Simultaneously Recorded EEG–fMRI: Removal of Gradient Artifacts by Subtraction of Head Movement Related Average Artifact Waveforms

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Abstract: Electroencephalograms (EEGs) recorded simultaneously with functional magnetic resonance imaging (fMRI) are corrupted by large repetitive artifacts generated by the switched MR gradients. Several methods have been proposed to remove these distortions by subtraction of averaged artifact templates from the ongoing EEG. Here, we present a modification of this approach which accounts for head movements to improve the extracted template. Using the fMRI analysis package statistical parametric mapping (SPM; FIL London) the head displacement is determined at each half fMRI-volume. The basic idea is to apply a moving average algorithm for template extraction but to include only epochs that were obtained at the same head position as the artefact to be removed. This approach was derived from phantom EEG measurements demonstrating substantial variations of the artefact waveform in response to movements of the phantom in the MRI magnet. To further reduce the residual noise, we applied a resampling algorithm which aligns the EEG samples in a strict adaptive manner to the fMRI timing. Finally, we propose a new algorithm to suppress residual artifacts such as those occasionally observed in case of brief strong movements, which are not reflected by the movement indicator because of the limited temporal resolution of the fMRI sequence. On the basis of EEG recordings of six subjects these measures combined reduce the residual artefact activity quantified in terms of the spectral power at the gradient repetition rate and its harmonics by roughly 20 to 50% (depending on the amount of movement) predominantly in frequencies beyond 30 Hz. Hum Brain Mapp 30:3361–3377, 2009. © 2009 Wiley-Liss, Inc.

Key words: simultaneous EEG–fMRI; gradient artifact; resampling; head movement

INTRODUCTION

The recording of the electroencephalogram (EEG) during functional magnetic resonance imaging (fMRI) using echo planar imaging (EPI) protocols has become increasingly popular since this technique was first introduced by Ives et al. in 1993 (refer Herrmann and Debener [2008] for a review). This task is technically challenging since the switched gradients needed for MR imaging continuously elicit EEG artifacts [gradient artifacts (GAR)] with extremely large amplitudes [Garreffa et al., 2003; Hoffmann et al., 2000]. Other less powerful types of artifacts are the ballistocardiogram (BCG), which is produced by the acceleration and abrupt directional changes in blood flow in the aortic arch during each heart beat [Ives et al.,

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1993], and from a subject's slight motions in the static magnetic field due to heart beat and respiration. In this article, we address the temporal fluctuations of the GAR and present an artifact removal technique which adapts itself to these ongoing variations. Such fluctuations in the GAR shape could be caused by minor MR-related instabilities as well as by subjects' movements (i.e., slowly occurring shifts in head position, nodding-like head movements related to the cardiac cycle, brief transient head movements due to swallowing, etc.).

Several groups have presented different approaches to eliminate or avoid the GAR. Among these methods are adaptive filtering [Hoffmann et al., 2000; Sijbers et al., 1999; Sijbers et al., 2000; Wan et al., 2006], and blanking the artifact contaminated periods [Goldman et al., 2000], and clustered MR-scanning [Kruggel et al., 2000]. Others have proposed techniques to remove the GAR by means of a principal component analysis (PCA) [Negishi et al., 2004], and by the averaged artifact subtraction (AAS) procedure introduced by Allen et al. [2000]. This technique has recently been adapted and modified by a variety of groups [Becker et al., 2005; Cohen et al., 2001; Goncalvez et al., 2007a,b; Laufs et al., 2003; Lemieux et al., 2001; Mandelkow et al., 2006; Niazy et al., 2005; Schmid et al., 2006]. The AAS approach assumes a stable repetitive artifact template is being superimposed on the ongoing EEG. This template can be estimated from the measured EEG by averaging over all gradient onset-related epochs. A subsequent subtraction of the extracted template leads in principle to an artifact-free EEG. However, in practice, it turns out that the artifact shape slightly varies over time, thus violating the stability assumption of the AAS. Therefore, Becker et al. [2005] introduced a weighted moving averaged template subtraction (MAS) method which takes into account potential time dependant fluctuations by only averaging over a limited number of surrounding artifacts with weighting factors decreasing with increasing temporal distance. Although the template subtraction method can eliminate most of the artifactual activity, some residual activity still persists, especially in higher frequency ranges due to the combined steep slope and extremely high amplitude of the GAR. To further reduce this residual artifact, post processing techniques have been proposed, including low pass or adaptive filtering [Allen et al., 2000; Becker et al., 2005] and the estimation of optimal basis functions by PCA with the aim of modelling the temporally varying details of the residual artifact [Niazy et al., 2005; Schmid et al., 2006].

Here, we analyze the strength and the temporal dynamics of these fluctuations and their relation to involuntary head movements. For this purpose both phantom and human EEG data were recorded during fMRI scanning. The results of these analyses lead to an optimization of the MAS method that we term movement adjusted moving averaged subtraction (MAMAS). In contrast to the MAS algorithm, the MAMAS method does not average over immediately adjacent EEG epochs but rather over epochs

obtained at similar head positions. A further improvement is achieved by a new algorithm that serves to remove artifacts with differing shapes that occur during brief transient head movements. To speed up the algorithm we replaced the upsampling algorithm usually applied in AAS and MAS by a resampling technique that precisely matches the sampling times of the artifact to be removed. Finally, we demonstrate the improvement achieved with this method over the standard MAS result.

