Magnetic Resonance Imaging, Vol. 17, No. 9, pp. 1383–1391, 1999

Printed in the USA.
0730-725X/99 $–see front matter

ELSEVIER

PII S0730-725X(99)00096-X

● *Technical Note*

RESTORATION OF MR-INDUCED ARTIFACTS IN SIMULTANEOUSLY RECORDED MR/EEG DATA

J. SIJBERS,* I. MICHIELS,† M. VERHOYE,† J. VAN AUDEKERKE,† A. VAN DER LINDEN,† AND D. VAN DYCK*

*Vision Lab, University of Antwerp, and †Bio-Imaging Lab, University of Antwerp, Belgium

During a Magnetic Resonance sequence, simultaneously acquired ElectroEncephaloGraphy (EEG) data are compromised by severe pollution due to artifacts originating from the switching of the magnetic field gradients. In this work, it is shown how these artifacts can be strongly reduced or even removed through application of an adaptive artifact restoration scheme. The method has proved to be fully automatic and to retain high frequency EEG information, which is indispensable for many EEG applications. © 1999 Elsevier Science Inc.

Keywords: **Restoration; Simultaneous EEG/MRI; Signal-to-artifact ratio.**

INTRODUCTION

Over the past few years, the interest in combining ElectroEncephaloGraphy (EEG) and Magnetic Resonance Imaging (MRI) simultaneously has grown considerably. Due to their complementary strengths, the combination of the two techniques is able to render a new and powerful tool in functional brain research. With functional MRI (fMRI), the localization of activated areas can be determined with very high accuracy. The time resolution of fMRI, however, is rather poor and thereby it is not adequate to fully explore the complex dynamics of neural processes. In contrast to fMRI, the time resolution of EEG is excellent but the resolution of spatial source allocation is low. Therefore, combining these two modalities is helpful in the study of the brain.

However, EEG and MRI severely interfere during simultaneous acquisitions. With some precautions taken, it has shown possible to obtain artifact free MR images.[1–4] Acquiring an artifact-free EEG in the MR environment, though, is not straightforward. Many interferences with the MRI system result in pollution of the EEG recordings:[1,5–7]

- Static magnetic field interference: movement in a magnetic field changes the electromagnetic flux through for example an inter-electrode loop or even the subject from which electrophysiological readings are taken, resulting in an unwanted contributions to the EEG recordings. Although it poses practical difficulties, the interference due to movement can simply be avoided by fixing the subject and wires and avoiding loops in the wires.
- Cardiac pulse interference: Although the true basis of this artifact is not fully distinguished, it is most probable a combination of
  - small cardiac related movements of the body,[7–9]
  - small but firm movement of the electrodes and scalp due to expansion and contraction of scalp arteries between systolic and diastolic phase.[8] Ives suggested these to originate from the acceleration and abrupt directional change of blood flow in the aortic arch.[1]
  - fluctuation of the Hall-voltage due to the pulsatile speed changes of the blood in the arteries.[7,8]

Without precaution, the cardiac pulse generates artifacts with a standard deviation considerably larger than the EEG variations.[7] These artifacts, however, can significantly be reduced by firmly bandaging the electrodes and wires to the subject.[1,8]

RECEIVED 6/4/99; ACCEPTED 8/7/99.
Address correspondence to Jan Sijbers, Groenenborgerlaan 171, B-2020 Antwerp, Belgium. Tel: +32 (0)3 218 0 452; Fax: +32 (0)3 218 0 318; E-mail: sijbers@ruca.ua.ac.be

- Dynamic magnetic field interference: By far the most important interference originates from switching magnetic field gradients during an MR imaging sequence. Gradient switching severely obscures the EEG. In fact, gradient related artifacts make the reading of the EEG signal almost impossible during the actual MR imaging. It poses one of the main problems in simultaneous EEG/MR recording.

