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D3.4 BRAIN – Advanced SSVEP signal processing tools

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<tr>
<td>BCI</td>
<td>Brain Computer Interface</td>
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<tr>
<td>EEG</td>
<td>Electroencephalogram</td>
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<td>SSVEP</td>
<td>Steady State Visual Evoked Potential</td>
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1. Summary

Brain-computer interfaces (BCI) based on Steady State Visual Evoked Potential (SSVEP) can provide higher information transfer rate than other BCI modalities. For the sake of safety and comfort, the frequency of the repetitive visual stimulus (RVS) necessary to elicit an SSVEP should be higher than 30 Hz. However, in the frequency range above 30 Hz, only a limited number of frequencies can elicit sufficiently strong SSVEPs for BCI purposes. Consequently, the conventional approach, consisting in presenting various repetitive visual stimuli having different frequency each, is not practical for SSVEP based BCI functioning. Indeed this would bring low communication bitrates. In order to increase the number of possible repetitive visual stimuli, we consider modulating the phase of the stimulus instead of the frequency. Thus, several stimuli sharing the same frequency, but with different phase can be presented to the user. The approach presented in this document, to detect the phase of the stimulus instead of the frequency. This consists in using as feature, the phase difference between the SSVEP and the stimulus. The phase is extracted through the Hilbert transform applied on an univariate signal resulting from spatially filtering the electroencephalogram. The spatial filter is determined in such a way that the SSVEP energy is enhanced through a linear combination of the signals recorded at different positions on the scalp.

2. Milestone 3.3 Second prototype of advanced SSVEP signal processing tools

1.1. Summary of task 3.1 Advanced SSVEP signal processing tools

BCI systems based on SSVEPs have many appealing features, but many subjects find the required flickering stimuli annoying. BCIs can be made less annoying by using high frequency stimulation patterns (beyond 30 Hz) or convenient frequency modulation (e.g. the stimulating signal can be a square wave at a certain frequency in which the duty cycle randomly changes between successive cycles). However, for higher frequencies and the randomized duty cycle, the amplitude of the SSVEP reduces considerably. Philips has already developed advanced self-adaptive SSVEP signal processing software that provides excellent signal quality at these frequencies. These tools will be adapted to work with BRAIN. In addition, we will leverage on the Philips expertise in lighting devices to develop a portable LED based SSVEP stimulator supporting different stimulation strategies that are comfortable and safe for the user. The three phases of this task include initial, advanced, and final system which will be ready after 6, 18, and 30 months respectively. They will also be tested with other BRAIN components across target users through work package 5.

1.2. Milestone 3.3 (18 months)

SSVEP elicited by high frequency repetitive visual stimulation is suitable for BCI operation. However, in the high frequency range only a small number of frequencies can be used to elicit SSVEPs of sufficient strength for BCI purposes. Consequently, we have decided to change not only the frequency but also the phase of the stimulus to overcome the frequency limitation. Thus, we have developed signal processing tools to detect the phase of the stimulus. Such tools are reported in this document along with experimental results to support their effectiveness. Such signal processing techniques constitute the core of milestone 3.3.

3. Introduction

The steady state visual evoked potential (SSVEP) refers to the response of the cerebral cortex to a repetitive visual stimulus (RVS) oscillating at a constant frequency and is characterized by peaks at the fundamental frequency and its harmonics in the power spectral density (PSD) of EEG signals. The SSVEP is an effective electrophysiological source that can be used as an input of a brain computer interface (BCI).

Previous studies have demonstrated that it is possible to detect the user’s focus of attention among several visual stimuli. This can be achieved by examining the amplitudes at the stimulus frequency (and harmonics) in the spectral content of the EEG [6, 9, 4].

Most SSVEP-based BCIs use stimulation frequencies in the 4 to 30 Hz range. The SSVEPs elicited by frequencies in this range have higher amplitude but can induce visual fatigue or
epileptic seizures [3]. Higher stimulation frequencies are preferable for safety and comfort. Yet, only a limited number of frequencies above 30 Hz can elicit a sufficiently strong SSVEP for BCI purposes. Thus, if a frequency per target is used, the number of choices (and consequently the information transfer rate) in a BCI is limited.

