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Original Research Article

EEG with a reduced number of electrodes: Where to detect and how to improve visually, auditory and somatosensory evoked potentials

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ABSTRACT

The measurement of evoked potentials has become a standard tool to test new hardware and software for electroencephalography (EEG). In this study, we investigate where to detect and how to improve visually, auditory and somatosensory evoked potentials with a reduced number of electrodes. We measured a total of 50 evoked potentials in healthy subjects, and we were able to detect visually, auditory and somatosensory evoked potentials with just three electrodes. We also investigated where to measure a combination of visually, auditory and somatosensory evoked potentials and found the best positions to be Oz, O1, O2, TP9 and TP10. In the second part of this study, we analyzed how the evoked potentials depend on the segmentation frequency selected to superpose EEG responses. We found that the detection of visually evoked potentials requires the segmentation frequency to match the stimulus frequency with an accuracy of at least 99.92 percent. The detection of auditory evoked potentials and somatosensory evoked potentials requires a matching of at least 99.95 percent. Therefore, a correct matching of the segmentation frequency with the stimulation frequency is the primary key to improving the quality of evoked potentials.

1. Introduction

Electroencephalography is a well-established clinical technique to investigate the activity of the brain and to diagnose neurological malfunctioning [1]. But the recording of an electroencephalogram (EEG) always involves a lot of effort: Frequently 32 or more electrodes are fastened to the scalp [2,3]. Each of these must be checked manually for a very good electrical contact to the skin—a very time-consuming procedure that is unacceptable for a fast clinical diagnosis. The many wires connected to these electrodes often cause cable clutter that must be cleaned up for the next measurement. The subsequent sterilization is very time-consuming too and carries the risk of breaking cables.

Recently more and more applications have been designed that require EEG signals from selected areas of the brain only, for example brain–computer interfaces [4–6] and neurofeedback [7–9]. In many of these applications, sensory perceptions

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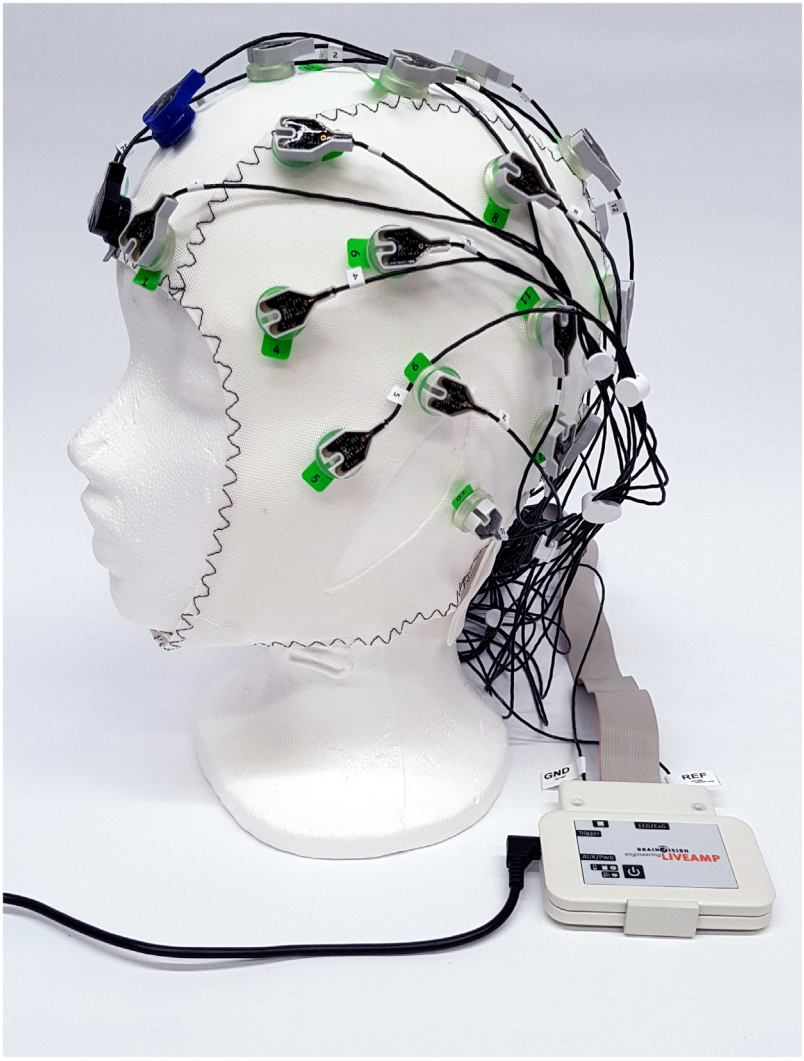

