Clinical Neurophysiology 118 (2007) 98–104

www.elsevier.com/locate/clinph

A fully automated correction method of EOG artifacts in EEG recordings

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Accepted 4 September 2006
Available online 7 November 2006

Abstract

Objective: A fully automated method for reducing EOG artifacts is presented and validated.
Methods: The correction method is based on regression analysis and was applied to 18 recordings with 22 channels and approx. 6 min each. Two independent experts scored the original and corrected EEG in a blinded evaluation.
Results: The expert scorers identified in 5.9% of the raw data some EOG artifacts; 4.7% were corrected. After applying the EOG correction, the expert scorers identified in another 1.9% of the data some EOG artifacts, which were not recognized in the uncorrected data.
Conclusions: The advantage of a fully automated reduction of EOG artifacts justifies the small additional effort of the proposed method and is a viable option for reducing EOG artifacts. The method has been implemented for offline and online analysis and is available through BioSig, an open source software library for biomedical signal processing.
Significance: Visual identification and rejection of EOG-contaminated EEG segments can miss many EOG artifacts, and is therefore not sufficient for removing EOG artifacts. The proposed method was able to reduce EOG artifacts by 80%.

Keywords: Electroencephalogram; Electrooculogram; Artifact processing; Missing values; Regression analysis

1. Introduction

Artifacts in general and specifically the electrooculographic (EOG) artifact is a major noise source in electroencephalogram (EEG) recordings. One can assume that every EEG recording is contaminated with EOG artifacts, because eye movements are difficult to suppress over a sustained period of time. For example, in a study on sleep EEG, 9.1% of the total recording length was contaminated with EOG artifacts (Schlögl et al., 1999b).

The origin of EOG is due to electrical eye activity (electroretinogram) which is propagated throughout the body via volume conduction and can be recorded at the body surface. EOG artifacts are caused by retinal dipole movement and eyelid movement (Croft and Barry, 2000), causing potential shifts on the body surface. A simplified model assumes an electric dipole (Berg and Scherg, 1991) within the eyeball. The direction of the dipole is aligned with the line of sight; the size of the dipole (i.e. the amplitude) is determined by the amount of light hitting the retina in the back of the eye. Because eye blinks and saccades (volitional and non-volitional ocular activity) cause "topographic and morphological differences" of EOG artifacts (e.g. Ille et al., 2000; Picton et al., 2000), some have suggested different correction methods. Although the appearance is different, both are caused by the same dipole in 3-dimensional space. For this reason, it is important to capture the EOG activity by no more and no less than three spatial dimensions of each eye.

In most cases, both eyes are in the same line of sight and observe the same luminance. Thus, the dipoles of both eyes are parallel and strongly coupled (i.e. highly correlated). Therefore the EOG can be modeled by a single dipole

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1388-2457/$32.00
doi:10.1016/j.clinph.2006.09.003

consisting of three spatial components (horizontal, vertical and radial). An exception is the occurrence of rapid eye movements (REM) during sleep, where the eye movements are not in synchrony; the vertical and horizontal movements of both eyes are independent. In that case, the horizontal and vertical eye movement must be captured separately for each eye.

EOG amplitude is attenuated approximately with the square of the distance (Croft and Barry, 2000), contaminating mostly the frontal EEG channels. Due to the volume conduction effect, EOG and EEG activity are propagated to the head surface where the superposition of both is recorded. The weighting coefficients are determined by the spatial relationships and the electrical properties of the tissue between the sources and the electrodes. These properties do not change during a recording session; in other words, the weighting coefficients are stationary. Exceptions are eye lid movements, which change the geometry of the surrounding tissue. However, the influence can be modeled as a change in the radial component (i.e. magnitude in axial direction) of the EOG. The assumption of stationary coefficients has also been investigated by several studies (Girton and Kamiya, 1973; Van den Berg-Lenssen et al., 1989; Croft and Barry, 2000) and no non-stationary property of the correction coefficients was observed.