MATERIALS

Artifact Recording From a Phantom

A watermelon with a diameter 140 mm and simulated electrode contacts was positioned inside the MR scanner (Fig. 1A,C) to serve as a phantom for the recording of pure GAR examples with no interfering physiological human signals. Twenty-nine EEG amplifier inputs plus a reference and ground were connected to the phantom (Fig. 1B) via 10 K Ω antimagnetic resistors (DALE CMF55) that were plugged into the melon tissue to simulate a real measurement situation. For the analyses described below we focused on the signal that was picked up at a position that corresponded to the P4 electrode of a human lying in the MR-bore during fMRI epi scanning, (refer Fig. 1C).

In four consecutive runs signals of 10 min/run (corresponding to 300 MRI-volumes) were recorded. Between each run the watermelon was rotated by 2° to simulate a human head movement. During the scans the watermelon was fixed with a block of plastic. As shown in Figure 1D the displacement of the P4 electrode per rotation step was about 0.5 mm in the vertical (i.e., y) direction resulting from a 2.4 mm displacement in tangential direction which is comparable to a realistic strong head movement.

Human EEG Measurements

Six subjects (average age 25, range 20–27 years, one female) with no neurological deficits participated in the study. The EEG was recorded over three sessions, each of 9:30 min duration: (i) while the subjects were sitting outside the shielded MR-cabin, (ii) lying inside the MR bore without active MRI scanning, and (iii) during active fMRI scanning. Signals were referentially recorded from 29 Ag/AgCl-electrodes at positions Fp1, Fp2, F3, F4, C3, C4, P3, P4, O1, O2, F7, F8, T7, T8, P7, P8, Fz, Cz, Pz, FC1, FC2, CP1, CP2, FC5, FC6, CP5, CP6, TP9, TP10 according to the international extended 10–20-system. The reference electrode was located between the Fz and the Cz electrode. Eye movements, recorded by means of an electrooculogram (EOG) and an electrocardiogram (ECG) were also included. For technical reasons (the subject's comfort) a modified electrode setting (F3, F4, C3, C4, P3, P4, PO3, PO4, FC5, FC6, CP5, CP6, F7, F8, T7, T8, P7, P8, TP9, TP10, PO9, PO10, Fpz, Fz, Cz, Pz, POz, Oz, Iz) was

Figure 1.

A: Schematic diagram of the phantom EEG recordings. **B:** EEG electrode connected to the watermelon. **C:** Position of the watermelon inside the MR-magnet. **D:** Relation of the melon's rotation angle and the displacement of the P4 electrode in vertical (y) direction. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

applied with two of the six subjects. Electrode-skin impedance was kept below 5 K Ω in all cases.

METHODS

EEG Acquisition

Human EEG and phantom data were recorded continuously using a BrainAmp MR EEG-amplifier (Brain Products, Munich), which was located close to the subjects head or to the phantom, respectively, but outside the magnet bore (on the subject's bed), and attached by adhesive tape. The signals were band-pass filtered with lower/upper cut off frequency of 0.016 Hz/250 Hz and subsequently digitized at 5000 Hz, with 16 bits of resolution (the least significant bit representing 500 nV). After digitization, the data were transferred to a PC outside the shielded magnetic resonance (MR) cabin by a fiber optic cable. All electrode cables were anchored using small sandbags to prevent movement.

Functional Magnetic Resonance Imaging

FMRI images were acquired in a 3T MR scanner (Siemens Trio) using an eight channel head coil and running an EPI protocol with the following parameters: Echo time (TE) 40 ms, repetition time (TR) 2000 ms, flip angle 80°, field of view (FOV) 20 × 20 cm, matrix size 64 × 64, slice thickness 5.0 mm, 20 slices (subjects 1–4) or 24 slices (subjects 5, 6, and phantom), with no gap between slices.

Determination of Artifact Onset and Duration

A trigger code was provided by the MR scanner at the onset of each volume imaging. It was recorded by the EEG device simultaneously with the biosignals. Given that the MR image acquisition requires one gradient per slice, the duration in terms of sampling points was defined as $TR/ns \times fs$ with ns representing the number of slices and fs representing the EEG recorder's sampling frequency. Thus, the onsets of each switched gradient can be determined by means of the trigger code and the duration of each slice

by the calculation of a cross correlation. In this manner, we were able to specify the artifact onset with a temporal precision of 200 μ s as defined by the sampling rate.

Movement Detection

Some head movements by the subjects can usually not be avoided during MR imaging. To include movement information in the artifact removal algorithm we monitored subjects' movements using the statistical parametric mapping (SPM99, FIL London) program, which derives movement parameters (displacement in three orthogonal direction and rotation around these axes) from the sequentially acquired MR images as part of the statistical analysis. Given an interleaved image acquisition, the temporal resolution is TR/2, i.e., 1 s. In a typical fMRI measurement environment, head motions are most likely to occur around the x-axis resulting in displacement along the y-direction (refer Fig. 1C). Therefore, we used the y-displacement provided by SPM as a movement indicator guiding the artifact averaging process (refer below). The orientation of the SPM y-direction is roughly parallel to the MRI-y-axis (refer Fig. 1C).

In addition, we applied a proprietary MR compatible video based eye tracker [Kanowski et al., 2007] that was modified to trace head-movements with a 20 ms temporal resolution, so that we were able to relate occasional large residual artifacts to transient head movements that were detected by visual inspection of the eye tracker signal. However, it should be acknowledged here that our movement related approach to removing the GAR is in general not able to cope with fast movements (e.g., those associated with nodding) that are not capable of being captured by the SPM movement detection algorithm, i.e., those persisting less than half a TR.