Fig. 1 – Mobile 32-channel device with electrode cap and trigger cable.

are to be detected such as visually, auditory and somatosensory evoked potentials [10]. Apparently the number of electrodes can be significantly reduced here, thereby minimizing the wiring effort and the amount of data to be analyzed [11,12]. We are well aware of the fact that reducing the number of electrodes will result in a loss of information [13]. But we are confident that with our strategy we can better focus on the relevant information. Most likely a reduction in EEG channels will trigger many new applications with a mobile hardware (so-called ''wearables'').

In this study, we investigate the positions on the scalp that are best suited for the measurement of evoked potentials. Most importantly we show at which positions all three kinds of evoked potentials (visually and auditory and somatosensory) can be detected. We also describe how sensitive the detection is with regard to selecting the correct segmentation frequency. Evoked potentials can be detected only as a superposition of many repeated single responses. If the segmentation frequency of the superposition does not exactly match the repetition frequency of the stimuli, the evoked potential will completely dissolve within the noise.

2. Evoked potentials

Generally speaking, evoked potentials (EP) are expressions of electrical activity within the brain that occurs with a temporal delay after a specific event. These can be sensory or mental events. In this study, we focus on sensory EP only. They are classified as visually evoked potentials (VEP), auditory evoked potentials (AEP) and somatosensory evoked potentials (SEP), respectively.

First experiments in 1875 already demonstrated that visual stimuli cause small electrical potentials within the brains of apes [14]. But it was not until Hans Berger's work on encephalography in 1929 [15] und the subsequent development of non-invasive signal detection techniques, that VEP, AEP and SEP could be measured in the human brain [16]. At the beginning of EP research, these signals were very difficult to detect because the very first EPs were single responses only. The primary task was to distinguish the small evoked EEG activity from the much stronger non-evoked brain activity. It was George Dawson who first used electrical superposition

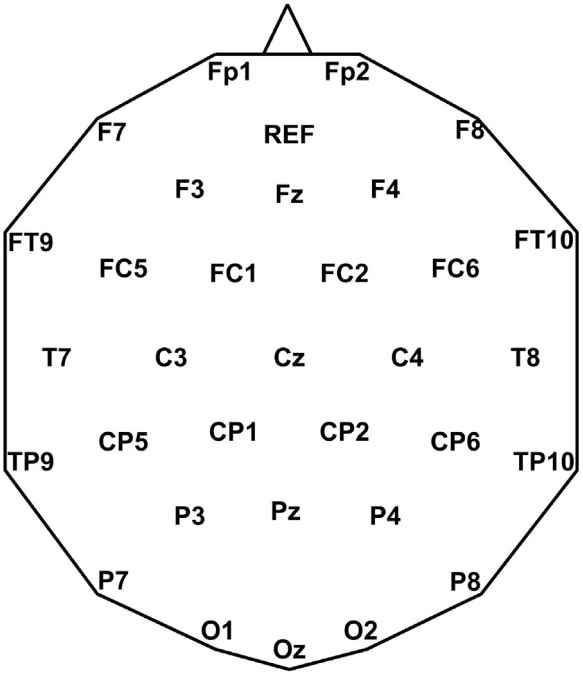

Fig. 2 – Array of electrodes according to the extended 10–20 system.

techniques in electroencephalography [17,18] enabling systematic studies of VEP, AEP und SEP in the human brain.