A possible way to tackle such limitation consists in combining several frequencies to drive a single visual stimulus [2, 12]. Thus, if N frequencies are used, a target may combine k frequencies selected among the N available. From combinatorial theory it is known that \( \binom{N}{k} > N \) if \( N > k + 1 \) and \( k > 1 \). An alternative way consists in using the same frequency for several stimuli but different phase [18, 8]. Detecting the phase of the stimulus that receives the user’s focus of attention is possible because the SSVEP is phase-locked with the stimulus.

The SSVEP phase can be obtained using methods based on the Discrete Fourier Transform (DFT) [18, 19, 8]. These methods however require relatively long signal segments containing a number of samples that is a multiple of the stimulus period. This effectively increases the latency period and therefore reduces the communication throughput.

In this study, the phase was identified using the Hilbert transform. To align the stimulus signal with the SSVEP, the oscillatory light emanating from one of the stimuli (without loss of generality, the stimulus with 0-phase was selected) was simultaneously recorded. In the following we refer to such a signal as stimulation-signal. The difference between the instantaneous phase of SSVEP and the stimulation-signal was calculated. Such phase difference is dependent solely on the subject and the frequency. Because the subtraction of the two phases from the SSVEP and the stimulus signal cancels the time variability, the synchronization of presenting stimuli and recording EEG signals between sessions is not necessary. An additional advantage of using the phase difference as a feature is that it can avoid the change in SSVEP phase distribution caused by the possible phase deviation of the stimuli.

Another aspect considered in this study is the comparison between the use of the bipolar signal Oz-Cz and a combination of several electrodes (through spatial filtering), to detect the SSVEP phase.

4. Methods

In this section, we summarize the method for spatial filtering based on [7, 5], and present the phase synchrony analysis based on the Hilbert transform.

1.3. SSVEP enhancement using spatial filtering

A signal recorded at a particular electrode location, that contains \( T \) samples can be seen as a vector in the space \( \mathbb{R}^T \). Following this interpretation, we use hereafter the terms vector and signal without explicit distinction.

The signal \( x_i \), where \( i \) indexes the electrode location, recorded while the subject focuses his/her attention on a RVS modulated at a certain frequency \( f \), can be written as the sum of the SSVEP component (denoted as \( s \)), background EEG and noise [5]. For convenience, the background EEG and the noise at electrode \( i \) are combined into a single term denoted by \( y_i \). Thus, the following relation holds:

\[
x_i = s_i + y_i,
\]

\[
x_i = \sum_{h=1}^{H} a_{h,i} \sin 2\pi h f t + b_{h,i} \cos 2\pi h f t + y_i,
\]

where the SSVEP component is modeled as a linear combination of vectors in the set:

\[
\Phi = \begin{bmatrix} \sin 2\pi h f t, \cos 2\pi h f t \end{bmatrix} \quad h = 1, ..., H
\]

Here \( t = 0, ..., T - 1 \) is a vector of sample indices, \( H \) is the number of harmonics that are considered in the model, and \( a_{h,i} \) and \( b_{h,i} \) are real numbers. Equation 1 can be generalized to the whole set of electrodes \( x_i | i = 1, ..., N \) (\( N \) is the number of electrodes) in the following matrix form:
where the matrix $X \in \mathbb{R}^{T \times N}$ (the notation $\mathbb{R}^{T \times N}$ refers to the space of real matrices having $T$ rows and $N$ columns) has as columns the vectors $x_i$, $Y \in \mathbb{R}^{T \times N}$ has as columns the vectors $y_i$, $S \in \mathbb{R}^{T \times 2H}$ has as columns the vectors in the set $\Phi$, and $A \in \mathbb{R}^{2H \times N}$ is the matrix of linear combination coefficients such that: $A_{h,j} = a_{h,j}$ if $h$ is odd and $A_{h,j} = b_{h,j}$ if $h$ is even. By means of the coefficients $a_{h,j}$ and $b_{h,j}$ the model in Equation 3 takes into account the differences of SSVEP-strengths across the scalp.