In this work, we focus on the problem of restoring EEG signal corrupted by MR gradient-induced artifacts. Conventional low-pass filters are able to reduce those artifacts significantly. However, along with the artifacts, a significant amount of underlying EEG information (i.e., all frequencies above the cut-off frequency) is removed. Hence, there is a need of developing filters that are able to reduce the artifact signal while retaining the high frequency information of the relevant EEG signal. In that context, we could only trace one very recent journal publication, where the restoration of ECG signals distorted by inductive effects of magnetic field gradients has been discussed.[10] The authors exploit the output of the MR gradient shapes, digitized with a sampling rate of 4 kHz along with an estimation of the gradient impulse response functions to restore the ECG signals. In this work, we propose an alternative restoration method, based on a simple experimental setup that only relies on 4 binary channels carrying codes representing the MR imaging scheme. The method was applied for the restoration of MR gradient corrupted EEG data.

ADAPTIVE RESTORATION SCHEME

In the following, the raw EEG signal is denoted by $f(t)$. This signal is assumed to be constituted of:

1. a term $s(t)$ containing the encephalographic information but also all non MR gradient related artifacts such as movement artifacts or static magnetic field artifacts. The disturbances present in $s(t)$ are generally much smaller and/or occur less frequent compared to MR gradient related artifacts and do not render the EEG signal unreadable. In the following, restoration refers to the restoration of $s(t)$.
2. artifacts $\{a_g(t)\}$ originating from MR gradient switching. In this work, we concentrate only on the removal of these kinds of artifacts. It was postulated and experimentally verified by Felblinger et al. that the gradient impulse response functions can be modeled as a linear time invariant system.[10] Hence, it is reasonably assumed that these artifacts are additive, such that they can be removed independent from other artifacts.
3. a white, gaussian distributed, and hence additive noise term $n(t)$.

In summary:

$$f(t) = s(t) + \sum_g a_g(t) + n(t) \tag{1}$$

The restoration scheme presented below consists of extraction of the artifacts, estimation of the artifact template, and adaptive filtering.

Extraction of the Artifacts

In general, artifacts originating from the switching of magnetic field gradients depend on the slope and magnitude of the gradients, the orientation of the electrodes with respect to the gradient direction, etc. Hence, it is clear that detection of gradient related artifacts in the EEG signal is a rather complex task.

An elegant solution to this problem is encoding the MR imaging sequence along with the EEG signals. Therefore, an additional link has been set up between the MRI- and the EEG-computer. The electrical signals at the output userport were converted to an optical signal, carried to the EEG-computer by fiberglass cables, reconverted to electrical signals, and fed to the EEG-computer. The MRI software was adapted so that a 4-bit-code for the current status of the MRI system was sent through the link to the EEG-computer. In that way, the EEG signals were recorded along with 4 binary channels carrying the MR codes: 1) on/off state of the imaging sequence; 2) the gradients along the slice; 3) phase encoding; and 4) frequency encoding direction. Owing to the MR codes, which were stored along with EEG signals, detection of gradient related artifacts now simplifies to detection of the MR codes related to the gradient switching. Obviously, this is far easier than searching for those artifacts in the raw EEG signals.

An MR imaging sequence consists of a repetitive switching of gradients. Hence, the artifacts appear as heavy disturbances at regular intervals. Because of the linear time invariant model, the distortion of the EEG signal will vanish as soon as the gradient switching stops. In general, when the gradient switching is halted, the artifact contributions become negligible after about 100 ms. For slow imaging sequences, where the distortions are vanished before the next artifact appears, the selection interval was chosen to be the distance between two subsequent code blocks. In case of fast imaging sequences, however, all artifacts merge into each other, resulting in large, heavily disturbed EEG channels. In that case, for each gradient code block, an interval of the EEG signal is selected enclosing multiple gradient code

blocks. In our experimental setup, the selection interval ranged from 150 to 300 ms. We remark here that only the code block distance (and hence the repetition time and the number of slices) is used by the restoration method.

Estimation of the Artifact Template

For each selected interval, after normalization, the power spectrum is computed. Each power spectrum, of course, contains the frequency contribution of $s(t)$, $n(t)$, as well as of the artifacts $\{a_g(t)\}$. The power spectra of L such intervals are stored into the rows of a matrix $P(L, u)$. As to retain only the artifact frequencies from the Fourier spectra, for each frequency u (i.e., for each column of the matrix P), a median filter is applied (Fig. 1). Because only the artifact frequencies contribute coherently to the median filtered spectrum, the result is a good approximation of the power spectrum of the artifact template. This template will be represented by $T(u)$. In our experimental setup, L ranged from 15 for slow MR sequences to 31 for faster ones.