Nowadays EP play a significant role in clinical diagnostics, especially in analyzing brain activity and signal processing in our nervous system. Significant EP parameters such as amplitude, signal width and latency between stimulus and response are measured and analyzed [19]. Since the response is temporally locked-in to the point in time when the stimulus was given, the EP can be distinguished from the background activity by superposing the responses to repeatedly triggered stimuli. The signal-to-noise ratio is improved due to the superposition itself: The deterministic component (the responses to the stimuli) will *constructively* superpose, while the stochastic component (the background activity of the brain) will *destructively* superpose, resulting in a clear EEG response to the EP. For this to happen, the segmentation frequency f_{seg} must closely match the repetition frequency f_{stim} of the stimuli. The segmentation frequency itself is defined as:

$$f_{seg} = \frac{1}{T_{seg}} \tag{1}$$

where T_{seg} is the duration of the superposed segments. The mismatch δ between f_{seg} and f_{stim} can be expressed as:

$$f_{seg} = f_{stim}(1-\delta) \tag{2}$$

3. Materials and methods

We measured a total of 50 evoked potentials in healthy subjects (28–53 years old, all of them males). Each subject agreed to participate in this study before measurements were made. VEP were measured by presenting a checkerboard (pattern reversal technique) at a flickering frequency of 2 Hz. The subject was sitting in a dark room 60 cm in front of a monitor. For the AEP measurements, we applied a 1000 Hz sinus signal synchronously to both ears via earphones. The sinus signal was sent in 30 ms bursts (including a 10 ms linear rise and a 10 ms linear fall segment) at a repetition frequency of 0.5 Hz. SEP were measured by stimulating the *Nervus Medianus* synchronously on both hand wrists with a maximum current of 50 mA, a stimulation period of 50 μs and a repetition frequency of 2 Hz.

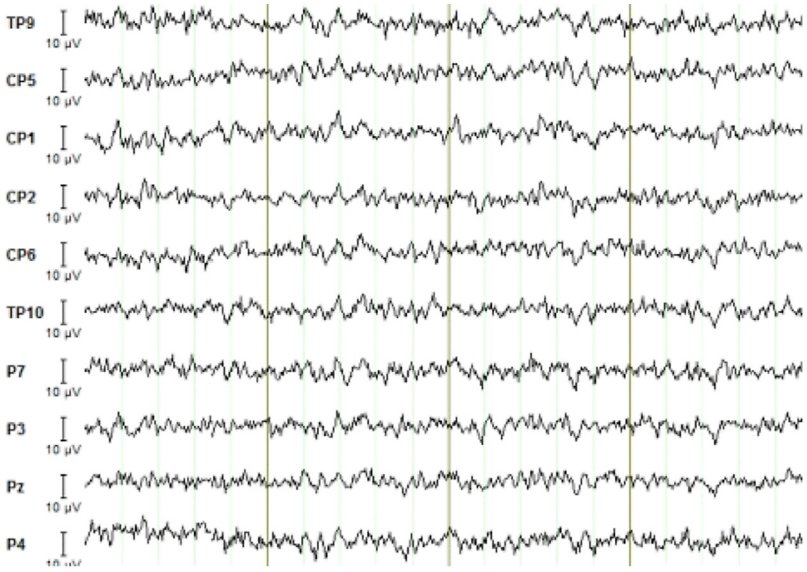

Fig. 3 – Monitoring mode of the BrainVision Recorder.

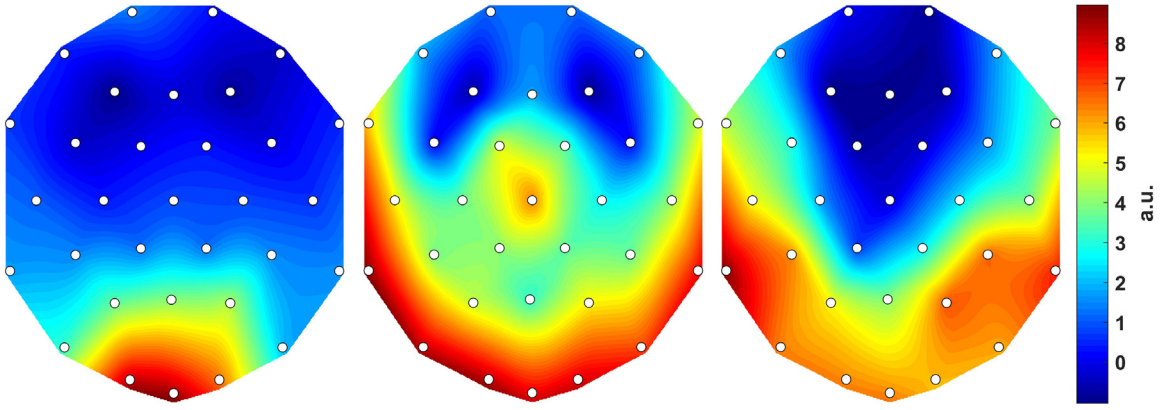

Fig. 4 – VEP (left), AEP (center) and SEP (right) as topographic maps.