There are several different approaches for artifact processing: avoiding the occurrence of artifacts, correction of artifacts and rejection after identification (Anderer et al., 1999). EOG artifacts can be hardly avoided, artifact rejection results in loss of data. Moreover, there are several methods for EOG reduction: the spatial filter (Lagerlund et al., 1997; Ille et al., 2000), blind source separation (BSS) (e.g. Joyce et al., 2004; Barbati et al., 2003) and the regression methods (e.g. Moretti et al., 2003; Wallstrom et al., 2004) for EOG subtraction. Some works (e.g. Ille et al., 2000; Joyce et al., 2004 and Barbati et al., 2003) are criticizing the regression approach because regression-based methods may reduce cortical activity as well. It was never investigated whether the alternative methods "guarantee" that no cortical activity is removed. Especially, in case of a small number of EEG channels it's very likely that principle component analysis (PCA) and independent component analysis (ICA) perform worse than regression analysis. Anyway, no quantitative comparison on a representative dataset is available. On the other hand, some recent works (e.g. Moretti et al., 2003 and Wallstrom et al., 2004) suggest that the regression method is appropriate for EOG reduction. For these reasons, the critics on regression-based methods are at least disputed. Moreover, this work will suggest some measures for minimizing possible negative effects of the regression methods, and the reduction of the cortical activity will be quantified, too.

In this work, a regression-based EOG reduction method is applied for removal of fast and slow EOG-related artifacts. The regression method is chosen because it is simple and robust, and it can be used with any number of EEG channels. In contrast with previous studies, validation is based on blind scoring, which is carried out by expert EEG analysts (see Section 2 for details). This ensures an independent validation of the presented EOG reduction method (cf. Croft and Barry, 2000).

2. Model

In agreement with the work of Elbert et al. (1985) and motivated by the previously described mechanism we assume the following linear model with three spatial (horizontal, vertical, and radial) EOG components:

$$Y(t,\mathrm{ch}) = S(t,\mathrm{ch}) + [\mathrm{EOG1}(t), \mathrm{EOG2}(t), \mathrm{EOG3}(t)] \cdot [b_1(\mathrm{ch}), b_2(\mathrm{ch}), b_3(\mathrm{ch})]^{\mathrm{T}}$$

whereby $Y(t,\mathrm{ch})$ is the recorded value of channel ch at time t, S is the source signal without artifact contamination, EOG123 indicates the noise source U of the three spatial EOG components, and $b(\mathrm{ch})$ indicates the weights of the EOG artifacts at the EEG channel ch and T means matrix transposed. Extending this to more EEG channels and using a matrix notation, we can write:

$$Y_{TxM} = S_{TxM} + N_{Txn} \cdot b_{nxM}$$

The indices indicate the size of each matrix; the signals Y and S have T time points and M channels, the noise U has n components and b denotes the weights from each EOG component to each EEG channel; subsequently these indices are omitted. In order to obtain the corrected signal $S = Y - U \cdot b$, the noise source U and its weighting coefficients b must be known. In this work, the noise source (i.e. EOG) was recorded by separate EOG channels. In order to identify the weighting coefficients b, we assume the signal S (i.e. EEG) and the noise U (i.e. *EOG*) are independent, then

$$< U^{\mathrm{T}}S > = < U^{\mathrm{T}}Y > - < U^{\mathrm{T}}U > b$$

with $<U^{\mathrm{T}}S> = 0$ results in

$$b = < U^{\mathrm{T}}U >^{-1} < U^{\mathrm{T}}Y > = C_{NN}^{-1}C_{NY}$$

with $C_{NN} = <U^{\mathrm{T}}U>$ is the auto-covariance matrix of the EOG channels and $C_{NY} = <U^{\mathrm{T}}Y>$ is the cross-covariance between the EEG and EOG channels. Accordingly, the EEG can be corrected by the following equation:

$$S = Y - U \cdot b$$

In order to obtain data for calculating the correction coefficients, a short data segment was recorded during which the subjects performed eye movements. Specifically, the subjects were instructed to move their eyes (according to Table 1). This provides EOG activity in all three spatial dimensions and the recorded data contains the propagation of the EOG activity throughout the head surface. The induced eye movements result also in large EOG activity, which reduced the influence of possible EEG interferences at the EOG electrodes. The procedure was applied once

Table 1
Instructions on performing eye movements. This procedure should ensure that large EOG artifacts are recorded. The data recorded during the eye movements were used to estimate the correction coefficients

(1) Perform clockwise and counter-clockwise rolling of the eyes for several seconds. The eyes should circumscribe the whole field of view without moving your head. Alternatively or additionally, horizontal and vertical eye movements can be performed.
(2) Perform repetitively eye blinks. The steps should be repeated for approximately 1 min.

after mounting all electrodes and before starting the actual EEG recordings.

