 Introduction

The electroencephalogram (EEG) is one of the most common techniques for brain activity assessment and it is very well established in the clinical practice for diagnostic purposes. Recently, it has received additional attention and developments in the scope of emergent techniques like Brain Computer Interfaces (BCI) or simultaneous EEG-fMRI. In these modalities, detection of event-related potentials (ERP) and associated event-related desynchronization / synchronization (ERD/ERS) are common goals in several scientific and clinical scopes, such as the detection of motor movement or epileptic activity.

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EEG signals are thought to reflect synchronized neuronal electrical activity, such that EEG power increases and decreases may be interpreted in terms of relative synchronization and desynchronization of the underlying neuronal populations, respectively [4,1,2]. The EEG power dependency on the synchronization state of neuronal masses can be explained by using a stochastic reasoning [8]. The traditional methods to detect event-related desynchronization / synchronization (ERD/ERS) are therefore based on measuring the EEG power changes at each electrode location in relation with an event onset.

During rest, neighbouring neurons oscillate synchronously, but when they become active the overall synchronization decreases. In particular, during a motor task like voluntary hand movement, desynchronization phenomena occur in the upper alpha and low beta frequency bands across the motor cortex [2]. Such desynchronization is observed predominantly in the hemisphere contralateral to the movement, close to the scalp projection of the hand area (electrode C3 or C4, for right or left hand movements, respectively), prior to movement onset. Immediately after movement onset, the desynchronization becomes bilateral and nearly symmetrical [3]. This phenomenon is usually quantified as the relative change in spectral power elicited by the event in relation to the baseline period [2], assuming that an increase / decrease of the EEG power in a certain frequency band corresponds to the synchronization / desynchronization of the underlying neuronal population's activity in those frequencies, respectively.

Synchronization state changes of the neuronal masses reflected at each electrode should also be reflected on the synchronization between electrodes. That is, the same phenomenon observed in time at each electrode is also observed in space, namely, by measuring the temporal synchronization between electrodes. Therefore, ERD/ERS could alternatively also be assessed based on the phase differences across EEG channels, which are thought to reflect different degrees of synchronization between the underlying neuronal populations [5]. The *Phase-Locking Factor* (PLF), which is based on the phase difference between two signals, has previously been used as a measure of synchronization between the EEG signals at two electrode locations [6,7].

In this paper, we propose a novel method for ERD/ERS assessment based on an extension of the PLF approach whereby phase synchronization is quantified along the scalp space in a continuous manner. A continuous phase map is estimated from the discrete set of EEG traces by using the Hilbert transform in the analytical signals framework. The synchronization at each arbitrary spatial location on the scalp space is then computed from the magnitude of the phase gradient at that location. A PLF field (PLFF) is hence defined for the entire scalp on a continuous basis. The application of the method is illustrated on real data acquired from one subject performing two different motor tasks and the results are compared with the traditional method based on the EEG power. It is shown that both methods lead to consistent results, but the proposed method is potentialy more discriminative in spatial terms with respect to the detection and quantification of ERD/ERS.

2 Methods

In this section, the problem formulation is first presented, leading to our definition of the PLFF, and the experimental procedure employed for the acquisition of the EEG data used to illustrated the applicability of the proposed method is then described.

2.1 Problem Formulation

Let $s_k(t)$ be the k^{th} EEG time signal located at position (x_k, y_k) of the scalp space, according the standard topographic map displayed in Fig.1.

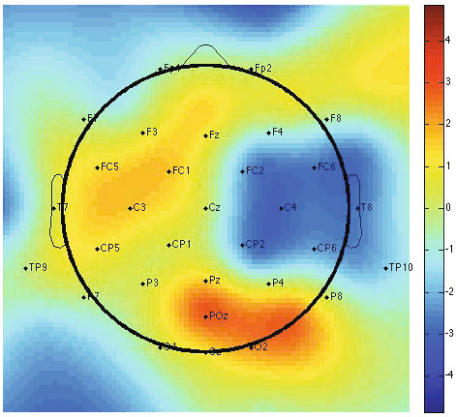

Fig. 1. Topographic spatial location of 31 EEG electrodes on the scalp space, according the standard 10-20 system, shown overlaid on an example of the interpolated phase field, $\Phi(x, y)$ (rad).