3.1. *EEG hardware*

All of the EEG signals were measured with the mobile 32-channel device *LiveAmp* by BrainVision Inc. (Morrisville, USA). The data were transmitted via Bluetooth and recorded on a PC. In addition to the 32 EEG channels, a 3-axes sensor detected any body movements of the subject and recorded them in separate channels 33 through 35. These channels enabled us to clearly identify motion artifacts. In order to record the exact moments of the stimuli, a trigger signal was generated from the stimuli that added a time stamp to all of the 35 channels. The sampling rate was 500 kHz, the detection bandwidth was ±341.6 mV, and the resolution was 40.7 nV/bit at a 24-bit AD-conversion. Fig. 1 shows the experimental setup with the 32-channel device, an EEG electrode cap and the trigger cable for adding the time stamps.

We used 32 *ActiCAP* sensors by BrainVision Inc. (Morrisville, USA) as EEG electrodes. These are active silver-silverchloride electrodes with an integrated noise reduction and an optical impedance control in real-time (RGB LEDs). All of the 32 EEG electrodes and the ground electrode were plugged into the designated sockets of a textile EEG cap (model *EasyCap* by BrainVision Inc.). The sockets were positioned according to the extended 10–20 system as illustrated in Fig. 2.

3.2. *EEG software*

We recorded the EEG measurements with the *BrainVision Recorder* by BrainVision Inc. (Morrisville, USA). One key feature of this software is that it provides multitasking: In a preview window, the impedance of each electrode-skin contact is decoded in color. In the monitoring mode, all of the 35 channels are shown on top of each other and on a moving time line (see Fig. 3). This mode was selected for monitoring the recorded signals and for judging the quality of the data during the measurement. Both time scale and amplitude of the signals were adjusted according to the type of measurement. The software also enabled us to connect the mobile measurement device via Bluetooth and to start, to interrupt or to stop a new measurement.

During the EEG measurements the software also enabled us to set markers and text comments that facilitated the recovery of specific events when evaluating the data. These markers were also used in MATLAB algorithms specially designed for this study, for example to indicate the start of a specific stimulation.

After recording the brain activity, the data were processed and analyzed with the MATLAB software by MathWorks Inc. (Natic, USA). A first algorithm converted the raw EEG data into MATLAB data sets. A second algorithm helped us to match the segmentation frequency with the repetition frequency of the stimuli. With a third algorithm we superposed up to 360 responses to single stimuli. Finally, a fourth algorithm enabled us to visualize the evoked potentials as topographic maps.

4. Results and discussion

Fig. 4 shows VEP, AEP and SEP as topographic maps. In this illustration, the maximum amplitude at each electrode is color-coded (red: highest activity, blue: lowest activity). The areas in between the electrodes were interpolated in terms of

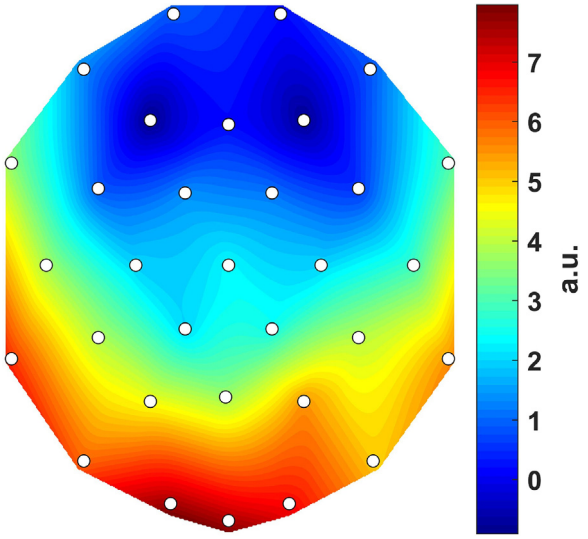

Fig. 5 – Superposition of VEP, AEP and SEP as topographic map.