Subtraction Algorithm and Optimized Estimation of the Artifact Template

Given temporal fluctuations of the artifact shape as reported by Allen et al. [2000], Becker et al. [2005], and others, we applied a weighted moving average window to extract the artifact template needed for the gradient locked subtraction algorithm [(MAS), Becker et al., 2005]. The goal of the present study was to optimize the MAS subtraction algorithm by:

- i. quantifying the temporal fluctuations of the artifact template,
- ii. comparing various weighting functions,
- iii. artifact resampling,
- iv. comparing slice selective versus uniform artifact templates,
- v. accounting for head movements, and
- vi. applying a special algorithm to reduce residual artifactual activity occurring in case of transient brief movements.

(i) Quantifying the Temporal Fluctuations of the Artifact Template as Observed in Watermelon Recordings

Initially, we observed that an average over at least 50 artifact free EEG epochs is required to get the amplitude of the residual EEG down to the technical noise of the amplifiers (being determined as 1 μ V STD at this bandwidth, measured with shortened inputs). This means, that at least 50 artifact epochs need to be averaged to extract an EEG-free artefact template. Moreover, difference amplitudes between two extracted artefact templates showing a standard deviation clearly exceeding 2 μ V can be attributed to changes in the artifact pattern. To quantify the temporal artifact fluctuations, we averaged 50 consecutive artifacts observed in the first run of the watermelon recordings starting slightly after the start of the MR scan. Next, this procedure was repeated, after shifting the sequence of artifact epochs by a delay of 20, 40, 60, and 80 epochs. Then each of these shifted average templates was subtracted from the first one leading to a sequence of difference amplitudes. Therefore, the resulting differences reflect changes due to temporal fluctuations within one run, i.e., while the position and the angle of the phantom remained fixed. The artefact templates were extracted applying the newly developed resampling scheme [refer subsection (iii) below].

To test the possibility that the temporal instability observed in these recordings (refer Results) was scanner specific we conducted the same measurements in a 1.5 T MR scanner GE signa LX and analyzed the data the same way. The functional MR sequences were as much as possible adapted to the sequence applied on the 3T scanner.

(ii) Comparison of Various Weighting Functions with Respect to the Efficiency of the Artifact Removal

The MAS algorithm as described by Becker et al. [2005] applies a symmetric window function defining the weights by which the artifacts are included in the moving average. As outlined in the previous section, at least 50 epochs should be averaged to derive an artifact template free of EEG residuals. Moreover, the bandwidth of the weighting function should be kept below 90 consecutive epochs [refer results of sub chapter (i)]. The exponentials weighting profile

$$W(n) = 0.9^{|n-i|}$$

proposed by Becker et al. [2005] (refer Fig. 2A) and later in the manuscript referenced as the "Becker window" meets this specification with n as the index of the consecutive epochs, i specifying the index of the window center, and the total width extending over 120 epochs. However, to yield an artifact estimate with an negligible residual

Figure 2.

A: Profile of the evaluated weighting functions. **B:** Residual EEG activity after weighted averaging of artifact free EEG applying the exponential profile proposed by Becker et al., 2005. **C:** Power spectrum averaged over 10 consecutive short term spectra and averaged over all 29 EEG channels; typical example after MAS

artifact removal. **D:** Same as C, obtained from 10 other EEG segments recorded during strong movement of the same subject. **E:** Short term EEG-epoch after removal of GAR using the various weighting profiles.

EEG it needs to be demonstrated that the residual EEG (after weighted averaging) (REEG) stays below the EEG amplifier noise. Therefore, we averaged separately for each EEG-channel N consecutive epochs (length 80 ms

each) of EEG signals recorded outside the scanner, applying the central N weighting coefficients of the function and then calculated the standard deviation according to Eq. (1).

$$\text{REEG} := \frac{1}{S \cdot C} \sum_{s=1}^S \sum_{c=1}^C \text{std} \left(\frac{1}{\sum_{n=1}^N W(n)} \sum_{n=1}^N W(n) * \text{EEG}_{\text{scn}} \right) \quad (1)$$

These numbers were averaged over all C ($=29$) EEG channels and all S ($=6$) subjects. On the basis of these results, we determined (i) if the weighted average keeps the residual EEG below the technical noise and (ii) the minimum N required to meet this constraint.

In addition, we compared other weighting functions (square root, linear, square, rectangular, each extending over $N = 120$ epochs, refer Fig. 2A) to the exponential approach of Becker et al. with respect to the efficiency of artifact removal. The reduction of spectral components at the gradient repetition frequency and its harmonics served as a quality criterion. A more efficient removal will result in a stronger decrease of these spectral peaks. Visual comparison of the EEG before and after artifact removal was used as an additional subjective criterion.

(iii) Artifact Resampling

Given the large amplitude and steep gradients of the artifacts under consideration, a missing synchronization between the EEG sampling clock and the timing of the gradient switching can lead to substantial residual artifacts after subtraction, unless each artifact included in the average is resampled to match precisely the phase of the sampling points of the artifact to be removed (subsequently called the “reference artifact”). Resampling here means the calculation of data samples at a predefined shifted grid of virtual sampling times. It has been proposed that this problem can be solved by massive oversampling of the signal [Becker et al., 2005], subsequent alignment (guided by cross correlation) and finally downsampling. Others [Allen et al., 2000] have resampled the EEG synchronously with the slice timing trigger by means of interpolation techniques. Recently, an electronic solution for the synchronization of the MR scanner and the EEG digitizer clock has been introduced by a commercial company (Brain Products, Munich) thereby providing a simple way to capture the true waveform at precisely the same relative sampling times as the reference artifact. Goncalves et al. [2007a,b] addressed this issue in the context of their approach of iteratively optimizing the MR-sequence parameters (time/volume, time/slice, onset delay before first slice) measured in “EEG-time”. Because in our MR environment a trigger is available at the beginning of each volume, we did not need to run this computationally costly procedure. Instead, we propose a software approach that avoids upsampling but appropriately resamples each artifact that is included in the average to precisely match the timing of the samples that represent the reference artifact. This technique broadly resembles the electronic solution and the approach of Allen et al. mentioned above, but is additionally capable of tracking potentially spurious jitter

that might occur during the MR slice acquisition. The new algorithm is also computationally more efficient than that used by Becker et al. [2005] and Goncalves et al. [2007a,b].