Furthermore, automated overflow detection was applied in order to identify a saturation of the amplifier or analog-digital converter. Unlike in the work of Schlögl et al. (1999a), the saturation values were provided by the recording system. Saturating sample values were encoded as "missing value" using the symbol Not-A-Number (NaN) according to the standard IEEE 754 (1985). The data with these missing values were handled by the NaN-toolbox (part of BioSig 2005). For the estimation of the covariance matrices, the missing values (encoded by NaNs) were ignored. Accordingly, samples with saturation effect were not considered for the computation of the correction coefficients b.

2.1. Data recording

The data recorded for this work was part of a series of experiments for an EEG-based brain–computer interface. Ten healthy subjects (4 female, 6 male, age between 17 and 31) (8 subjects with 2 sessions and 2 subjects with 1 session) took part in these experiments. Each volunteer was sitting in front of a LCD monitor and was instructed not to move. Twenty-two EEG channels, 3 monopolar EOG channels, one ECG and one respiratory channel were recorded. In this work, only the EEG and EOG channels were used, they shared a common reference electrode at the left mastoid and a ground electrode at the right mastoid (Fig. 1). The three EOG electrodes were positioned above the nasion, and below the outer canthi of the eyes, forming a rectangular triangle. Note that bipolar EOG channels "left-central" and "central-right" were able to capture horizontal and the vertical EOG components.

The EEG and EOG data were band pass filtered with a broadband anti-aliasing filter from 0.5 to 100 Hz and a 50 Hz notch filter, sampled with 250 Hz and 12 bit quantization. The dynamic ranges for EEG and EOG were $\pm 100\ \mu V$ and ± 1 mV, respectively.

The recording system consisted of two 16-channel amplifiers (g.tec, Graz, Austria), two data acquisition cards (National Instruments Corporation, Austin, USA) and a commercial desktop PC running under WindowsXP. The software for data recording, analysis and controlling the BCI experiment were implemented in MATLAB 6.5 and Simulink 5.0 (The MathWorks, Inc., Natick, USA) using rtsBCI (Scherer, 2005) and the open source package BioSig (2005).

The cue-based (synchronous) paradigm consisted of four imagery classes: motor imagery of left hand, right hand, foot, and tongue, whereby the subjects were instructed to imagine the desired movement depending on the cue. Each trial started with a fixation cross and an additional short acoustic stimulus. In total 18 recording sessions were analyzed. Each session consisted of a calibration recording (ca. two minutes) followed by a BCI experiment of 6 runs with 48 trials each. In this work, only the first run of each session was used for scoring and validation. The duration of each trial was 6 s with a random inter trial interval of

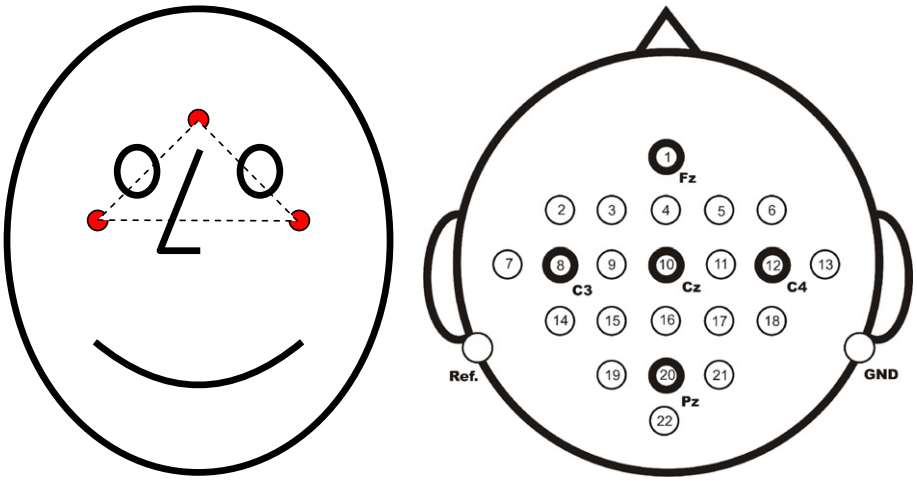

Fig. 1. Position of EOG (left) and EEG (right) electrodes. The three EOG electrodes are positioned at the corners of a right-angled triangle; the legs of the triangle form two spatially orthogonal components. The corresponding bipolar EOG components are able to capture the horizontal and vertical EOG components. The EOG electrodes were positioned close to the eyes in order to minimize the influence of non-EOG components.