Adaptive Filtering

The template function T was then adapted to each artifact power spectrum by minimizing the difference (MSE) with respect to the parameter b as given by:

$$\text{MSE} = \sum_u [F(u) - bT(u)]^2 \qquad (2)$$

Using the optimized b-parameter, the template function is subtracted from the artifact power spectrum and inverse Fourier transformed. Each time a new artifact interval is selected, its power spectrum is computed and stored in the next row of the matrix P. In this way, the artifact template $T(u)$ is updated adaptively while processing the EEG channels. The whole procedure is completed for each channel of the EEG data.

MATERIALS AND METHODS

Subjects

Tests were performed on adult Whistar rats and electrophysiological inactive phantoms, respectively. The rats were anesthetized with a mixture of 87.5% Ketalar (Ketaminium and Benzethonii chloride, Warner-Lambert manufacturing Ltd, Dublin, Ireland) and 12.5% Rompun (Xylazin. Hydrochlorid, Sodium chloride, Sodium Bicarbonate, and Methyl. Parahydroxybenzene, Bayer AG, Leverkusen, Germany). An initial intramuscular dose of 1.33 mL/kg was used and a maintenance dose of 0.33 mL/kg was administered through a subcutane catheter every 30 min. Body temperature was kept at 38°C by a heating pad (T/Pad, Gaymar Institute, Kent Scientific Corporation, Litchfield, CT, USA) and heating system fed back by a rectal thermometer (T/Pump, Gaymar Institute, Kent Scientific Corporation, Litchfield, CT, USA). The rat's head was immobilized by a custom made stereotactic device with earplugs and mouthpiece.

In our measurements, no pulse artifacts were observed. To our opinion, it is a result of the immobilization of the rat with anaestetics and stereotactic fixation. Also, less expansion occurs with the smaller veins during the systolic phase, hence leading to smaller movements of the electrodes in the magnetic field.

EEG/MRI Hardware

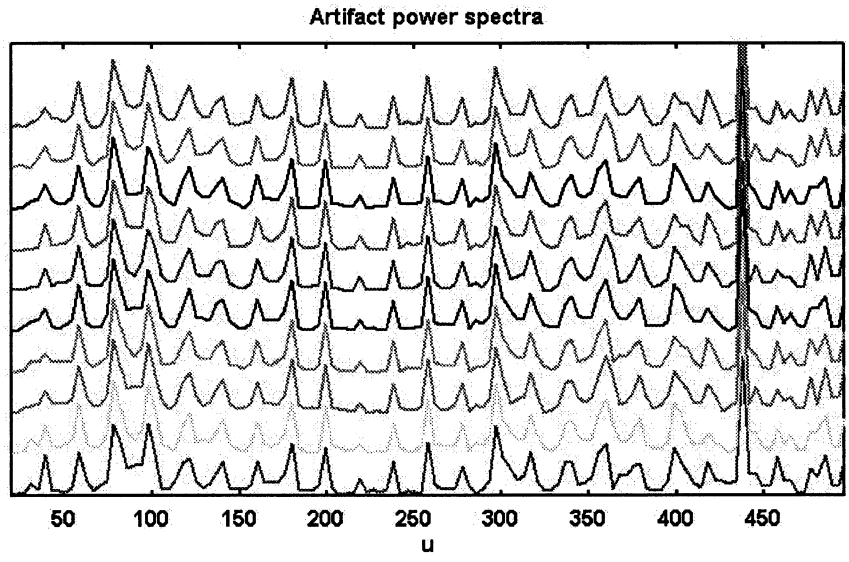
The experiments were done on a 7T MRI system (SMIS, Guildford, UK) with a 80-mm aperture and self-shielded gradients. Images where taken with size 256×128, maximum gradient strengths $G_r = 0.017$ T/m, $G_p = 0.027$ T/m, $G_{sl} = 0.07$ T/m, and ramp time 100 μs.

