

A METHOD FOR INVESTIGATING ALPHA BAND PHASE SYNCHRONIZATION IN THE CORTEX DURING A FATIGUING MUSCLE CONTRACTION USING EEG

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1. Abstract

Phase synchronization (PS) was calculated from EEG signals measured during an exercise-until fatigue trial, and a method to extract meaningful information from the data is shown. The purpose was to elucidate the brain's role in exercise fatigue, as there is evidence that indicates that the brain has a major role in regulating the work rate in exercise. EEG was measured during an exercise-until-fatigue trial as well as for a relaxed eyes-closed and eyes-open state, which served as baseline levels of phase synchronization. PS coefficients (which provide a measure of phase synchronization) between 2701 electrode pairs for 8 subjects were calculated for each condition at each epoch at the alpha (7-13Hz) frequency band. The method involved displaying the smallest PS and largest PS values of the differentiated PS coefficients on head maps. The head maps indicated that there was strong alpha band phase desynchronization followed by strong phase synchronization. The investigation showed that clear changes occur in the brain (specifically PS) and is the first known study that investigates PS in exercise and the progression to fatigue.

Keywords: EEG, phase synchronization, exercise fatigue, alpha

2. Introduction

The use of so called "high resolution" EEG has become popular in brain research. These EEG systems are able to sample brain wave activity at 200Hz using 64 electrodes or more. In studies that require recordings of more than a few seconds, these systems provide very large data sets; it may become very challenging for researchers to extract any meaningful information from these large data sets. In this paper we provide a method to extract patterns in the EEG recordings in exercise-fatigue experiments that may be of interest in the study of the brains role in exercise fatigue.

Traditionally researchers have seen the cause of fatigue during exercise as a result of limitations in physiological systems in the periphery of the body (outside the central nervous system) such as the

alterations of pH levels and inadequate ATP production in the cells [1][2]. There is growing evidence that the central nervous system (CNS) plays a major role in the onset of fatigue [1-4]. It is thought that the brain plays a part in regulating the work rate of the muscles during exercise fatigue [2]. The aim of this study was to determine if there were any measurable changes in brain wave activity in the progression of exercise fatigue, normalized to exercise trial length, which could lend support to this hypothesis.

The trial involved the sub-maximal contraction of the quadriceps muscle until fatigue was reached and the particular variable of interest was *phase synchronization*. Phase synchronization occurs when rhythmic or oscillatory systems (such as the brain regions), through some kind of interaction, adjust their behaviours relative to one another so as to attain a state where they work in unison [5]. Phase synchronization was thus investigated to determine which regions of the brain display coherent, dependent behaviour during a fatiguing muscle contraction. This is evaluated by means of the high-resolution electroencephalogram (EEG). Data collected from each EEG electrode represents the oscillatory electrical activity of the brain areas underlying it. Thus phase synchronization between two EEG channels is indicative of communication between the underlying structures or of a functional relationship between the underlying structures [7]. In this study phase synchronization was quantified between every EEG electrode pair.

EEG has high temporal resolution making it an invaluable tool for measuring phase synchronization. Changes in phase synchronization values can be calculated at time scales of milliseconds, which correlate to the same time frames that these changes are occurring in the brain [6]. Since it typically takes hundreds of seconds to reach fatigue, there is a large amount of data produced from phase synchronization calculations at millisecond time scales and this can make it difficult to formulate meaningful interpretations from the data. Furthermore, data sets are enlarged by the use of so called high-resolution EEG (74 channels in this study, which results in 2701 electrode combinations). The method presented here is aimed at showing the data in a manner that allows for comparisons with controls and is not necessarily an exhaustive analysis. Since the brain activity during the fatiguing process is highly complex and extends over a relatively long period, a highly detailed analysis may not be feasible; a method that was conducive to a more superficial description of the process was selected. No known studies to date have investigated phase synchronization in the cortex during exercise fatigue.

3. Methods

Subject selection, the trial and EEG measurement were all performed by Harley, YXR and colleagues [8].

Subjects

Twenty-five healthy subjects participated in the exercise-until-fatigue trial with the subjects having a range of physical fitness levels. After close investigation, only data from eight of the subjects (5 male and 3 female) was determined to be suitable for analysis due to data artifacts. These remaining subjects had a mean age of 28 ± 4.7 years.