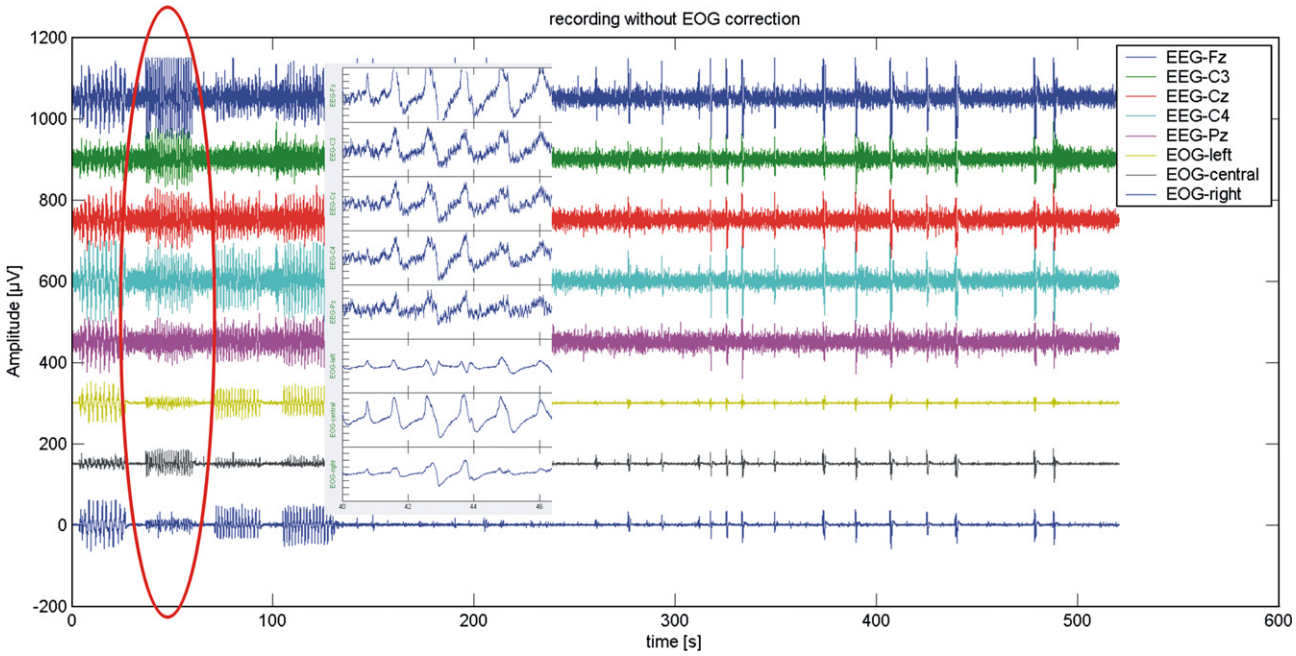

Fig. 2. Original recording. Five out of 22 EEG channels and 3 EOG channels are shown (the EOG is scaled 1:10). During the first 120 s, the subject performed eye movements; the 4 EOG bursts correspond to eye blinks and clockwise and counterclockwise rolling. Afterwards the subject performed the BCI experiment.

1.5 to 2.5 s. The subjects were asked not to move their eyes from cue onset ($t = 2$ s) until the end of the trial. Accordingly, the subjects blinked with the eyes during the break, resulting in a repetitive pattern of EOG artifacts. Fig. 2 shows the two minute calibration recording and the EEG recording of the first run (ca. 6.5 min).

2.2. Validation

For the purpose of validation, the 22 EEG channels of 19 experiments of approx 400 s (ca. 6–7 min each) have been transformed into new data files. All non-EEG channels (including EOG) have been removed.

The filenames were randomized, such that the experts could not know whether the correction method was applied or not (blinded evaluation). A total of 38,307,060 samples (18 recordings, 22 channels, 250 Hz sampling rate, and an average recording length of 6.45 min) have been scored twice (with and without EOG correction) by two expert scorers. For data scoring, the viewing and scoring software SVIEWER from the BioSig (2005) was used.

Two types of EOG artifacts have been distinguished: (i) the fast EOG artifacts which usually have also large amplitude, and (ii) slow EOG artifacts which usually have smaller amplitude. The former is caused mostly by eye blinks, the latter by eye movements. In order to distinguish (small) EOG artifacts from slow wave EEG, the distribution of the wave patter was considered – EOG artifacts have large amplitude in frontal electrodes and smaller amplitude at parietal channels.