The EEG data were collected with a nonmagnetic preamplifier (Schwarzer, Munich, Germany) through five Ag/AgCl sintered electrodes (Schwarzer, Munich, Germany) with a sample rate of 1 kHz. The electrodes were slightly modified to fit the small head. They were applied with EEG paste (EEGsol, Graphic Controls, Buffalo, NY, USA) and held on the skin by an elastic tubular net bandage (Bandafix, Zutphen, the Netherlands) around the rat's head. The electrode leads—0.9 mm thin and 85 cm long coax cables—were bundled together and firmly fixed. Precaution was taken not to put any loops in the leads. Next, they were attached to the inputs of the EEG preamplifier just outside the magnet's bore. The amplified signal was then carried over a fiber-optic cable to the EEG-computer. The EEG-computer was equipped with special cards to operate with the EEG preamplifier (PTMS1 and PTMS3, Schwarzer, Munich, Germany). The software used for measuring, displaying, and analyzing the signal was Brainlab (OSG, Rumst, Belgium). The software for filtering the data was written on site under Visual C++ and has been integrated into the Brainlab software.

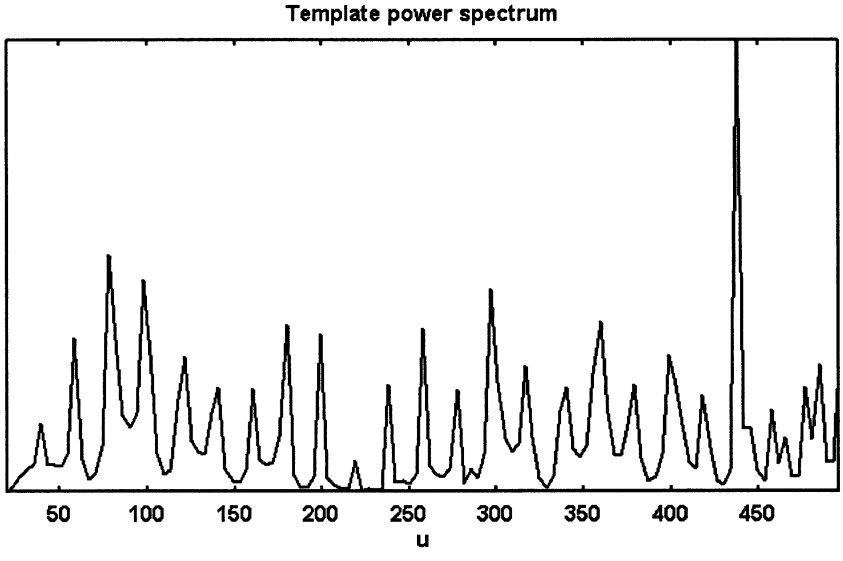
Test Procedure

As to validate the performance of the proposed restoration scheme, knowledge of the MR gradient artifact free EEG signal is required, because only in that case, the residual signal after restoration can be analyzed. As to accomplish this, a simultaneous EEG/MRI acquisition was made of a dead rat. In this way, the recording conditions were made identical compared to a normal experiment (impedance, electrode positions, etc.) while, of course, recording no EEG signal (or movement related artifacts) at all. Hence, only artifacts related to the MR gradient switching were recorded.