Task

The trial involved a seated subject performing an isometric contraction (a contraction where the muscle length does not change) of the right quadriceps muscle, by extending the right knee until they were unable to continue to do so i.e. until they reached fatigue (*the exercise condition*). The EEG of each subject was recorded during the trial. Before the commencement of the trial the EEG recordings were taken of the subjects in both a relaxed eyes closed state (*eyes closed condition*) and a relaxed eyes open state (*eyes open condition*). These recordings lasted approximately 2 minutes each per subject and serve as a baseline for the analysis of the EEG data collected during the trial.

EEG Measurement

A dense array EEG system, the Geodesic System 200 [9] was used for EEG recordings. The system has 128 recording channels and a reference channel, allowing for high spatial sampling of scalp EEG recordings. The EEG signals were recorded at a sampling frequency 200Hz, with the electrode at the vertex of the head used as a reference. The data was filtered online with a bandpass filter with frequency cut-offs at 0.1 and 70Hz and was saved in a Matlab [10] file format for further offline processing.

EEG Data Pre-processing

The EEG data was spatially enhanced using *spherical spline interpolation* as developed by Perrin et al and outlined in [11][12][13]. A digital implementation of a sixth order Butterworth filter was chosen to filter the data in the alpha band (8-13 Hz).

Since it took each subject a different length of time to reach fatigue, the data was normalized so that it could be averaged. This was achieved by dividing each subject's data into 6 equal time periods. This resulted in 6 windows of fixed length for each subject with these windows having varying lengths across subjects. A coefficient of phase synchronization was then calculated using the windows that corresponded in time and the coefficients were then averaged and further analyzed as described in the following section.

Quantifying phase synchronization

The method is based on the work of Tass and colleagues [14] and Rosenblum and colleagues [15] and shall be briefly described here. Synchronization of noisy or chaotic systems such as the EEG can be understood by observing preferred values of the generalized phase difference of the signals by looking at the distribution of the *cyclic relative phase* defined as

$$\Psi_{n,m} = \varphi_{n,m} \bmod 2\pi$$

where

$$\varphi_{n,m}(t) = n\phi_1(t) - m\phi_2(t), \quad (3-1)$$

n and m are some integers and will be restricted to $n = m = 1$ and $\phi_{1,2}$ are the phases of the two oscillators. Both $\varphi_{n,m}$ and $\phi_{1,2}$ are defined on the whole real line and not just on the circle $[-\pi, \pi]$.

The distribution of $\Psi_{n,m}$ can be statistically quantified and provides a useful 'measure' of phase synchronization. Strong noise (whether external noise or deterministic noise) can introduce rapid jumps in the relative phases of the oscillators and $\varphi_{n,m}(t)$ will exhibit a random-walk-like motion [14]. Phase locking will be determined if there are preferred values for $\Psi_{n,m}$ or peaks in the distribution of $\Psi_{n,m}$.

The first step in calculating $\Psi_{n,m}$ was to determine the phases of the oscillators. This was achieved using the analytic signal $\psi(t)$ which is a complex function of time where the instantaneous amplitude $A(t)$ and phase $\phi(t)$ of a signal $s(t)$ can be determined. It is defined as

$$\psi(t) = s(t) + j\tilde{s}(t) = A(t)e^{j\phi(t)} \quad (3-2)$$

where $\tilde{s}(t)$ is the Hilbert transform of $s(t)$

$$\tilde{s}(t) = \pi^{-1} PV \int_{-\infty}^{\infty} \frac{s(\tau)}{t - \tau} d\tau \quad (3-3)$$

where PV means that the integral is taken in the sense of the Cauchy principal value.

To quantify the deviation of the relative cyclic phase $\Psi_{1,1}$ from a uniform distribution the method developed by Tass et al [14], based on Shannon entropy was applied to the EEG data. A phase synchronization coefficient

$$\tilde{\rho}_{n,m} = (S_{\max} - S) / S_{\max} \quad (3-4)$$

is defined where

$$S = - \sum_{k=1}^N p_k \ln p_k \quad (3-5)$$

is the entropy of the distribution of $\Psi_{n,m}$,

$$S_{\max} = \ln N \quad (3-6)$$

and N is the number of bins in the distribution. This results in a normalized value representing the degree of phase synchronization, where $\tilde{\rho}_{n,m} = 0$ corresponds to a flat distribution of $\Psi_{n,m}$ and no phase synchronization and $\tilde{\rho}_{n,m} = 1$ corresponds to a sharp peak in the distribution of $\Psi_{n,m}$ and perfect phase synchronization.