3. Results

Fig. 2 displays 5 out of 22 EEG channels and 3 EOG channels of the original recording. In the first segment (up to ca. 120 s) the subject performed various eye movements (rolling, blinking, horizontal and vertical). Then, after a short break, the EEG experiment was started. One can clearly see that the EEG data is frequently corrupted by EOG artifacts.

In the next step, the previously described artifact reduction method, based on regression analysis, has been applied. The EOG channels were re-referenced, and two bipolar EOG channels "EOG central-left" and "EOG central-right" were obtained. These were sufficient to capture the horizontal and the vertical EOG component. In order to avoid removing global EEG activity, the radial EOG component has not been used. For estimating the weighting coefficients, the data from the first segment with large eye-movements has been used. These weighting coefficients have been used for correcting the whole recording implicitly assuming that the correction coefficients do not change with time. The results are shown in Fig. 3. The EEG channels do not show any large EOG artifacts.

The two independent expert scorers distinguished between fast EOG artifacts, slow EOG artifacts and no EOG artifact. The agreement between both expert scorers was 95.6% and had a kappa of 0.607. In the (uncorrected) raw data, the expert scorers identified in 4.8% of the time fast EOG artifacts, in 1.1% slow EOG and in 94.1% no EOG artifact was marked (Table 2). From these 4.5 Mio samples

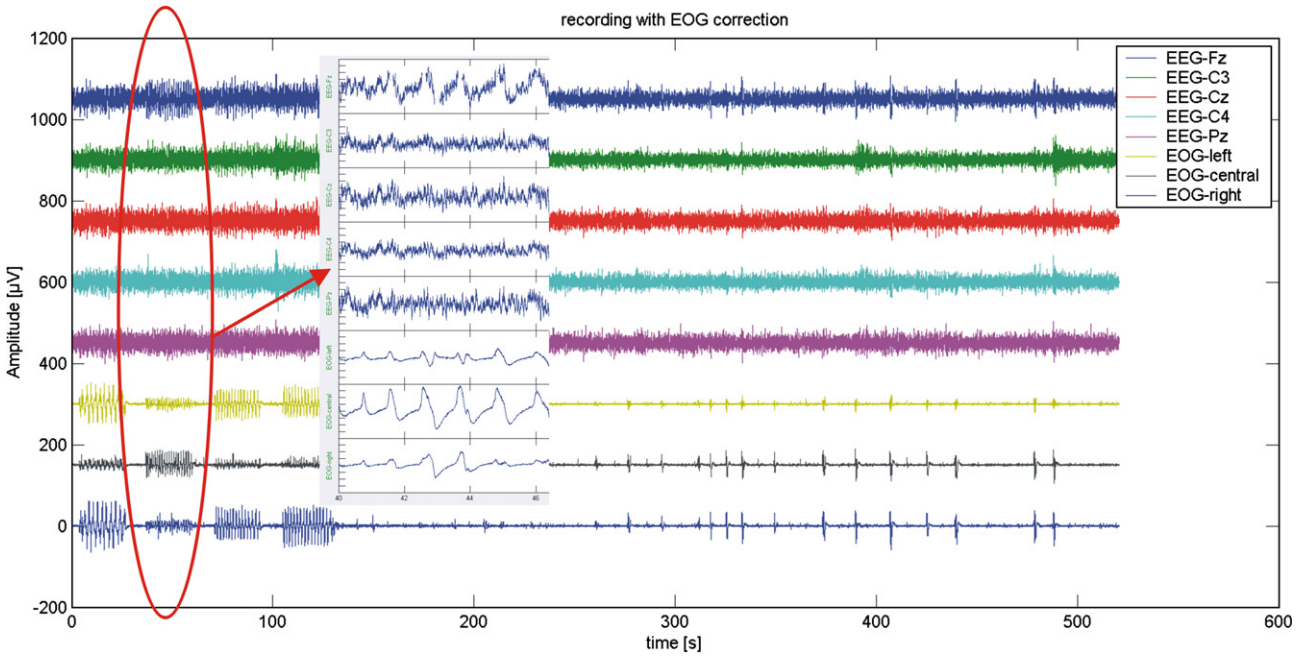

Fig. 3. EEG recording with EOG correction. The EOG correction has been applied to the data from Fig. 2. The EOG channels indicate the eye movements, no large EOG artifacts are observed in the EEG channels.