### 1.3.1. Construction of the spatial filter

The elements of $A$ in Equation 3 cannot be determined from $X$ and $S$ only. Thus, to determine the SSVEP strength at different electrode locations, a signal $x_w$ is constructed such that: $x_w = \sum w_i x_i = Xw$, where $w = w_1, \ldots, w_N$. The signal $x_w$ can be considered to be the result of a spatial filter (e.g., filtering across the various electrodes) with coefficients $w_i$ applied to the measured signals $x_i$. The spatial filter is determined in such a way that it simultaneously maximizes the energy in the SSVEP frequencies and minimizes the energy in the nuisance signals [5]. This ratio can be determined by relying on the geometric interpretation described as follows.

The linearly independent vectors in the set $\Phi$ generate a vector-space $\Pi$ in $\mathbb{R}^T$ of dimension $2H$. In this article's framework, we assume $\Pi$ to be the space where the SSVEP components lie. Thus, $\Pi$'s orthogonal complement $\Pi^\perp$ contains the non-SSVEP components. Since the vectors in $\Pi$ are linearly independent, the projection matrix $Q$ on $\Pi$ can be written as $Q = S S^\top S^{-1} S^\top$ [11]. The component of $Xw$ in $\Pi^\perp$ is equal to $Xw - QXw$. The Euclidean norm of the latter: $\|Xw - QXw\|^2$ divided by $T$ represents the power of the non-SSVEP related activity.

The power of the SSVEP related activity in $Xw$ can be approximated by $w'X'Xw$. The spatial filter $w$ corresponds to the argument that maximizes the ratio $\rho = \frac{w'X'Xw}{\|X - QXw\|^2}$.

$$w = \arg \max \frac{w'X'Xw}{\|X - QXw\|^2}.$$  

The ratio in Equation 4 is a generalized Rayleigh quotient [15], whose maximum can be found through a generalized eigendecomposition of the matrices $X'X$ and $X - QX$. This results in two matrices $W, \Lambda \in \mathbb{R}^{N \times N}$, such that

$$X'XW = X - QX \quad X - QX \quad W_{\Lambda},$$

where $\Lambda$ is a diagonal matrix whose diagonal contains the eigenvalues. The corresponding eigenvectors are in the columns of $W$. By construction, eigenvalues are larger than one [17]. The largest element in $\Lambda$ corresponds to the maximum of the quotient in Equation 4. The column of $W$ corresponding to such maximum is the sought spatial filter $w$.

In this study, we used the training strategy detailed in [7] to obtain $w$. Using such a filter we obtained the univariate signal $x_w = \sum w_i x_i = Xw$ on which the phase synchrony analysis is
performed. For convenience of presentation of the phase synchrony analysis we use the notation \( x_w(t) \) to refer to \( x_w \).

### 1.3.2. Phase synchrony analysis

The phase difference between the SSVEP and the stimulation-signal was estimated for two univariate signals, namely the bipolar signal \( x_b(t) \) resulting from subtracting the signal recorded at electrode Oz from that recorded at electrode Cz, and the spatially filtered signal \( x_w(t) \). Given that the same signal processing steps are applied on \( x_b(t) \) and \( x_w(t) \), we use (in this section) the notation \( x(t) \) to refer to any of them. Thus \( x(t) \) and the stimulation-signal were filtered through a 1-Hz wide bandpass linear-phase FIR filter centered at the stimulation frequency \( f \). This resulted in the signals \( s(t) \) and \( l(t) \) which are the bandpass filtered versions of \( x(t) \) and the stimulation-signal respectively. We consider \( s(t) \) to be an estimate of the SSVEP. Such an estimate is termed bipolar or spatially-filtered following that it results from \( x_b(t) \) or \( x_w(t) \).