Various imaging sequences were applied such as Spin



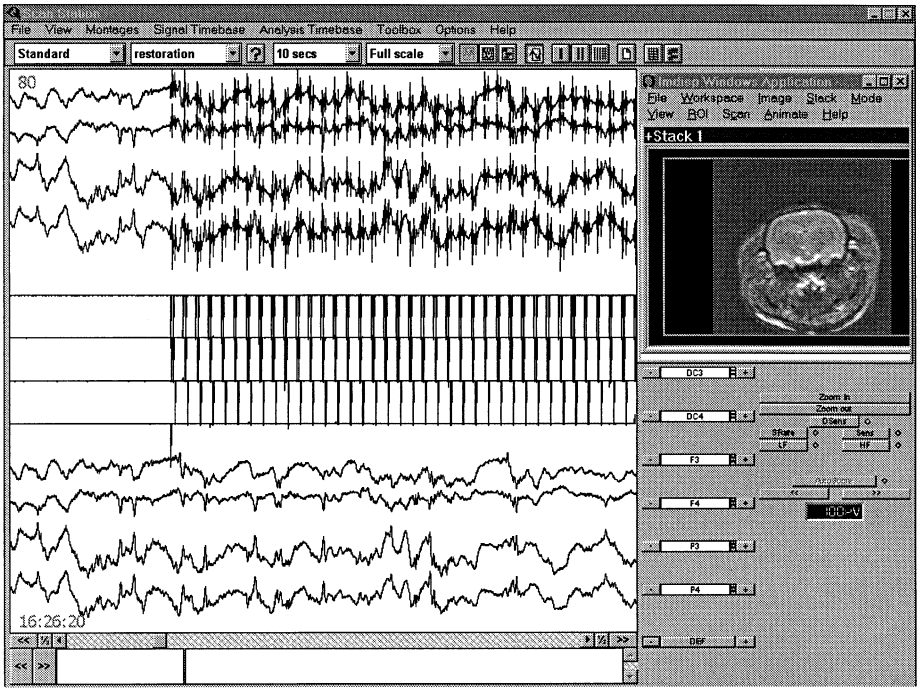
(a) Artifact power spectra



(b) Template function

Fig. 1. Estimation of an artifact template from subsequent artifact power spectra.

Echo (SE), Gradient Echo (GE), and diffusion-weighted (DW) imaging, with various echo (TE) and repetition times (TR). These recordings took 75 minutes in which 19 SE, 15 GE, and 9 DW multi-slice images were acquired, all with 10 images taken. The artifact recordings were then superimposed onto EEG signals from a living rat, which were acquired without MR imaging. Next, to these artificially corrupted data, the proposed filtering scheme was applied. As to quantify the restoration performance, the signal-to-artifact ratio (SAR) was computed before and after the restoration. Thereby, the SAR was defined as the ratio of the standard deviation of the



(a) SE (TE/TR = 100/2000 ms)

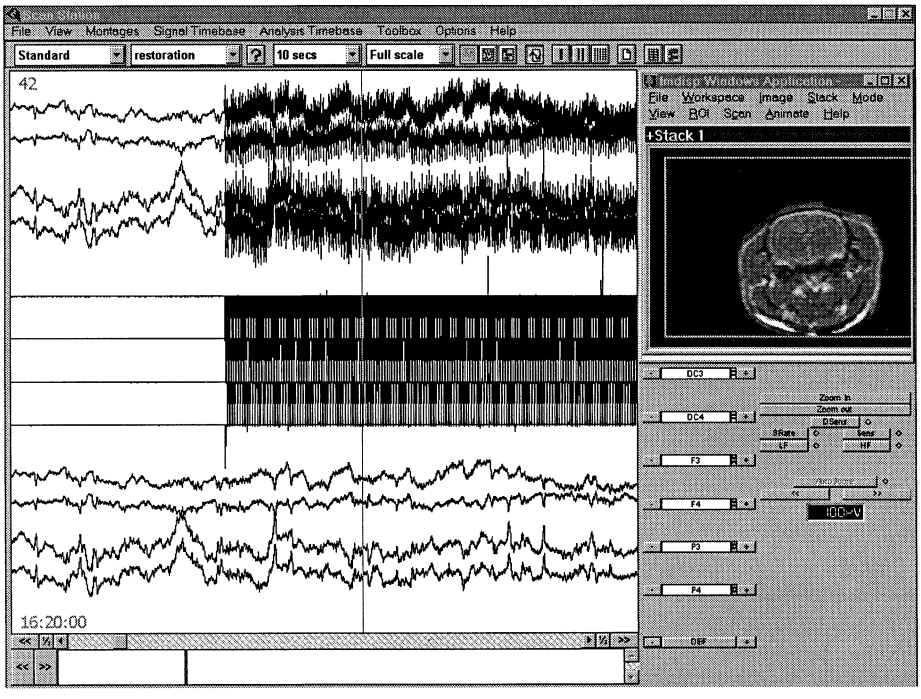


Fig. 2. Restoration results of MR gradient induced artifacts. The figure represents two multi slice SE sequences with 10 slices taken and TE/TR = 100/2000 ms (a) and 30/500 ms (b), respectively.