(5.9%) of artifact-contaminated raw data, 3.6 Mio (3.8% fast + 0.9% slow) were successfully corrected by using the proposed method; this is a reduction of approx. 80%.

However, after applying the correction method, the experts identified EOG artifacts in another 1.9% (1.2% fast + 0.7% slow) of the whole data set. Accordingly, 3.1% (2.2% fast and 0.9% slow) were identified as artifacts in the corrected data set.

In order to investigate the effect of the EOG-reduction method on artifact-free data, the discrepancy of the power spectral density (PSD) is computed. For this purpose, those data segments were marked as missing, which were marked by at least one of the expert scorers in either the corrected or uncorrected recording. Then, the PSD was computed from the remaining data samples using an autoregressive model of order $p = 50$. The resulting PSD of one subject (v4) on a frontal (channel 1, Fz) and a most occipital electrode (see channel configuration in Fig. 1) is shown in Fig. 4. Furthermore, the discrepancy (difference between corrected and uncorrected data) as well as the ratio between corrected and uncorrected data is shown. It demonstrates that the discrepancy is largest in the frequency range below 5 Hz and is larger in the frontal than in the occipital electrodes. The discrepancy is almost zero for larger frequencies (8 Hz and above), partly because the PSD becomes small, too.

In order to quantify the relative changes, the PSD ratio between corrected and uncorrected data is presented. Fig. 4 shows that, for higher frequencies, the PSD ratio is close to 1; even in the frontal electrode no more than 5% of the PSD are removed. In the low frequency range, the ratio can be much smaller indicating a larger reduction. This indicates that some EOG activity was recorded during the "artifact-free" data (a pure volume conduction of EEG activity towards EOG electrodes would have caused a constant ratio for the whole frequency range). This indicates that EOG activity was also reduced in segments where the expert scorers did not identify any artifact.

4. Discussion and conclusion

In this work, a fully automated method for correcting EOG artifacts is proposed. The proposed method for EOG reduction requires the following provisions:

- Montage of three EOG electrodes.
- One minute instruction and 2 min (or even less) additional recording time.

Table 2
Confusion matrix between scoring of raw data and corrected data

Samples (%)	No artifact (corrected)	Fast/large EOG (corrected)	Slow/small (corrected)	Total of raw data scoring
No artifact (raw data)	70,647,770 (92.2%)	947,364 (1.2%)	521,092 (0.7%)	72,116,226 (94.1%)
Fast/large (raw data)	2,947,606 (3.8%)	655,086 (0.9%)	69,344 (0.1%)	3,672,036 (4.8%)
Slow/small (raw data)	681,538 (0.9%)	61,314 (0.1%)	83,006 (0.1%)	825,858 (1.1%)
Total of corrected data	74,276,914 (96.9%)	1,663,764 (2.2%)	673,442 (0.9%)	76,614,120 (100%)

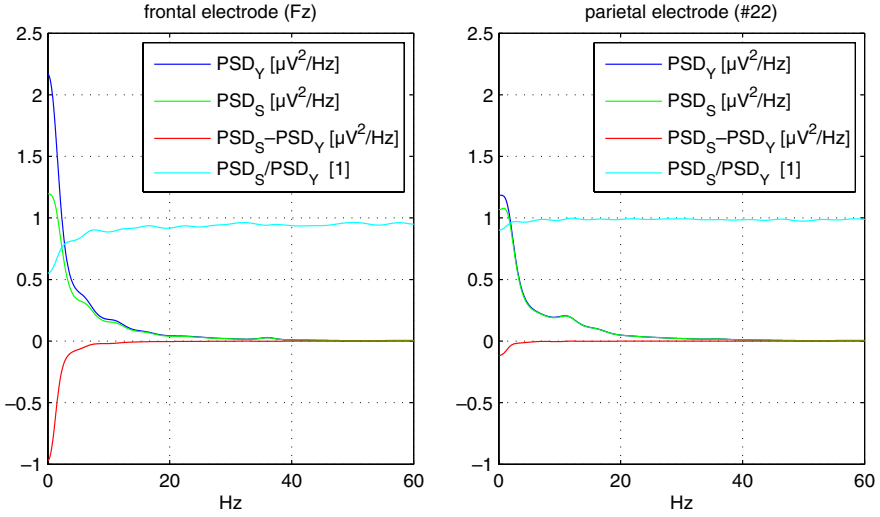

Fig. 4. Discrepancy of "artifact-free" data in the frequency domain. All segments marked as artifacts were removed (marked as missing samples); the remaining data was used to compute the power spectral density (PSD) obtained through an AR(50) method. PSD_Y and PSD_S indicate the PSD without and with EOG correction, respectively. Moreover, the discrepancy (PSD_S–PSD_Y) and the ratio PSD_S/PSD_Y is shown.

