DETECTION OF HIGH-FREQUENCY STEADY STATE VISUAL EVOKED POTENTIALS USING PHASE RECTIFIED RECONSTRUCTION

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ABSTRACT

Brain-computer interfaces (BCIs) based on the steady state visual evoked potential (SSVEP) can provide higher bitrates and require shorter training than other BCIs. The oscillating visual stimulus required to elicit an SSVEP can be (depending on the stimulation frequency) annoying for the user and even dangerous since it can trigger photoepileptic seizures. These issues can be surmounted by using high-frequency stimulation (beyond 40 Hz). However, the nature of the SSVEP is such that it decreases with increased stimulation frequency. Conventional spectral methods are not always suitable for detecting high-frequency SSVEPs. In this paper, it is shown that the recently developed phase rectified reconstruction average (PRSA) technique effectively detects SSVEPs elicited by stimulation frequencies between 43 and 46 Hz.

1. INTRODUCTION

In addition to its clinical application for diagnosing impairments in the visual-nerve pathway [1], the steady state visual evoked potential (SSVEP) can serve as a basis for electroencephalogram (EEG) based brain computer interfaces (BCI) [2, 3, 4] [1].

Compared to other types of BCIs, SSVEP based BCIs provide higher information transfer rates (*bitrate*) with minimal user training, and require fewer EEG channels [5]. For these reasons, this type of BCI seems to be the most promising for deployment in the consumer's market in the near future.

The SSVEP is the oscillatory wave appearing in the occipital leads of the EEG in response to a visual stimulus modulated at a certain frequency. Examples of visual stimuli are: reversal of checkerboard patterns, and flickering lights [6]. In this paper we focus on stimulation based on flickering lights.

The frequency of the SSVEP matches the frequency of the stimulus or its harmonics. SSVEP based BCIs operate by simultaneously presenting to the user visual stimuli modulated at different frequencies. Each stimulus is associated with an action in an output device. The spectral content of the user's EEG is continuously computed for time-windows whose duration depends on the BCI configuration (typical durations are in the order of seconds [7]). When the user focuses his/her attention on a certain stimulus, the corresponding stimulating frequency and/or its harmonics dominantly appear in the spectral representation of the EEG. Thus, a *dominant SSVEP* is visible in the EEG. This is particularly evident in the signals recorded at occipital sites, electrodes O1, O2, Oz of the 10/20 EEG international system [8] (see Figure 1). If the power associated to the dominant SSVEP (*SSVEP-power*) is above a threshold (*action threshold*) the corresponding action is executed. Otherwise the system does not react. The action threshold is experimentally adjusted for each stimulation frequency and is subject dependent [9].

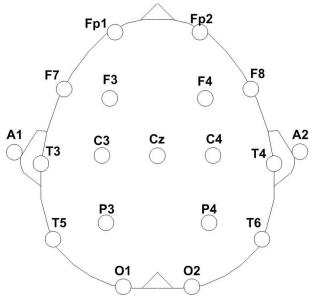

Fig. 1. EEG electrodes placement according to the 10-20 international system [8].

The choice of the stimulation frequencies in a BCI application must ensure that the corresponding SSVEPs are readily detectable. This implies that the stimulating frequencies are neither harmonics nor sub-harmonics from each other.

The SSVEP-power depends on the stimulation frequency. When flashing lights are used for stimulation, the highest SSVEP-power occurs for a stimulation frequency of about 15 Hz [9]. The relation between the SSVEP-power and the stimulation frequency is subject dependent. A detailed discussion on the subject variability of the SSVEP-power can be found in [9].

Because of relative straightforward SSVEP detection, the stimulation frequencies are usually chosen in the 5-30 Hz band. In spite of its favorable detection properties, this band presents two major inconveniences, namely:

- stimulation frequencies in this band are rather annoying for the subject.
- the risk for inducing photoepileptic seizures is higher for stimulation frequencies in the 15-25 Hz band [10].

To be usable, SSVEP BCIs must overcome these inconveniences. A simple solution consists in applying higher stimulation frequencies (beyond 40 Hz). Detecting SSVEPs elicited by high stimulation frequencies using conventional spectral methods (e.g. Fourier transform to estimate the power spectral density) is challenging because of the poor signal-to-noise ratio and the non-stationary nature of EEG. In this paper we propose to apply the recently developed Phase-Rectified Signal Averaging (PRSA [11]) to detect SSVEPs

[1] In this paper, the term BCI is henceforth used to refer to an EEG based BCI.

elicited by high stimulation frequencies.