Subsequently the spectral-temporal representation of \( s(t) \) and \( l(t) \) was obtained to estimate their instantaneous phases. This can be achieved using the Hilbert transform \[ \text{13] from which, one can derive the analytic signals \( a_{st}(t) \) and \( a_{lt}(t) \) corresponding to \( s(t) \) and \( l(t) \) respectively.

\[
\begin{align*}
s_a(t) &= s(t) + js_b(t) = R_s(t) \exp j \varphi_s(t) \\
l_a(t) &= l(t) + jl_b(t) = R_l(t) \exp j \varphi_l(t) ,
\end{align*}
\]

where \( s_b(t) \) and \( l_b(t) \) are the Hilbert transforms of \( s(t) \) and \( l(t) \).

The phase difference \( \Delta \varphi_{st} \), \( f \), \( t \) at frequency \( f \) and time \( t \) between the SSVEP estimate and the stimulation-signal can be obtained from

\[
\begin{align*}
\Delta \varphi_{sl} \ f \ , \ t &= \arg \exp j \varphi_s(t) - \varphi_l(t) \\
\Delta \varphi_{sl} \ f \ , \ t &= \arg \left\{ \frac{l_a(t)s_b(t)}{R_s(t)R_l(t)} \right\}
\end{align*}
\]

where the * operator stands for the complex conjugate. In practice, the phase is estimated within a time window. In this study, we select the mode of the distribution of \( \Delta \varphi_{sl} \ f \ , \ t \) within a given time window.

### 5. Experimental setup

Two 10×10 cm luminous panels 30 cm apart from each other were used to simultaneously render the visual stimulation. Each panel consisted of a 1-watt power green LED shining through a diffusion panel. The LED was driven by a square-shaped oscillating current. The maximum luminance of each LED was 1714 nits. The background luminance was 69.7 nits. The corresponding modulation depth was then: \( \frac{1714 - 69.7}{1714 + 69.7} \approx 92\% \). The SSVEP amplitude is positively correlated with the modulation depth \[ \text{14] . The SSVEP frequency for each subject was determined by presenting stimuli at all integer frequencies ranging from 30 to 40 Hz, recording the SSVEP, and identifying the frequency which elicited the highest SSVEP. Four subjects (three male and one female) participated in this study. Subjects had normal vision or corrected to normal vision. All subjects signed an informed consent before engaging in this study and had the right to quit at any time.
Subjects were asked to pay attention to one of the two luminous panels for 3 seconds. We refer to such a 3-second long period as a trial. We recorded forty trials separated from each other by a resting period of a random duration between 3 and 5 seconds.

In each trial, subjects were asked to focus their attention on a randomly selected panel. The random sequence was such that it resulted in 20 trials for each panel (phase). Subjects participated in two recording sessions, which resulted in a total of 40 trials per phase condition.

EEG signals were collected using a BioSemi Active-two EEG acquisition device in a normal office environment with curtains closed and lights on. Signals from thirty-two electrodes arranged according to the international 10-20 system were recorded at a sampling frequency of 2048 Hz. During the recording of EEG signals, subjects were requested to avoid movement or eye-blinking during the trials and advised to blink during the rest period between two consecutive trials. In addition, the light signal from the 0-phase panel was simultaneously recorded using a photodiode connected to the amplifier.

6. Results

Table 1 reports the optimal stimulation frequency for each subject. In addition, Figure 1 represents the distribution of the SSVEP energy for different stimulation frequencies (from 30 to 40 Hz) for the first two subjects (Figure 1a and b respectively). The box plots resulted from ten trials. As expected, the SSVEP amplitude has a decreasing trend as the stimulation frequency increases. Thus, the best stimulation frequencies for all subjects are respectively 30, 31, 32, and 32 Hz.