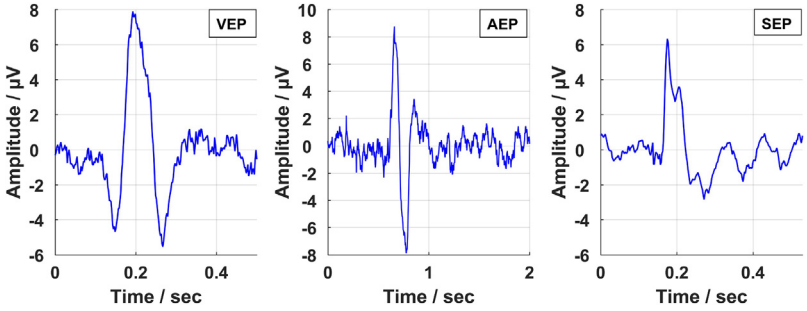

Fig. 6 – VEP, AEP and SEP at channel Oz.

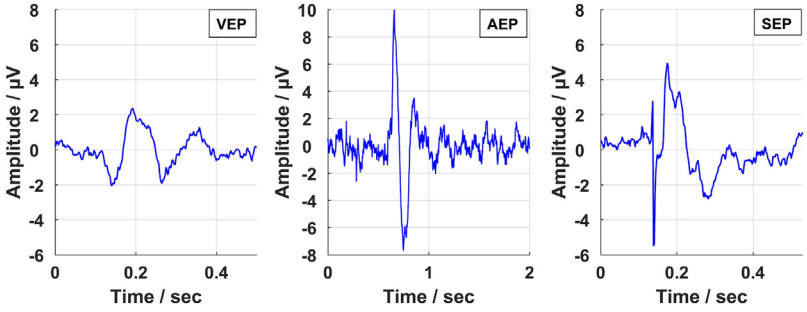

Fig. 7 – VEP, AEP and SEP at channel TP9.

color. It is evident that the main activity evoked by visual, auditory or somatosensory stimulation is measured in the *Lobus Occipitalis* and in the *Lobus temporalis*. There are almost no evoked potentials present in the *Lobus frontalis*. We found the best visually evoked potentials to be detected at the positions Oz, O1 and O2 in the extended 10–20 system. The best auditory evoked potentials were detected at TP9, TP10, P7, P8, O1, O2 and Oz. The best somatosensory evoked potentials were detected at TP9, TP10, CP5 and CP6.

We have also superposed all three EPs as of Fig. 4 because several EEG applications involve all three types of EPs: VEP, AEP as well as SEP. Fig. 5 shows the superposed topographic map at a uniform weighting (one third of each). The best positions to detect a combination of VEP, AEP and SEP are Oz, O1, O2, TP9 and TP10.

Fig. 6 shows the signal amplitude of the evoked potentials measured at the electrode position Oz. The EPs clearly stand out from the non-evoked brain activity. Fig. 7 shows the signal amplitude of the evoked potentials measured at the electrode position TP9 (left mastoid). Even at this position next to the ear, all three types of EPs have been detected. These results indicate that "wearables" with a few electrodes only should be suitable for many EEG applications.

We also analyzed how the evoked potentials depend on the segmentation frequency selected to superpose EEG responses. Figs. 8–10 show examples of VEP, AEP and SEP for three different values of the mismatch δ.

To estimate the maximum permissible mismatch δ we developed a MATLAB algorithm that automates the superposing of the EEG responses while varying the segmentation

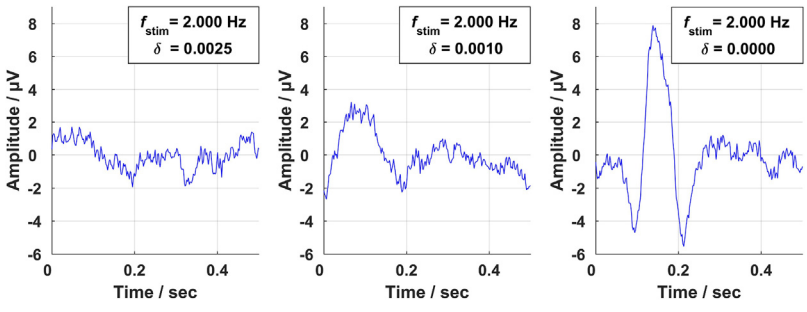

Fig. 8 – VEP sensitivity at channel Oz.