The corresponding analytical signals $a_k(t)$ are obtained using the Hilbert transform operator ($\mathcal{TH}$) [10], as:

$$a_k(t) = s_k(t) + i\mathcal{TH}(s_k(t)) = \rho_k(t)e^{i\phi_k(t)} \tag{1}$$

where $\rho_k(t)$ and $\phi_k(t)$ are the magnitude and phase of $s_k(t)$, respectively. The PLF for a pair of signals [7] is then defined as:

$$PLF_{kl} = \int_t e^{i|\phi_k(t)-\phi_l(t)|}dt \tag{2}$$

Let us consider the phase signals $\phi_k(t)$ as temporal samples of a continuous $2D$ scalar field, $\Phi(x, y, t)$ at (x_k, y_k) locations:

$$\Phi(x_k, y_k, t) = \phi_k(t). \tag{3}$$

The scalar function $\Phi(x, y, t)$ can now be obtained from $\Phi(x_k, y_k, t)$ using the biharmonc interpolation method [9], for each time instant. The generalized continuous PLF field (PLFF), is then computed according to:

$$PLFF(x, y) = \int_t e^{i\|\nabla_{x,y}\Phi(x,y,t)\|} dt \tag{4}$$

where $(x, y) \in \Omega \subset R^2$ and the difference of phases in Eq.(2) is replaced by the norm of the gradient vector of the phase field, $\|\nabla_{x,y}\Phi(x, y, t)\|$.

On a discrete time basis, the PLFF is defined as

$$PLFF(x, y) = \left| \frac{1}{N} \sum_n e^{i\|\nabla_{x,y}\Phi(x,y,n)\|} \right| \tag{5}$$

where n are the sample indices of the time interval considered for the analysis.

2.2 Experimental Procedure

EEG data were collected from one healthy female subject while performing two hand motor tasks, involving index finger brisk movements. In one task, self-paced movements were performed (*Self-Paced Movement*) and in the other task visual cues were provided (*Cued Movement*), as described below. All movements consisted of a computer button press, which was recorded by the EEG recorder while the subject sat on a chair in front of a computer screen and keyboard.

Motor Tasks. In the *Self-Paced Movement*, the subject was instructed to keep her eyes closed and to voluntary perform self-paced left index brisk movements at intervals no shorter than 6s. A total of 76 trials were performed. The *Cued Movement* started with the presentation of a fixation cross at the centre of the monitor during a period that varied randomly between 8 and 10s, followed by the letter R or L that appeared on the right or left side of the cross for 1s, respectively. Depending on the letter presented on the screen, the subject was required to perform the movement with her right or left index finger. A total of 90 trials were performed, in 3 runs of 30 trials each, with an equal number of right and left index finger movements in a random order. Only the data correspondent to the right index movement was considered for this analysis.

EEG Acquisition and Analysis. The EEG signals were recorded through a 32 channels system mounted on a scalp cap connected to a BrainAmp MR Plus amplifier (Brain Products). The electrodes impedance was maintained between 11 to 15kΩ and the signals were recorded at a sampling rate of 5kHz, with a low-pass filter at 250Hz and high-pass filter with a cutoff at 10s, using the

Brain Vision Recorder software (Brain Products, Germany). The EEG data were analysed using the EEGlab toolbox (http://sccn.ucsd.edu/eeglab/). The data were down-sampled from 5kHz to 250Hz and segmented into epochs time-locked to the movement onsets.