Calculation of Phase Synchronization Coefficients

Alpha band phase synchronization (ABPS) was quantified between 2 EEG signals by calculating the phase synchronization coefficient, $\tilde{\rho}$ as described above. Phase synchronization coefficients were calculated to determine ABPS between all electrode sites in a particular condition (exercise-fatigue, eyes-closed and eyes-open) and subject. This was done on a per epoch basis (6 epochs per channel for each condition and subject). Since there were 74 channels used the number of channel combinations pairs was $\binom{74}{2} = 2701$.

Once the coefficients were calculated a method was developed to compare values of $\tilde{\rho}$ between conditions, epochs and electrode locations. This would provide a basic description of ABPS in the brain in exercise-until-fatigue versus the baseline eyes closed and eyes open conditions. In the literature coherence values are often plotted on head maps. Head maps, which are 2-dimensional representations of the electrode locations on the scalp, provide a way to view the phase synchronization between two electrode locations in two-dimensions, and a series of head maps may indicate how ABPS between two electrodes changes in time.

In order to highlight changes in ABPS between epochs, the value of $\tilde{\rho}$ of each epoch was subtracted from the value of $\tilde{\rho}$ of each successive epoch resulting in 5 values of differentiated $\tilde{\rho}$, denoted $\Delta\tilde{\rho}$ for the 6 epochs or 5 epoch transitions. This was done for each channel combination, condition and subject. The aim

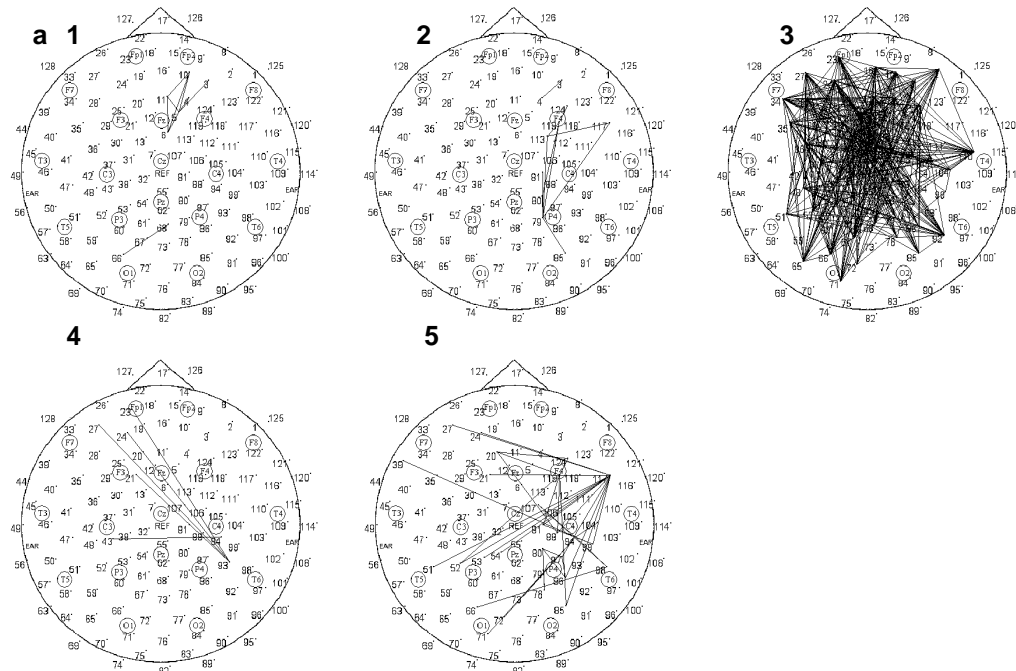
of the process was to see which particular channel combinations displayed the greatest increases or decreases in ABPS and whether these increases or decreases were stronger for a particular condition.

The $\Delta\tilde{\rho}$ values were averaged over subjects and all the subject-averaged $\Delta\tilde{\rho}$ values were pooled together and ranked. The highest 0.5% and the lowest 0.5% of the ranked values were retained and plotted on head maps. The highest 0.5% represents phase synchronization (all these values were positive) and the lowest 0.5% represents phase desynchronization (all these values were negative).

The retained subject-averaged $\Delta\tilde{\rho}$ values corresponded to a particular condition, channel combination and epoch transition. Each retained subject-averaged $\Delta\tilde{\rho}$ value was plotted on a head map that indicated its position on the map (a line drawn between 2 channel locations representing a channel combination). There were 5 head maps representing the time course of $\Delta\tilde{\rho}$ for each condition, forming a single graphical unit referred to as a head map series. Each retained $\Delta\tilde{\rho}$ value was plotted on a head map that corresponded to its correct epoch transition number and condition. The process resulted in 6 head map series, one synchronization map and one desynchronization head map series per condition.