This paper is organized as follows. In Section 2 the basic PRSA concepts are presented. Section 3 summarizes the experimental protocol and Section 4 discusses the results. The conclusions and prospective future research sujects are presented in Section 5.

2. PHASE-RECTIFIED SIGNAL AVERAGING (PRSA)

The application of PRSA to physiological signals (particularly ECG) was introduced in [11]. Here, only the main concepts in [11] are summarized. PRSA is capable of detecting and quantifying quasi-periodic oscillations masked by the non-stationary nature of composite signals and noise.

A signal $(x_i), i = 1,\ldots,N$ is considered. In addition to relevant periodicities and correlations, x_i may exhibit non-stationarities, artefacts and noise. EEG signals containing SSVEPs constitute examples of such signals. PRSA aims at compressing the signal into a much shorter sequence, keeping all relevant quasi-periodicities but eliminating non-stationarities, artefacts, and noise.

The basic PRSA principle is the aligning of sections of the signal x_i relative to selected *anchor points* (i.e. some of the samples x_i) followed by a signal averaging. The method consists of three steps:

In the first step, the anchor points are selected according to certain properties of the signal. Different selection criteria are discussed in [11]. Here, x_i is considered as an anchor point if:

$$\frac{1}{T}\sum_{j=0}^{T-1} x_{i+j} > \frac{1}{T}\sum_{j=1}^{T} x_{i-j}, \tag{1}$$

where T is a parameter whose value is set depending on the frequency of the periodicity that has to be detected. The anchor points are thus defined by comparing averages of T consecutive samples of the signal x.

The PRSA is most sensitive for strictly periodic oscillations with frequency $f \approx 1/(2.7T)$. Indeed, the sums in (1) can be approximated by the integral:

$$\frac{1}{T}\int_0^T \sin(2\pi f x)dx = \frac{1-\cos(2\pi fT)}{2\pi fT}, \tag{2}$$

finding the maximum of this expression yields $T \approx 1/(2.7f)$.

In the second step of the PRSA method, windows of length $2L$ (which we call PRSA-windows) are defined around each anchor point. Anchor points close to the beginning or the end of the signal, where no full surroundings of length $2L$ are available, are disregarded. If the indices of all regarded anchor points are denoted as $i_\nu, \nu = 1,\ldots,M$, the points in window number ν, corresponding to the anchor point x_{i_ν} are:

$$x_{i_\nu-L}, x_{i_\nu-L+1}, \ldots, x_{i_\nu}, \ldots, x_{i_\nu+L-2}, x_{i_\nu+L-1}. \tag{3}$$

Several of the PRSA-windows will overlap, since many anchor points are close to each other. The parameter L should exceed the expected coherence time of the periodicities in the data and it must definitely be larger than the period of the slowest oscillation that one wants to detect. In our experiments, we found that L is about one order of magnitude larger than the slowest relevant periodicity in the signal.

In the third step, the PRSA $\tilde{x}(k)$ is obtained by averaging over all PRSA-windows,

$$\tilde{x}(k) = \frac{1}{M}\sum_{\nu=1}^{M} x_{i-\nu+k} \;\; k = -L, -L+1, \ldots, 0, \ldots, L-2, L-1 \tag{4}$$

In this average, non-periodic components that are not phase synchronized with the anchor points, i.e. non-stationarities, artefacts, and noise, cancel out, and only events that have a fixed phase relationship with the anchor points, i.e. all periodicities and quasi-periodicities, "survive" the procedure.

3. EXPERIMENTAL PROTOCOL

Two male subjects S1, S2 aged 23 and 25 respectively participated in four recording sessions where the SSVEP was measured for stimulating frequencies in the range 40 to 49 Hz.

The EEG was acquired using a BIOSEMI [12] system at a sampling rate of 2048 Hz . In this study, the bipolar signal between Oz and Cz was used to detect the SSVEP. This choice was motivated by the possibility of integrating electrodes into a consumer headphone device which would record signals at Oz and Cz. The signal was band-pass filtered in the 0.5-100 Hz band and subsequently resampled at 256 Hz.