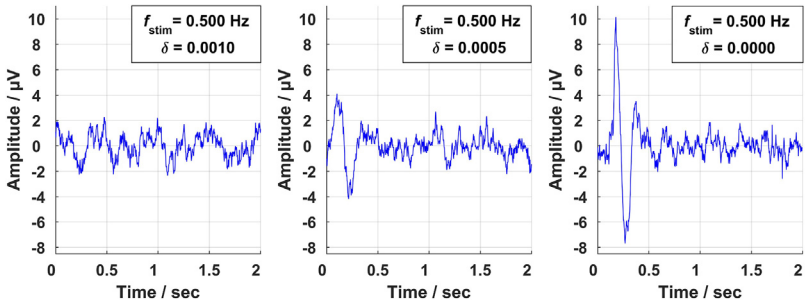

Fig. 9 – AEP sensitivity at channel TP9.

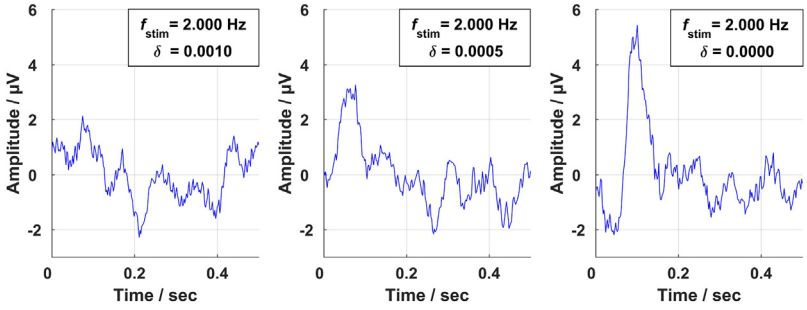

Fig. 10 – SEP sensitivity at channel CP6.

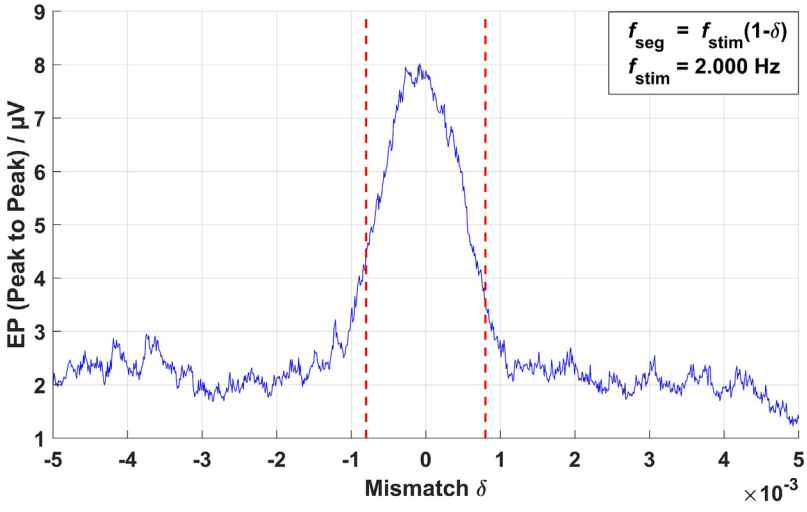

Fig. 11 – VEP amplitude affected by mismatch δ at channel Oz.

frequency f_{seg}. The algorithm works with increments of 0.001 percent of the stimulation frequency f_{stim}. Figs. 11–13 illustrate the amplitude of the evoked potentials (peak to peak) versus the mismatch δ. In each of these figures, the dotted red line marks the maximum permissible mismatch δ that still gives a valid EEG response (50 percent of the highest peak to peak amplitude). Fig. 11 shows that the detection of VEP requires the segmentation frequency to match the stimulus frequency with an accuracy of at least 99.92 percent (δ = 0.0008). Fig. 12 shows that the detection of AEP requires the segmentation frequency to match the stimulus frequency with an accuracy of at least 99.95 percent (δ = 0.0005). Finally, Fig. 10 shows that the detection of SEP also requires the segmentation frequency to match the stimulus frequency with an accuracy of at least 99.95 percent (δ = 0.0005).