The resampling algorithm derives for each artifact an individual temporal adjustment of the signal sampling (while keeping the original sampling frequency) by the application of a cubic spline interpolation scheme, i.e., it determines sampling times and corresponding amplitude values best matching the sampled reference artifact, which is to be removed. For reasons of computational efficiency, the precision of this procedure should be kept at the lowest acceptable limit, and the search algorithm that determines the best adjustment should be optimized. Details regarding this optimized algorithm and the simulated artefact waveform are described in the supplementary material.

(iv) Comparison of Slice Selective Versus Uniform Artifact Template

As shown by Goncalves et al. [2007a,b], the subtraction of MR-slice-selective rather than a uniform artifact template may lead to an improved artifact removal. We checked this approach in the context of our setting by applying the following procedure to the phantom data: Artifact templates of slices number 2–24 were subtracted from the artifact observed at slice number 1. The standard deviation of the resulting differences was compared with the amplifier’s noise level to test for systematic differences.

(v) Accounting for Head Movements

During a fMRI experiment subjects usually move their head involuntarily from time to time, altering the position and orientation of the electrodes and cables with respect to the magnetic gradient. Different artifact shapes can be produced at these different positions, and this may cause distortions in the averaged artifact template, leading to a less efficient artifact removal by subtraction. We quantified this effect by recording a full 32 channel signal during fMRI scanning from the watermelon phantom in four 10-min-sessions at different positions, as described in the Materials section.

Before further analysis these data were filtered with 34-order FIR high pass of 10 Hz cut off frequency to avoid distortions due to low pass fluctuations which might result from slow electrochemical processes. Next, the artifacts were separately averaged for each position. As will be demonstrated in the results section we encountered significant differences between the four positions.

Given these observations, we modified the MAS-algorithm to extract the averaged artifact as follows: Since the subjects are usually properly fixed with respect to left-right-movements, empirically they tend to make movements around the x-axis (producing y-direction shifts according to the SPM notation). Therefore, we used the y-component of the SPM movement vector as an indicator of

Figure 3.

A: SPM-movement indicator [vertical (y) direction] covering a 9 min fMRI recording session as indicated by the horizontal grid the different position classes used for the position dependent template extraction technique are equally distributed over the

full range. **B:** Example of the position guided selection of artifact epochs included in the average to extract the artifact template. The weighting coefficients are distributed over epochs pertaining to the same position class.

the subject's head position. The total geometric span of the head movement observed within a scan period was split into 10 sub intervals leading to different groups of artifacts ("position groups") that pertained to different position levels (refer Fig. 3A).

As before, the average artifact was estimated by a window of artefacts centered on the actual reference artifact epoch to be removed. However, only those epochs were included which were recorded in the same position group as the reference epoch. If an epoch did not match this constraint, it was replaced by the temporally nearest one pertaining to the same position group while still applying the same weighting factor (refer Fig. 3B). In case of longer lasting movements, the length of the averaging interval needed to reach at least the minimum number of averaged epochs (refer section ii) could be larger than the actual duration defined by the maximum number of consecutive useable artefact epochs [which was set to 200, based on the results of section (i) and taking into account the shape of the Becker window which strongly focuses on the central epochs (refer Fig. 2A)]. In this case, epochs with position values of the adjacent position groups are accepted as well. If needed, this rule is further relaxed (however, this situation occurs extremely rarely). In the remainder of this article, we label this procedure the MAMAS algorithm.

The improvement produced by MAMAS was evaluated by a modified version of the "improvement normalized power spectrum ratio" (INPS) introduced by Tong et al. [2001] in the context of removing ECG artifacts from the EEG. For the present study this measure was modified to quantify the percentage of relative improvement of

MAMAS over the MAS algorithm. This was done by defining the following ratio:

$$\text{RINPS} = 100 * \frac{\sum_{j=N_{\text{low}}}^{N_{\text{up}}} P_j^{\text{MAS}} - \sum_{j=N_{\text{low}}}^{N_{\text{up}}} P_j^{\text{MAMAS}}}{\sum_{j=N_{\text{low}}}^{N_{\text{up}}} P_j^{\text{MAS}}}$$

with P_j representing the spectral power at the j th harmonic (averaged over a 1 Hz interval) of the gradient repetition rate g , with the superscripts "MAS" and "MAMAS" indicating the two algorithms and the summation extending over the frequency range from $N_{\text{low}} * g$ to $N_{\text{up}} * g$ Hz. This measure was calculated for the following frequency bands: $g = 30$ Hz, 30–100 Hz, 100–200 Hz. It was derived twice by averaging over spectra estimated for all "strong movement" and for all "mild movement" 6-s epochs of each of the 6 subjects. Movements were qualified as "mild" if the standard deviation of the difference sequence of y-displacements over 6-s-epochs stayed below 0.1 mm over the 6 s. In contrast, if the standard deviation was larger than 0.2 mm, the epoch was defined as "strong". These two thresholds roughly represent the 40- and 80-percentile limits of the empirical distribution of the standard deviation values. The improvement documented with RINPS can exclusively be attributed to the movement guided average, since the resampling was applied in both the MAMAS and the MAS algorithm.

To rule out the possibility that the artefact reduction is achieved at the cost of a distortion of the underlying true EEG signal (refer Debener et al., [2007] for further discussion) we applied the MAS and the MAMAS algorithm to

Figure 4.