In each of the four sessions, subjects were presented with stimuli at frequencies $f_n = 40 + n$, $n = 0,\ldots,9$ Hz. Ten-second long periods of stimulation at a randomly chosen frequency among the f_n were followed by break periods of about 15 seconds. Each frequency was presented six times. Thus, the total dataset per subject consisted of (6×4) ten-second long *trials* per stimulation frequency.

4. RESULTS

For each trial, the periodogram (using Welch's method [13]) and the power spectrum density (PSD) of the PRSA were estimated.

Figure 2a depicts the periodogram of a trial recorded during flicker stimulation at 46 Hz for subject S1. The corresponding PRSA signal in Figure 2b exhibits the oscillation at 46 Hz which appears clearly in the PSD of Figure 2c. In the PRSA signal, the sample index 0 correspond to the center of the PRSA-windows. As in [11], the peak at 46 Hz in the PRSA's PSD is sharper than in the periodogram.

To characterize the extent to which the SSVEP corresponding to the stimulation frequency can be detected in each trial, the ratio (which we call *detectability ratio*) between the power at the stimulation frequency and the highest power in the 40-128 Hz band was computed for both the periodogram and the PSD of the PRSA. When the ratio was equal to one, i.e. the highest power coincided with the SSVEP power, the comparison was made with respect to the second highest power. Thus, the higher the detectability ratio is, the easier to detect the SSVEP.

Figure 3 depicts the average detection ratio (over all trials corresponding to a given stimulation frequency) versus the stimulation frequency for both, the PSD of the PRSA signal and the periodogram. Figure 3a(b) shows the detection ratio for subject S1(S2).

For subject S1, the PRSA is significantly superior in detecting the SSVEP for stimulation frequencies between 43 and 46 Hz. Outside this band, the detection ratio is below 1 which indicates that other peaks mask the SSVEP. Nonetheless, in general the PRSA appears to perform better than the periodogram for SSVEP detection.

For subject S2, the PRSA can effectively detect SSVEPs elicited by stimulation frequencies between 44 and 46 Hz. Outside this band, the detection rate falls below 1. The detection ratio associated with the periodogram does not exceed 1 for any stimulation frequency that was tested.

The shape of the detection ratio curve in function of the stimulation frequency is similar for both subjects. Indeed, they both indicate that the SSVEP is more prominent in a narrow band in the vicinity of 45 Hz. This could be explained by the fact that 45 Hz corresponds to the third harmonic of 15 Hz which, according to [9], is the stimulation frequency at which the SSVEP is maximum. Further experiments involving stimulation frequencies around 30 Hz (the second 15 Hz harmonic) need to be performed in order to test this hypothesis.

5. CONCLUSIONS AND FUTURE WORK

To be suitable for consumer applications, BCIs based on SSVEPs need to address two major issues, namely the annoyance caused by flickering stimuli and the risk of inducing photoepileptic seizures. Such issues can be surmounted by using high frequency (beyond 40 Hz) stimulation frequencies.

Detection of SSVEPs at high stimulation frequencies is challenging using transitional spectral methods because the amplitude of the SSVEP significantly decreases in the frequency band beyond 40 Hz. In this paper, the PRSA technique has been applied for high-frequency SSVEP detection. The results show that the PRSA performs better than the conventional periodogram. Yet, effective detection of SSVEPs occurs for a narrow band in the vicinity of 45 Hz.

In spite of the narrow band in which the PRSA is effective, the results are very promising for developing a practical SSVEP based BCI using high frequency stimulation. Indeed, several stimulation frequencies in this band can be used for stimulation by improving the frequency resolution. This has certainly need to be traded with longer reaction times since a higher frequency resolution entails lower time resolution.

The future steps in this research include:

- Optimization of the recording sites. Which can be subject dependent. Using several electrodes can improve the detectability of the SSVEP by applying some form of spatial (across the electrodes) filtering and also allow for the design of a universal (non-subject dependent) electrode positioning device which would only need to update the spatial filter coefficients to account for subject variability.
- Using more electrodes to increase the SSVEP detectability. To this end we plan to modify the PRSA technique to include multiple channels.