Table 1. SAR improvements as a result of the restoration procedure

SE (TE/TR)	SAR before	SAR after
SE (30/2000)	3.2	12.9
SE (30/1500)	2.7	12.3
SE (30/1000)	2.1	11.8
SE (30/500)	1.6	10.7
DE (30/1500)	2.6	11.2
DE (30/1200)	2.2	10.4
DE (30/1000)	2.1	8.9
DE (30/500)	1.4	5.6
DW (40/2000)	3.4	13.8
DW (40/1500)	3.0	11.7
DW (40/1000)	2.3	10.4
DW (40/500)	1.9	7.4

The signal-to-artifact ratio's before and after restoration for multi slice SE, GE, and DW sequence with various repetition times. The relative error was of the order of 2%.

artifact free signal and the standard deviation of the gradient related artifacts:

$$\mathrm{SAR} = \sqrt{\frac{\sigma_s^2}{\sigma_{\{a_g\}}}}, \qquad (3)$$

where the variance was computed over the total duration of the applied MR imaging sequence. For each imaging sequence, the SAR was computed before and after the restoration procedure.

RESULTS AND DISCUSSION

Figure 2 shows typical results of the proposed restoration scheme. The figure represents four raw EEG channels of the rat (each electrode versus the Goldman reference), followed by four DC channels carrying the binary codes from the MRI sequence. The start code of the MR imaging is clearly seen in these channels. As is obvious from the figure, the EEG recordings are heavily disturbed by the MR acquisition. The imaging sequences shown are two multi-slice SE sequences with TE/TR = 100/2000 and 30/500 ms, respectively. Since 10 images were taken during the acquisition, the distance between two artifacts (which also equals the distance between two gradient code blocks) is 200 and 50 ms, respectively.

The last four channels of Fig. 2(a) and 2(b) are the restored versions of the raw EEG channels. As can be observed, the artifacts induced by the switching MR gradients are strongly reduced while high frequency information is retained. In order to quantify the performance of the restoration scheme, the SAR was computed before and after the restoration. Table 1 shows the SAR improvements for a multi slice spin echo (SE), gradient echo (GE), and diffusion weighted SE sequence, all with various repetition times. The artifact contributions that were not removed by the restoration scheme are shown in Fig. 3. In that figure, the original, uncorrupted signals (first 4 channels), along with the residual signals after restoration (last 4 channels), are shown. For slow imaging sequences, the variance of the 'leftovers' was found to be orders of magnitude smaller than the variance of the original EEG signal. For fast imaging sequences, the variance of the residuals slightly increases. In general however, the SAR improvement due to the restoration scheme was found to be of the order of 4 to 5.

Beside improvement of the SAR, it is important that, during restoration, typical characteristics are retained in the EEG signals. The restoration scheme as described above indeed does retain high frequency information as can be appreciated from Fig. 4. In that figure, two EEG/MRI signal pages of an epileptic rat are shown, where a GE multi-slice imaging sequence was applied (TE/TR = 30/2000 ms, 10 slices). After restoration, epileptic features such as 3Hz activity as well as typical spike-wave complexes are well retained.

In summary, the proposed method is based on a simple experimental setup and does not require information from the MR gradients such as the shape, amplitude, rise-time, etc. Only knowledge of the on/off state is used. The echo time for example, though influencing the appearance of the artifact, is not relevant in the restoration scheme.