Example of an EEG epoch showing strong residual artifact activity after the application of the MAMAS algorithm in the context of a brief movement not resolved by the SPM-movement vector. Middle: EEG epoch. Upper: Zoomed (50-fold) version of the residual artifact. Lower: Real time position tracking (blue) with

high temporal resolution (25 ms) versus position information as provided by SPM (red) with a resolution defined by the fMRI scanning procedure (1 s). [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

an EEG recorded outside the MR scanner. For this purpose, we defined a sequence of virtual GARs, i.e., a sequence of time points (simulating gradient onsets) spaced by 100 ms (according to a TR of 2 s and 20 slices/volume) and then added a random fraction of the EEG sampling interval to each artefact onset time. Next, we applied the MAS and the MAMAS algorithms, using the defined time points as the gradient onset times, and resampling each EEG epoch to match the sampling times of the artefact to be removed. To incorporate the effects of movement the MAMAS algorithm was guided by a SPM-movement trace taken from a fMRI data set obtained from a strongly moving subject. Finally, to compare the original EEG to the two processed versions, the power spectrum

was estimated by using the algorithm of Welch [1967] (40 epochs of 6 s duration, Hamming window).

(vi) Removing Residual Artifactual Activity Occurring in Case of Transient Brief Movements

As previously noted by Allen et al. [2000], we occasionally observed atypical artifacts that occurred during brief subject movements. Because the temporal resolution of the SPM-information is limited, these movements were not reliably detected. Consequently, the estimated reference artifact does not match the actual template, leading to a large residual artifact after subtraction. In the example shown in Figure 4, the relatively fast video based

movement detector clearly shows a transient brief movement which is associated with an almost stable SPM movement index.

Artifacts of this type can be easily detected if an amplitude threshold criterion applied to the ongoing EEG signal. Here, we set the threshold to twice the maximum peak-to-peak amplitude observed in a short period before start of MR scanning. However, these residual artifacts usually persist for only a few slices, making it impossible to extract an artifact template by averaging over neighboring slices. To overcome this problem, we exploited the fact that empirically the actual artifact template is very much similar (although differing in absolute amplitude and polarity) across all channels (refer Fig. 4), with the strongest opposing polarity artifacts occurring at electrode locations F3 and Iz, respectively. Given this observation we replaced the weighted temporal averaging by a weighted average over channels to extract an artefact template needed for subtraction. Details are found in the supplementary material section.

This method reduced the residual artifact enough to allow for a visual inspection of the ongoing EEG. However, with respect to quantitative analyses, the more conservative approach of discarding these epochs may be more appropriate because the amount of residual artefact that is not readily visible as well as residual grand average EEG overlaid on each EEG-signal by the subtraction cannot be controlled.

(vii) Removal of Vibration and of Residual High Frequency Activity

The vacuum pump of the scanner gives rise to another continuous and generally (but not strictly) periodic artifact with a frequency range between 40 and 50 Hz, usually peaking around 47 Hz. This artifact does not occur in all types of scanners. If it cannot be avoided it may be suppressed by an appropriate band-stop filter. Here we applied a third order band stop of Butterworth characteristic (cut off 40 Hz and 50 Hz). Residuals of the GARs at higher frequencies (beyond 120 Hz) were suppressed by an order 9 Chebychev Type II low pass filter with 120 Hz cut off frequency and 20 dB stop band ripple.

RESULTS

(i) Temporal Fluctuations of the Artifact Template as Observed in the Watermelon Recordings

The differences between the average artifact template obtained for the sequence starting at artifact epoch 300 after the onset of the MR scan and those observed at later epochs are depicted in Figure 5A. The data represent the mean of all standard deviation values calculated separately for each channel of the watermelon. In Figure 5C,

the difference waveform of the artifacts observed at electrode P4 during period 300 and 340 (i.e., 2 s apart) is shown. Obviously, fluctuations of the artifact's shape occur even over short intervals and tend to increase with longer temporal shifts. Given the 2 μ V threshold (refer methods) fluctuations can only be neglected for temporal shifts of less than 40 epochs (i.e., 40 slices). That means, that given that 50 artifact periods were included in the current average the number of periods used to extract the artifact template should be kept below 90 artifact periods. Correspondingly, if a weighted average is applied the kernel function should be shaped to focus on this limited number of periods.

One could argue that the temporal fluctuations are specific to the scanner we used. However, similar measurements conducted in a second MR scanner with a different field strength and supplied by different company show a very much comparable variability of the GAR waveform (refer Fig. 5B) suggesting that this observation reflects a more general effect.

(ii) Comparison of Various Weighting Functions with Respect to the Efficiency of the Artifact Removal

The results of the weighted averaged EEG epochs are shown in Figure 2B for the Becker profile. Obviously, the residual EEG is below the noise floor if the number of averaged epochs exceeds 45. This means that the Becker-window meets the requirement of a negligible residual EEG.

In terms of the residual activity at the gradient repetition rate and its harmonics remaining after artefact removal the Becker window outperformed all other profiles under any circumstances (strong/mild movement, all channels, all subjects). In most cases, this advantage was marginal (refer Fig. 2C presenting the power spectrum averaged over 10 subsequent EEG epochs of 6 s duration) but occasionally a reduction of more than 50% (Fig. 2D) compared to all other profiles was achieved especially at higher harmonics. This effect is further illustrated in Figure 2E showing a short EEG-epoch of 50 ms after application of the various weighting profiles. Again the advantage of the Becker is most prominent at high frequency components of the signal.

These results demonstrate that the Becker-profile achieves the most effective artifact suppression in terms of the residual spectral peaks at the harmonics of the gradient repetition rate, thus leading to less residual high frequency activity after artifact removal.