- Calculation of the correction factors b requires the computation of two correlation matrices, one matrix inversion and one matrix multiplication. Today's computers can compute this within a fraction of a second. In this work, it took approx. 0.18 s for 22 EEG and 2 bipolar EOG channels on a 1600 MHz CPU.
- The filters of the recording system must be the same for EEG and EOG channels. Otherwise, the assumption of a linear superposition will not hold.
- No saturation (overflow) shall occur in any EOG channel, or saturated sample values must be encoded as missing values.

This additional effort reduces possible negative effects of the regression method, because the use of large EOG signals minimizes the influence of possible EEG contamination of the EOG channels when learning the correction coefficients. The additional effort is minimized by using a scheme of three instead of four EOG electrodes, and by providing the software for this procedure through the BioSig project [BioSig, 2005].

The independent evaluation of expert scorers demonstrated that EOG artifacts can be reduced by approx. 80%, only 1.2% out of 5.9% were not corrected. After applying the correction method, the expert scorers identified EOG artifacts in another 1.9% of the whole data set. It can be excluded that the additional contamination is due to the correction method, because without any EOG amplitude, the corrected and uncorrected signals are the same. Therefore, the only possible explanation is the subjective interpretation. This was confirmed by the expert scorers, saying that for data sets with fewer artifacts, it was more likely to identify also small artifacts. Nevertheless, the experts are confident that these were actual EOG artifacts. This statement is supported by the relative high agreement between both scorers ($\kappa = 0.6$). Moreover, EOG artifacts could be also reduced in segments, where visual artifact scoring did not identify any EOG artifact. Consequently, a simple EOG rejection approach is not a viable option when dealing with EOG artifacts. Alternatively, the fully automated correction method should be applied, and the visual scoring should be performed on the corrected data set.

Nevertheless, the present method reduced EOG artifacts but did not eliminate them. The most likely explanation is the fact that only 2 instead of 3 spatial EOG components have been removed. This choice was done on purpose, because the third component can not be (easily) assessed but is contaminated by global EEG activity. This problem of a remaining radial component would be solved if someone finds a clever way for identifying the third (radial) EOG component clean of any EEG. If the number of EEG channels is sufficiently large, maybe PCA or ICA can be useful for this purpose.

The major critique on regression-based EOG correction is based on the fact that EEG activity could be picked up by the EOG channel; this could cause removal of EEG activity, too. This issue was addressed in this work twofold: (i) the EOG electrodes were mounted in the proximity of the eyes, and (ii) the correction coefficients were calculated from segments with large EOG artifacts. A quantitative analysis shows that the PSD reduction ranges from approx. 5% for frontal EEG channels to less than 1% for occipital channels. In order to cope with this attenuation, Croft and Barry (2000) suggested applying correction factors.

In summary, it is demonstrated that the proposed method reduces significantly EOG artifacts. Unlike PCA or ICA, the present method can be applied to any number of EEG channels (even for single channel EEG), whereas PCA, ICA and BSS can be used only if the number of EEG electrodes is sufficiently large. Furthermore, no expert decisions are necessary and no subjective knowledge is needed. Thus, the method is fully automated and the result

is determined by the data only. The method is simple and robust, no time-delay is introduced and the method is well suited for online processing. Actually, the method has been implemented for offline and online processing. The offline analysis is available through BioSig (2005), the online implementation of the correction method is available through "rtsBCI" (Scherer, 2005).

In summary, the advantages of the proposed method for reducing the EOG artifacts are worth the additional effort, and the proposed approach is a viable method for EOG reduction. Moreover, the presented validation procedure can be used to quantify the performance of alternative EOG correction methods.

Acknowledgements

This work was partly supported by the EU-Project IST – 2001 – 37027 "Presencia", Allgemeine Unfallversicherungsanstalt (Austria), "Landesregierung Steiermark" (Federal State of Styria in Austria) and the Austrian Science Foundation (Fonds zur Förderung der Wissenschaftlichen Forschung) project P16326-B0.

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