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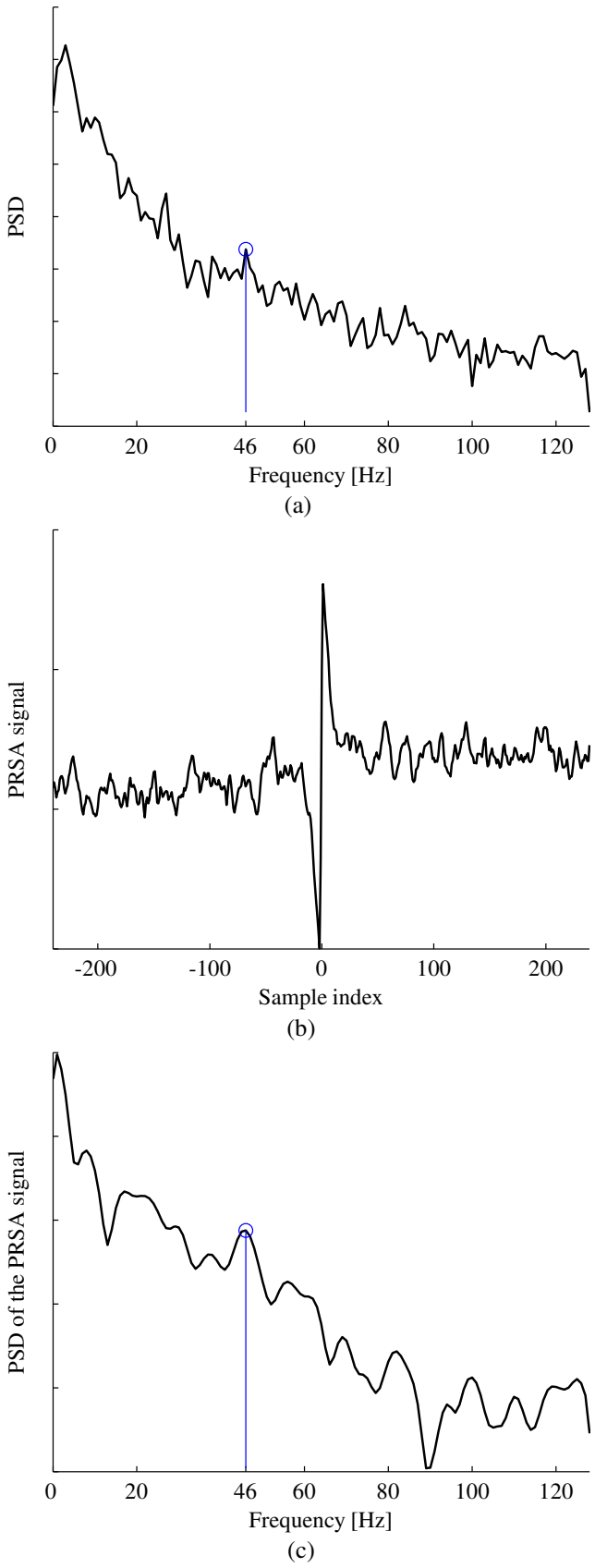
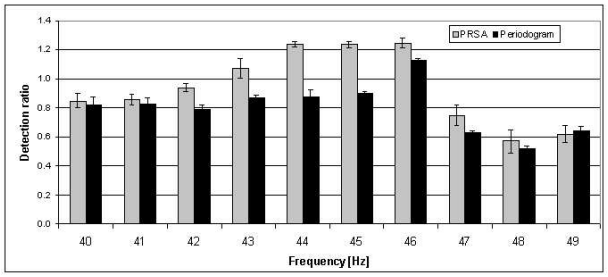
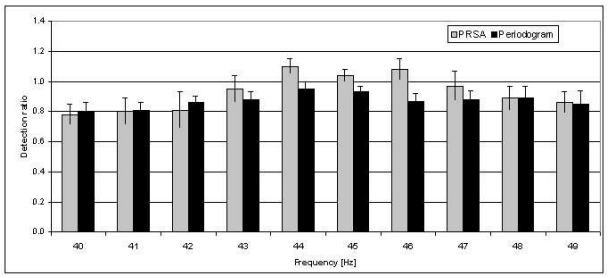


Fig. 2. Periodogram (a), PRSA signal (b), and PSD of the PRSA signal (c) of a trial recorded during 46 Hz stimulation



(a)



(b)

Fig. 3. Detection ratio computed on both, the PSD of the PRSA and the periodogram for subject S1 (a) and subject S2 (b).

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