(iii) Artifact Resampling

The analyses using the simulated artifact template showed that a resampling with a temporal precision (regarding the alignment of the virtual samples) of at least 0.0002 ms is needed to avoid additional noise due to

Figure 5.

A: Standard deviation of the difference between artifact templates (phantom measurements) estimated at different times spaced by 20 TR (= 40 s) starting at GAR epoch 300 (i.e., the 15th volume = 30 s) of the first run, each extracted from 50

subsequent artifact epochs recorded in the 3T TRIO MR scanner. **B:** Same as A, but measured in the 1.5 T GE MR scanner. **C:** Difference waveforms of artifacts obtained at the 3.0 T MRI.

misaligned artifact sampling (refer supplementary materials for details). In principle this procedure is comparable to upsampling of the GAR at 5 MHz, aligning it to the reference artifact, and subsequently downsampling, while avoiding the boost in data volume and computational load.

Figure 6A shows an extreme example to illustrate (in the time domain) how much reduction in residual artifact activity can be achieved by replacing the upsampling by the resampling algorithm. Visually, the improvement is most prominent for high frequency components exceeding 100 Hz, but the signal's power spectrum is an average over ten 6-s-epochs recorded at electrode position Fp1 of one subject. Figure 6B demonstrates an artifact reduction that extends over all harmonics of the gradient repetition frequency (10 Hz in this case).

Besides an improved artifact removal, we observed a reduction of computing time by a factor of 4 with resam-

pling compared to upsampling to 50 KHZ, aligning, and downsampling. In both cases interpolation was done by means of a cubic spline.

(iv) Comparison of Slice Selective Versus Uniform Artifact Template

The standard deviation of the difference amplitude between the various slice selective artifacts and the template observed at the first slice fluctuates by about 1 μV, and is thus comparable with the inherent system noise level. Consequently, the slice selective artifact templates might only marginally improve the artifact removal. Moreover, bearing in mind the temporal fluctuations of the template, the larger period needed to gather a sufficient number of artifacts for averaging would likely counteract

Figure 6.

A: Example (real EEG) illustrating to what extent the residual artifact activity can be reduced by replacing the upsampling (10-fold) by the resampling algorithm. **B:** Same effect shown in the frequency domain. The power spectra represent the average over all subsequent 6 s epochs of a 60 s EEG segment recorded at electrode Fp1 in subject I.

this improvement. Therefore, we implemented the algorithm using a uniform artifact template.

(v) Accounting for Head Movements

The change of the artifact template observed in the watermelon data due to the displacement of the P4 electrode (refer Fig. 1C) after slight rotation of the melon is illustrated in Figure 7. The differences between the various templates clearly exceed the typical EEG amplitude range and thus justify the modification of the MAS algorithm towards a movement-dependent version.

To illustrate the effect of the movement aligned correction we applied the MAS and the MAMAS algorithm to a representative 6 s human EEG epoch with a clear (>0.4 mm) displacement in y-direction during this epoch according to the SPM movement detection. The SPM-displacement trace is shown in Figure 8A, the EEG epoch resulting from the application of MAS and MAMAS is displayed in Figure 8B and the corresponding power spectra are shown in Figure 8C.

Comparing the MAMAS and the MAS result, the time domain representation shows a prominent reduction of residual artifactual activity with the MAMAS algorithm, especially at higher frequencies. Correspondingly, the peaks at the harmonics of the gradient repetition rate are reduced. Figure 8D shows that EEG signals recorded outside the MR scanner (i.e., without gradient artefact) are only marginally affected by the MAS and the MAMAS processing. This means that the artefact reduction provided by the two algorithms leaves the original EEG unchanged.

Accordingly, the RINPS values indicate the improvement achieved by the MAMAS compared to the MAS-approach is in the range of 20–50% (refer Fig. 9). This improvement can exclusively be attributed to the movement related mechanism of MAMAS since both MAS and MAMAS were applied with resampling rather than upsampling. It is slightly larger for the high frequency fraction of the artefact activity (>30 Hz) and is more pronounced for periods with strong movements.

Figure 7.

GEAR template (phantom measurements) observed at electrode P4 at the different phantom-angles specified in Figure 1C.

Figure 8.

Effect of the MAS and the MAMAS algorithm to remove GARS. **A:** SPM position information (vertical/y direction). **B:** Resulting EEG after application of MAS and MAMAS, respectively. **C:** Resulting spectrum (10 subsequent epochs of 6 s duration) before and after artifact removal. **D:** Spectra derived from an EEG recorded outside the MR scanner (i.e., free of GAR). The different colors represent different processing steps (no proc-

essing, MAS processing, MAMAS processing). Each spectrum was estimated as the average over 40 subsequent epochs of 6 s duration starting at the very beginning of the recording, average over all channels. Left: Frequency range, 0–100 Hz. Right: Frequency range, 30–100 Hz, amplitude scale adapted to illustrate the concordance of the three spectra at these higher frequencies.

(vi) Reducing Residual Artifactual Activity Occurring in Case of Transient Brief Movements

After application of the additional algorithm, the spectral power at harmonics of the gradient repetition rate are clearly reduced but still not perfectly removed. Therefore, as a final step, we applied a low pass filter (for details refer the supplementary materials). This means, in contrast to the standard EEG (i.e., without instantaneous movement), a sufficient artifact suppression is only achieved for a limited bandwidth.