For the computation of the conventional ERD/ERS measure based on power, the time-frequency analysis of the EEG data is first performed using a Morlet Wavelet transform, as implemented in EEGlab, in a frequency band 5 to 45Hz. The *event-related spectral perturbations* (ERSP) [8] are then computed as the power $P(t, x_k, y_k)$ at each time point in each trial normalized by the mean power during the baseline period, $P_B(x_k, y_k)$, at each electrode location (x_k, y_k), as a function of frequency. This is subsequently averaged across trials and expressed in % (or dB), for each time point t over the trial period, at each electrode location, as:

$$ERSP(t, x_k, y_k) = \frac{P(t, x_k, y_k) - P_B(x_k, y_k)}{P_B(x_k, y_k)} \tag{6}$$

The $PLFF$ is computed according to Eq.(5), for a period of time around each event onset (trial), in 200ms windows with 50% overlap, normalized by the mean $PLFF$ during the baseline period ($PLFF_B(x, y)$) and subsequently averaged across trials and expressed in %, for each time point t over the trial period, at each location in the continuous scalp space (x, y), as:

$$ERD/ERS_{PLFF}(t, x, y) = \frac{PLFF(t, x, y) - PLFF_B(x, y)}{PLFF_B(x, y)} \tag{7}$$

3 Experimental Results

In this section, the results obtained by the proposed method based on the PLFF metric are compared with the results obtained with the traditional power based methods, in data acquired during two motor tasks.

3.1 Trial Time-Courses

The $ERSP$ over the trial period, averaged across the 76 self-paced movement trials, is shown in Fig.2, for channel C4, the one more closely located to the right primary motor cortex, known to be associated with the left hand movement. A decrease in $ERSP$ can clearly be observed prior to movement onset, at 0s, lasting for approximately 500ms, in the β band (14-25Hz). The ERD/ERS time courses obtained using both $ERSP$ in the β band and $PLFF$, in selected locations on the right and left hemispheres, are shown in Fig.3. The expected asymmetry across hemispheres can be found with both measures. The task (A) and baseline (B) periods considered for the analysis were [-0.6;-0.4]s and [0.0;0.2]s, respectively.

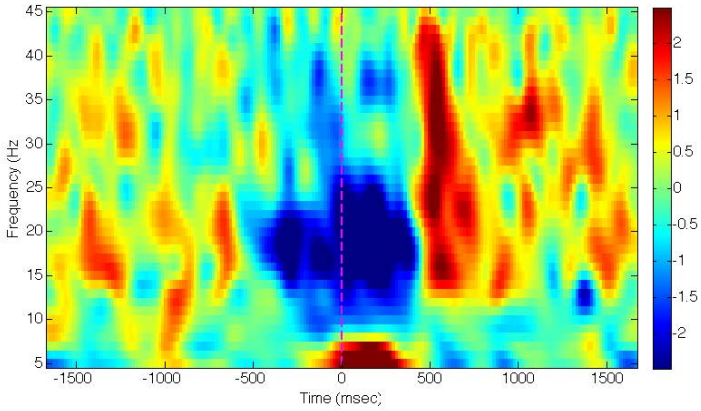

Fig. 2. Trial-average $ERSP$ (dB) of channel C4 for the self-paced movement.

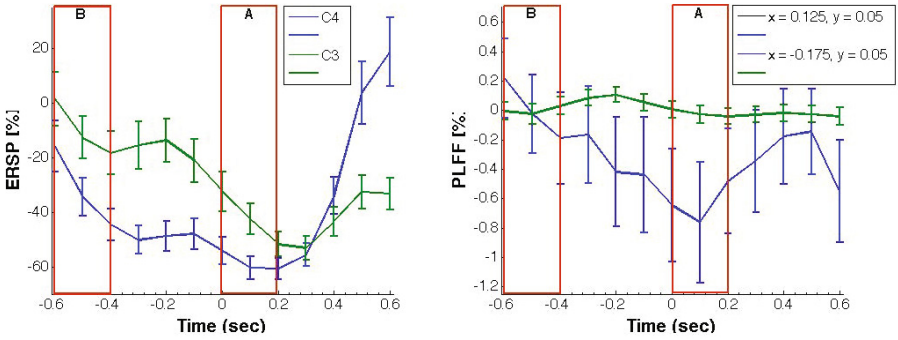

Fig. 3. ERD/ERS (%) time courses, for the self-paced movement, of: (*Left*) $ERSP$ in the β band in channels C4 (right hemisphere) and C3 (right hemisphere); and (*Right*) $PLFF$ in the indicated left and right hemisphere scalp positions. The task (A) and baseline (B) periods considered in the analysis are shown in red boxes.