Figure 2 shows the phase difference between the SSVEP and the light signal extracted from 1-second long EEG segment recorded while the attention of subject 1 was focused on panel 1 (LED 1, 0-phase, thick line) and on panel 2 (LED 2, $\frac{2}{3}\pi$-phase, dashed line).

It is clear that the phase difference can be extracted by the phase synchrony analysis as it fluctuates around a certain constant value. The measured phase difference values can, therefore, be used to identify which panel received the subject’s attention.

In our study, we have also examined the effect of spatial filtering on the detectability of the phase information in the SSVEP signal. We used the algorithm discussed in Section Error! Reference source not found. to obtain the spatial filters for each subject. The spatial filters can be represented in a topographic map as shown in Figure 3. Given that each coefficient of the filter is associated with an electrode location, it can be represented using a color code which facilitates the interpretation of the result. As expected, the coefficients with high absolute values correspond to occipital locations which mainly capture the activity of the primary visual cortex. Subject variability of the spatial filters is clear from Figure 3. Even if subjects 3 and 4 share the same optimum frequency, their respective spatial filter configurations differ from each other.

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Table 1. Phase classification accuracy for the bipolar (Oz-Cz) and spatially filtered estimates of the SSVEP.
Figure 1. Box plot of the SSVEP energy for each stimulation frequency for subjects 1 (a) and 2 (b). These distributions were obtained from ten trials at each stimulation frequency.

Figure 2. Instantaneous phase difference extracted from a one-second long EEG segment recorded while the attention of subject 1 was focused on: panel 1 (LED 1 thick line) and panel 2 (LED 2, dashed line).

Figure 3. Topographic representation of the spatial filters for all subjects at their respective optimum stimulation frequency.
Figure 4 depicts the distributions of the phase differences extracted from the bipolar signal Oz-Cz (Figure 4a) and the spatially filtered signal (Figure 4b) for subject 1. Two phase differences were estimated per trial. Each phase difference resulted from analyzing, using the phase coherence method, a one-second long window. The first second of each trial was discarded because the SSVEP establishes few hundred milliseconds after stimulus onset [16]. As it can be seen from Figure 4, phase detectability is higher when the spatially filtered signal is used in the phase coherence analysis.

To assess the accuracy of phase detection, a support-vector-machine (SVM) classifier was used [1]. Half of the trials were used for training the SVM-classifier and the remaining half to assess the accuracy of phase detection. The results, reported in Table 1, show that spatially filtering the signals to enhance the SSVEP amplitude, facilitates the detection of the phase.

![Figure 4. Distributions of the phase differences of subject 1: (a) calculated from Oz-Cz and (b) calculated from the one channel signal after spatial filtering.](image)
7. Conclusions

The phase synchrony analysis can effectively extract the phase difference between the SSVEP and the light signal. The difference between these two values deviates slightly from the expected value $\frac{2}{3}$, but the difference is sufficient for detection.

We have used, the Hilbert transform to obtain the spectro-temporal representation of a signal. In principle, the Hilbert transform can be applied to any arbitrary signal to extract its instantaneous phase. Yet, the phase has a physical meaning only if the signal is a narrow-band signal. This is why we use an FIR filter centered at the frequency of interest. A Gabor wavelet convolution can alternatively be used to analyze neural synchrony as in [10]. The difference between these two methods is minor and they are fundamentally equivalent for the study of neuroelectrical signals. However, the Hilbert transform is slightly more efficient from the computational viewpoint.

As shown in Table 1, spatial filtering can significantly increase the classification accuracy. This indicates that spatial filtering is important not only for estimating SSVEP energy [7] but also SSVEP phase. In this study, the coefficients of the spatial filter for each subject are fixed. Thus, the phase difference after spatial filtering is non-linearly and invariably related to the SSVEP phase of the electrode signals which are used to design a spatial filter.

The stimulus phase can deviate from the pre-established value, especially if the stimulus is presented for long time. Using the phase difference between the SSVEP and the stimulus signal as a feature can compensate for this deviation, because the SSVEP is phase-locked.
8. References