(vii) Practical Demonstration of the Contribution of the Various Processing Steps Including Removal of Vibration Artifact

The result of the various processing steps including the suppression of the vibration artifact is shown in Figure

10A–E. The traces obtained after artifact suppression but before low pass filtering illustrates the improvement of the MAMAS over the MAS algorithm. After vibration, artifact removal and low pass filtering with 120 Hz no artifact residuals are visible in the MAMAS result. Note that the EEG shown in Figure 10B was processed with the MAS algorithm using the resampling technique (rather than the 10-fold upsampling as originally proposed by Becker et al., [2005]). Consequently, the improved artefact removal provided by the MAMAS algorithm as compared to MAS (Fig. 10C vs. Fig. 10B) can exclusively be attributed to the movement dependent algorithm.

DISCUSSION

The aim of the current work was to optimize the subtraction procedures used to remove MR switching artifacts

Figure 9.

Relative reduction of spectral power at the gradient repetition rate and its harmonics achieved with MAMAS in comparison with MAS, averaged over all 6 subjects. The reduction factor was computed by restriction of the RINPS measure to the bandwidth specified in each of the subfigures. In total, 214 epochs

with mild movement and 87 epochs with strong movement were included. Each pair of bar represents the improvement observed at the electrode labeled on the abscissa. Subfigures **A**, **B**, and **C** illustrate the improvement for different frequency bands as indicated by the inserts.

Figure 10.

A–E: Example of a subset of all 29 EEG channels after the different processing steps (refer inserts for details). [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

from EEG recordings made during MR scans (refer to Becker et al. [2005] and Herrmann and Debener [2008]). The residual artifacts may occur due to both technical instabilities in the EEG-MR-setting and subjects' head movements. The contribution of each of these error sources cannot be differentiated in physiological EEG recordings because subtle head movements can in general not be

avoided, mixing the effects of the two sources of artifact. Therefore, we conducted phantom measurements with electrodes applied to a watermelon that was anchored at a fixed position in the scanner at four different orientations (leading to a different orientation of the electrodes and the cables with respect to the gradient magnetic field). To simulate a kind of worst case scenario we rotated the

watermelon by 2° steps (up to a total of 8°, which is the likely upper limit of realistic movements). The amplitude differences between the artifact templates observed at different orientations were surprisingly high and reached up to about 10% of the absolute artifact amplitude, underscoring the need for taking these movements into account.

However, even more surprising the phantom measurements demonstrated that the shape of the artifact template changed over periods of only a few seconds (refer Fig. 5A,B) in the absence of clear movement related fluctuations. Although this variability is much lower than the movement related changes in the artifact waveform, this observation could explain the fact that among the evaluated weighting functions, the Becker version, which focuses on those epochs positioned closest to the artifact to be removed, proved the most effective. The mechanisms underlying these changes remain unclear, but one can speculate that the warming of the gradient coils leads to very small changes in the magnetic field, giving rise to tiny but not negligible variations of the artifact shape which are not detected and thus not corrected by the MAMAS algorithm. Since a similar instability was observed on a 1.5T scanner constructed by a different company, the instability does not appear to be specific to this particular scanner or type of scanner.

Together, these observations (i) justify the extraction of the artifact template by a moving average algorithm and (ii) indicate the need for a head position dependent averaging algorithm. To implement the latter we used the movement vector provided by the SPM-program in the context of functional MR analysis. Although this information is available only every half volume, its temporal resolution is sufficient except in the case of unusually fast movements. Such fast movements can be traced correctly by a direct position tracking system. We applied such a video based device (resolution 20 ms) to identify momentary fast movements as a source of artifact template distortions that cannot be adequately removed by the standard subtraction method. However, this tracking tool is usually not available in standard MRI-units. Therefore, we declined to include this device as an essential part of the artifact removal scheme although we are aware that such short term fluctuations, even if rare, may lead to distortions of the artifact template and thus compromise the corrected signal. MAMAS in principle could easily be modified to include this high resolution movement detector thereby providing a more efficient GAR removal.

Restricting the extraction of the subtracted average to a limited number of epochs in the vicinity of the actual artifact conflicts with the need to keep the residual EEG noise below a level which can only be achieved by averaging over a sufficiently large number of epochs (taking into account that the EEG is not correlated with the gradient timing). Our analysis of EEGs recorded outside the scanner indicates this constraint is met by the exponentially decaying Becker window. In further tests (not presented here) we observed in cases of no or little movement that

an average over only 10 epochs leads to an artifact reduction that is comparable (both by visual inspection and by spectral measures) to the Becker-algorithm. However, the potential distortion of the ongoing background EEG due to residual EEG activity carried by the artifact template cannot be controlled in this case. Thus, single trial or continuous EEG analyses would be degraded by residual activity from other epochs.

Various methods have been proposed to cope with the extremely steep slopes of the artifact templates. Becker et al. [2005], Allen et al. [2000], and Niazy et al. [2005] increased the sampling rate by a fixed interpolation up to 50 KHz (Becker), 100 KHz (Allen), and 20 KHz (Niazy) to capture the slope and the peak with sufficient precision. However, according to our analysis a much higher rate of about 5 MHz would be needed to keep the residual error (due to an imperfect match between the EEG sampling times and the gradient timing) below the noise floor. The method presented in this article avoids this extreme boost of data by adapting the sampling times for individual epochs in response to the artifact that is to be removed. Recently, Goncalves et al. [2007a,b] presented an alternative approach that avoids any mismatch between the sampling grid applied to the different artifact epochs: Before any processing they precisely determined the exact interslice interval (ISI) in terms of the EEG sampling interval. From this they calculate the number of samples per slice plus a fraction of the EEG sampling interval, thereby specifying the difference between the exact ISI and the nearest integer number of EEG samples. These authors show this approach (which has an effect similar to that of a recent commercial device that electronically locks the clocks of the MR- and the EEG systems) leads to a perfect temporal match of the samples representing the artifact epochs. However, this only works under the assumption of a perfectly stable ISI. In practice, in our fMRI experiments we observed a spread of about 5 μ s in the ISIs observed over the full fMRI measurement period (quantified by determining the GAR's onset slope after increasing the sampling by means of interpolation beyond 1 MHz), which clearly exceeds the critical threshold determined for our resampling scheme. The epoch specific resampling is quite capable of tracking even this tiny jitter to minimize this source of residual artifact activity. Because of the optimization of the resampling algorithm (refer above), this process is numerically more efficient than the 10-fold upsampling (used by Allen et al. [2000]). To further reduce the computational load of the complete artifact removal algorithm one could apply an appropriate low pass filter and then reduce the sampling rate prior to further processing. However, the goal here was to optimize the removal of artifacts from EEGs made with EEG recorders that used their standard settings.