3.2 Spatial Maps

The ERD/ERS maps obtained using the $ERSP$ in the β band and $PLFF$, for the self-paced and the cued movements, are shown in Fig.4 and Fig.5 , respectively. In the case of the cued movement experiment, a bilateral desynchronization was found with the $ERSP$ method while a contralateral desynchronization was found by $PLFF$ method. Despite this apparent inconsistency between the two methods, ipsilateral synchronization using $ERSP$ has in fact been previously reported and used to increase the power of the ERD/ERS detection [11].

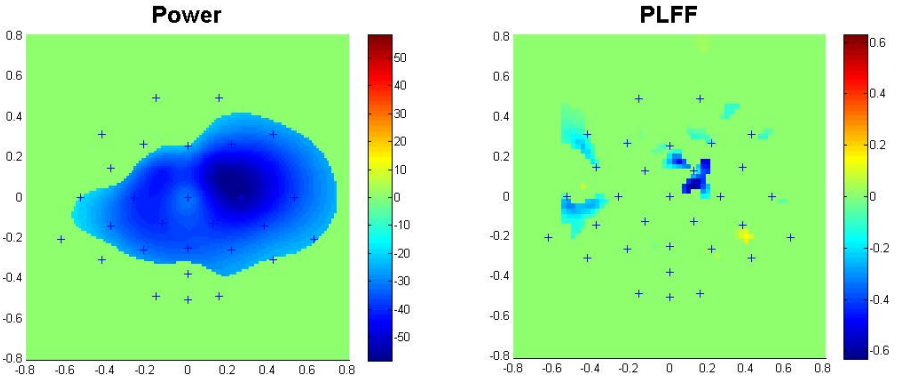

Fig. 4. ERD/ERS (%) maps, for the self-paced movement, of: (*Left*) $ERSP$ in the β band, interpolated; and (*Right*) $PLFF$. Only locations with statistically significant changes between task and baseline ($p < 0.01$) are shown. The (+) represent electrodes.

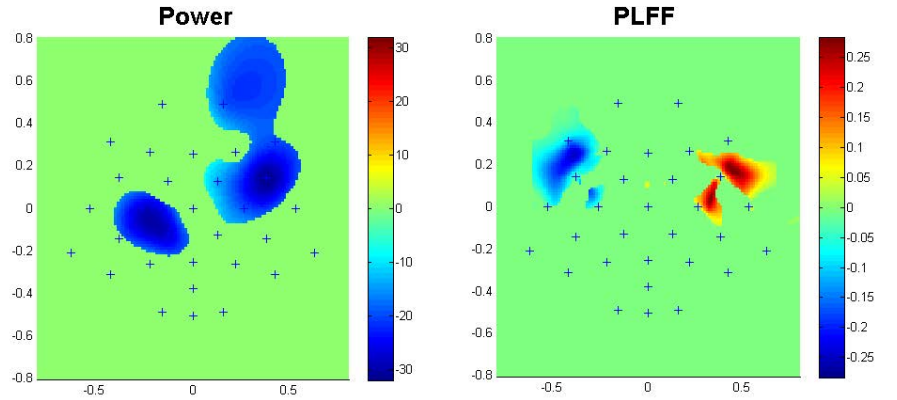

Fig. 5. ERD/ERS (%) maps, for the cued movement, of: (*Left*) $ERSP$ in the β band, interpolated; and (*Right*) $PLFF$. Only locations with statistically significant changes between task and baseline ($p < 0.01$) are shown. The (+) represent electrodes.

4 Discussion and Conclusions

This paper proposes a 2D continuous extension of the PLF metric for ERD/ERS quantification and brain activation detection from EEG data, which we call *Phase Locking Factor Field*. The method relies on the estimation of an interpolated 2D field of phases computed from each electrode by using the Hilbert transform in the analytical signals framework. Two motor tasks were used for illustrative purposes and comparison with the traditional method based on the EEG spectral power. Both methods were consistent with respect to the contralateral desynchronization but they did not agree on the ipsilateral synchronization. It is important to notice

that complete agreement is not necessarily observed because two different features are being computed; power and phase locking. More experiments are needed in order to validate the methodology.

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