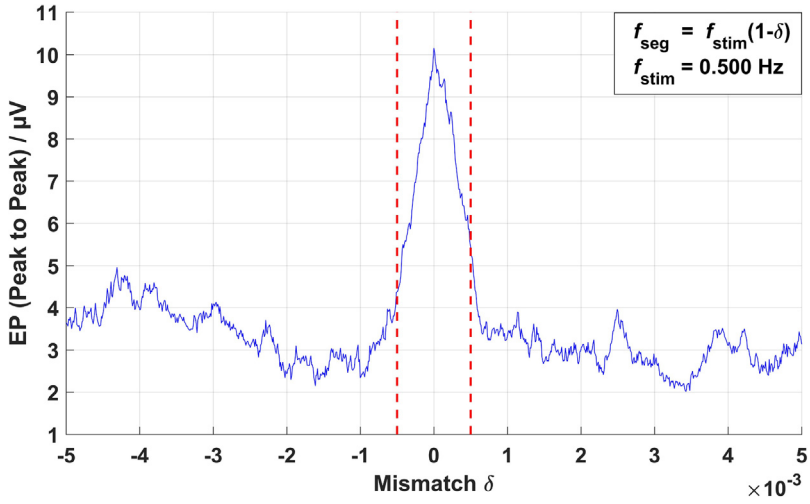

Fig. 12 – AEP amplitude affected by mismatch δ at channel TP9.

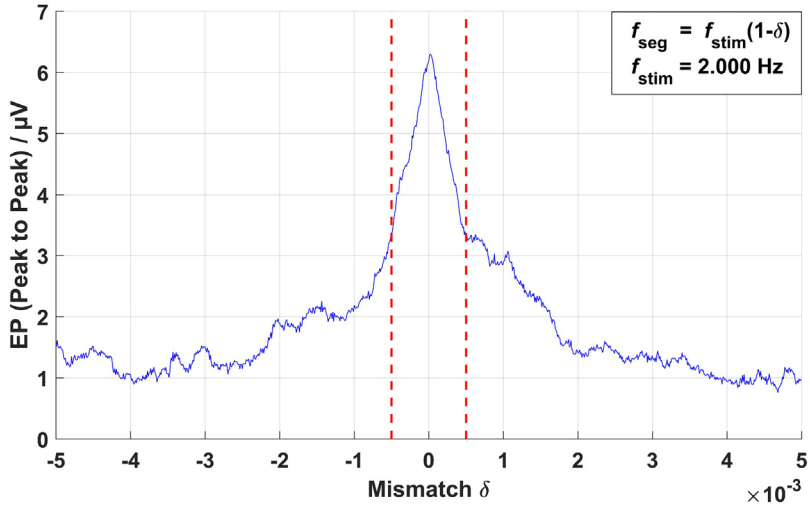

Fig. 13 – SEP amplitude affected by mismatch δ at channel CP6.

5. Conclusion

This study confirms our original assumption that evoked potentials can be detected with a few EEG electrodes only. Figs. 4 and 5 will assist developers of future EEG applications in selecting the best electrode positions. Our results indicate that the position Oz will be the most favorable electrode position in many cases but—depending on the type of application—other positions in the *Lobus Occipitalis* and in the *Lobus temporalis* will probably yield a sufficient signal-to-noise ratio too. The positions TP9 and TP10, for example, are very promising as these electrodes can easily be connected to temples of glasses.

But our study also shows that one needs to be very careful when matching the segmentation frequency with the stimulation frequency. A correct matching of these two frequencies is the primary key to improving the quality of evoked potentials. When taking this into account, we believe that reducing the number of channels will give rise to many novel EEG applications in the future, especially in the segment of "wearables".

Conflict of interest

None.

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