4. Results and discussion

The set of head maps presented show alpha band phase synchronization (ABPS) and alpha band phase desynchronization (ABPD) for the exercise-until-fatigue conditions. The results for the eyes-open and eyes-closed conditions are not shown but show no remarkable patterns.



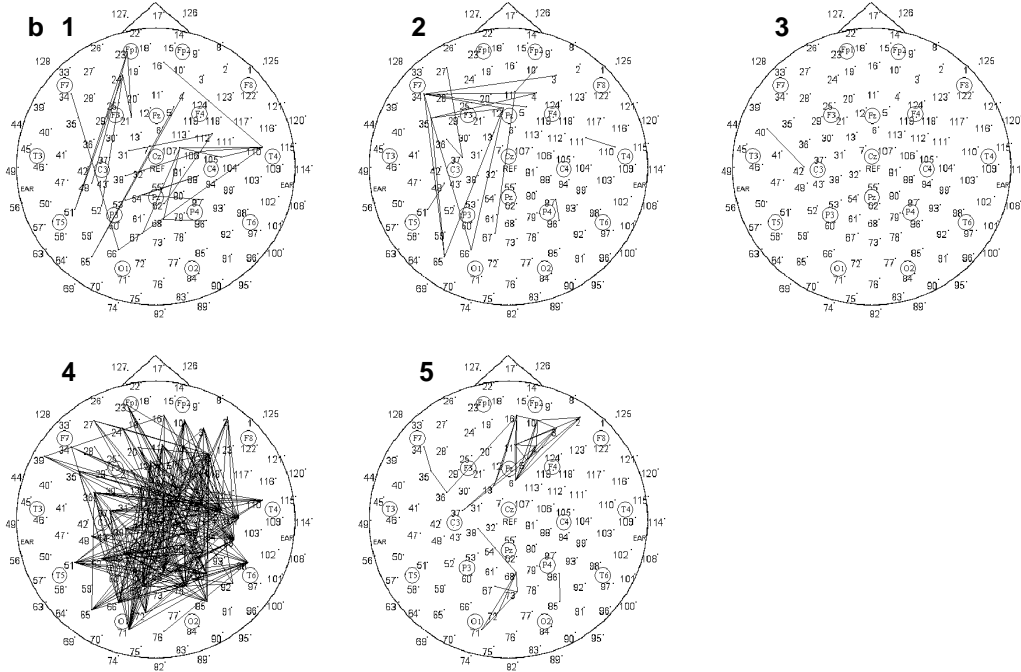


Figure 1: Alpha phase desynchronization for the exercise-until-fatigue condition (a) and alpha phase synchronization for the same condition (b) for each epoch transition 1-5 (indicated on each head map). Note that a line indicates a $\Delta\tilde{\rho}_\alpha$ value above (synchronization case) or below (desynchronization case) a certain threshold.

The maps provided a way of observing where the greatest amount of ABPS and ABPD is occurring temporally and spatially and for what condition. What is particularly noteworthy is the large scale ABPD pattern at epoch transition 3 (Figure 1 a no. 3) followed by a large scale ABPS pattern at epoch transition 4 (Figure 1 b no. 4) in the exercise-until-fatigue condition. This pattern is absent from all epochs in the other conditions. The ABPD pattern observed in Figure 1 a (no. 3) shows the participation of many electrode combination pairs across the entire cortex. The ABPS pattern that follows (Figure 1 b no. 4) seems to recruit less electrode pairs although it is still widespread across the cortex.

One should be cautious when trying to make interpretations from the head maps shown. The activity represented by the maps is very complex, representing the activity of billions of interconnected neurons whose electrical characteristics are changing in time frames of milliseconds. The head maps are a very coarse representation of these processes. Yet the method provides a way to show that there are clear changes that occur in the brain (ABPS and ABPD) in exercise and the progression to fatigue compared to the baseline eyes-open and eyes-closed states which deserve further investigation.

Since this study involved both motor and cognitive brain processes, a methodology that would attempt to discern the two processes should be investigated. As a starting point, since the exercise was performed with a one sided leg contraction, the differences in hemispheres should be investigated, in a similar way to how the different areas was determined. Since movement creates specific changes in ipsi-lateral and contra-lateral brain areas [16][17], these could have been used to help differentiate between motor and cognitive effects. Other methods could be devised to help understand the endogenous and sensory processing effects during exercise, such as the concomitant use of Rate of Perceived Exertion (RPE), heart rates, EMG and body temperature measurements for example.

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