Especially during patient measurements, substantial head movements often cannot be avoided. Therefore, clinical EEG-fMRI-measurements will probably derive the greatest benefit from the improvement achieved with our

modified method. In the case of EEG-guided MR imaging (for instance spike triggered fMRI in epileptic patients), a real time artifact removal is required. The algorithm described here is non causal and thus not suitable for this purpose. However, in case of movements that are not too fast a causal variant of this method that relies on past epochs should be sufficient. In support of this supposition, we have observed that the performance of the current version of the algorithm does not change if the position vector being temporally assigned to the center of a current half volume is assigned to the end of that half volume. As part of a "real time movement correction" procedure most MR scanners continuously monitor head movements, using a method similar to that used by SPM, and thus provide the position information needed for the artifact removal. Furthermore, a direct head movement monitor with a much better temporal precision that uses an external optical motion tracking has recently been described [Speck et al., 2006]. This device could be used for an improved real-time movement corrected subtraction algorithm.

In case of brief strong movements, sufficient samples of the artifact are not available to allow the derivation of the reference template because (i) the sequence of artifacts is too short and (ii) each artifact has its own shape and is thus not suitable for the subtraction process. Therefore, we tried to extract a representative template by averaging over channels rather than over time. This approach produces a clear reduction of the artifact amplitude, which is sufficient for a coarse visual EEG evaluation but certainly not for a quantitative analysis. However, the remaining residual artifact is still large compared to signal periods without a strong movement. Since these special cases have a different shape each time they occur a more efficient method of removal does not appear to be feasible. Fortunately, these special cases occur infrequently and in some subjects not at all. As a rough estimate typically up to 10 such events (duration < 1 s/event) occurred during a 600 s recording period summing up to less than 2% of the total signal.

Given the successful application of independent component analysis (ICA) based methods to the removal, other EEG artifacts such as the EOG or BCG [refer Mantini et al., 2007; Tong et al., 2001], the question arises if blind separation methods could replace the subtraction method. Several groups [Mantini et al., 2007; Niazy et al., 2005; Schmid et al., 2006] have applied ICA or PCA to reduce residual artifacts after application of the subtraction method. However, all these methods require the determination of the number of components to include in the artifact processing. Although this task is well suited to an interactive procedure, it is less suitable for a fully automatic processing system. Therefore, we chose not to use this method.

Looking at the improvement observed at higher frequencies after MAMAS processing, one might ask if this attenuation could be due to a plain filtering effect. However, the results obtained (refer Fig. 8D) after processing of

an artifact free EEG recorded outside the scanner argue against this mechanism.

With respect to the Gamma-EEG frequency band (i.e., beyond about 30 Hz), a question that arises is the extent to which EEG activity at these frequencies is suitable for a valid quantitative analysis after artifact removal. If the strength of the artifact activity is stable over some time, its residuum after the removal process might be assumed to be stable as well, and thus just add a constant to the total power in this frequency range. If this stability cannot be assumed, only the spectral components that fall between the harmonics of the gradient repetition rate should be included in the analysis. But even in this case the validity of the results would benefit from the improved removal algorithm because the reduced residual harmonics lead to a lower spread of the spectral power of the artifact specific components. A second problem is produced by the periodic artifact generated by the vacuum pump of the MR scanner ("vibration artifact"). We observed this in the 3-T-MR scanner used in the current investigation but it is much weaker in our lab's 1.5T-GE scanner. This component with its frequency fluctuating slightly between about 45 and 50 Hz can only be suppressed by an appropriate band-pass filter. Alternatively, the cryocooling system could be switched off for brief recording periods. The processing sequence, i.e., first GAR removal and then vibration suppression versus the inverse order, has no influence on the artifact removal performance.

One can only speculate about possible reasons for the imperfect artifact removal we observed. Given the discrepancy between the almost perfect resampling we observed with the simulated artifact and the residual amplitudes after resampling and removal of the real artifact, it could be the case that very low level noise or a limited reproducibility of the gradient's time course contributes to the high frequency residual which are pronounced close to the steep slopes of the artifact gradient. In addition, very small head movements (for instance, head nodding caused by the cardiac cycle) which cannot be detected by the movement monitor could nevertheless cause a slight deformation of the artifact which is not captured by the average template (refer above) used for subtraction. Finally, although we did not observe severe differences in the removal performance for different interpolation schemes including a SINC-interpolation [$\sin(x)/x$ as proposed by Allen et al. 2000], we cannot control for potential errors around the artifact peak introduced by the interpolation algorithm since the true shape of the GAR is not known in closed form.

SUMMARY

In the present study, we (i) determined the technically optimal parameters for a subtraction based GAR removal and (ii) improved this method by including the current head position as a parameter controlling the artifact epochs used to extract the actual reference artifact

template. The result is a clear improvement in the visual quality of the ongoing EEG, and an improvement of up to 50% with quantitative power spectral measures, especially at higher frequencies.

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