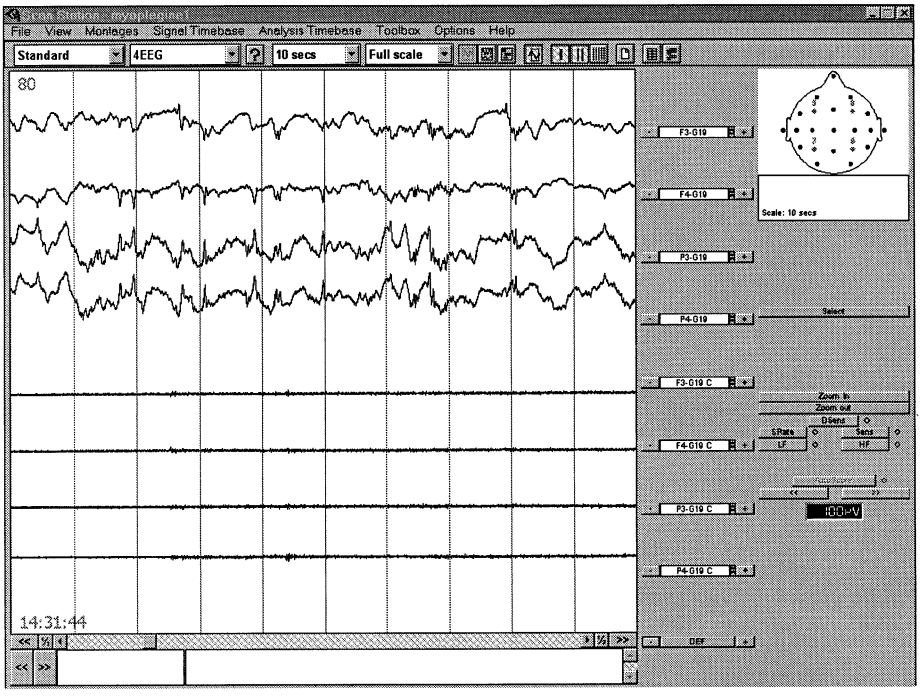
The method is invariant to small time shifts of the artifacts as it only relies on the power spectra of the artifacts. Also, owing to the continuous updating of the artifact power spectrum matrix, the method adapts itself to unforeseen changes of the gradient impulse response functions (e.g., when an electrode changes place, the appearance of the gradient artifact in the EEG recordings changes too).

In addition, the method has proved to be (relatively) computationally inexpensive. All data processing was performed on a 300 MHz PC. The 4-channel EEG recordings of 80 minutes, in which not less than 430 MR images were acquired, were processed within 15 minutes.

Finally, we remark that the filtering scheme as a whole is fully automatic, requiring no user-interaction at all.

CONCLUSIONS

Simultaneous EEG/MRI acquisitions suffer from severe pollution of EEG recordings, mainly due to the switching of the magnetic field gradients. It is clear that without handling this problem, interpretation of the EEG



(a) SE (TE/TR = 100/2000 ms)

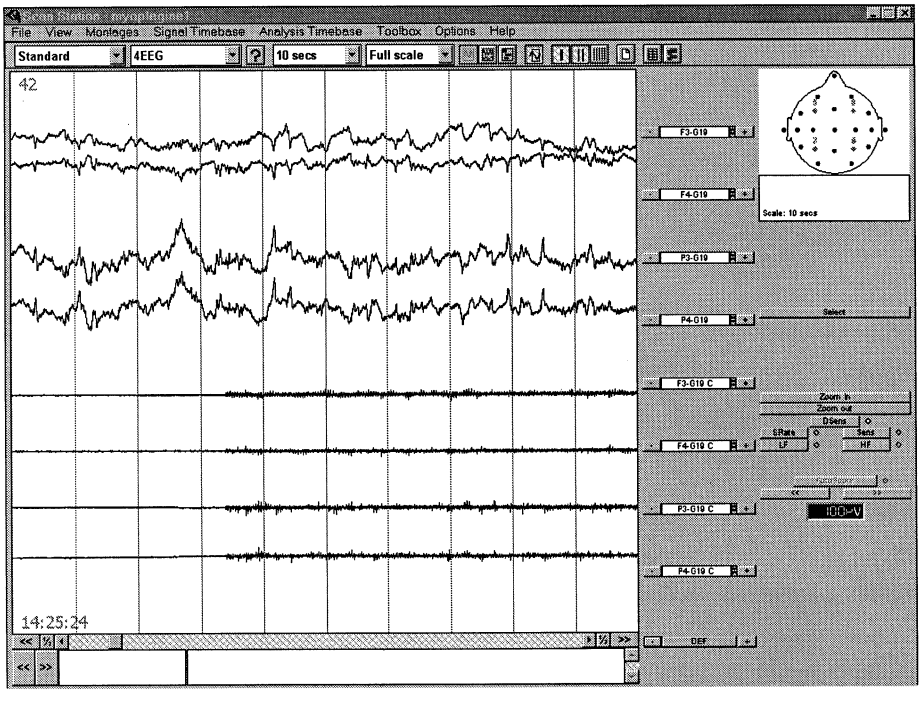


Fig. 3. The first 4 channels show the original, uncorrupted EEG signals (cfr. Fig. 2) whereas the next 4 channels represent the residual signals after restoration.

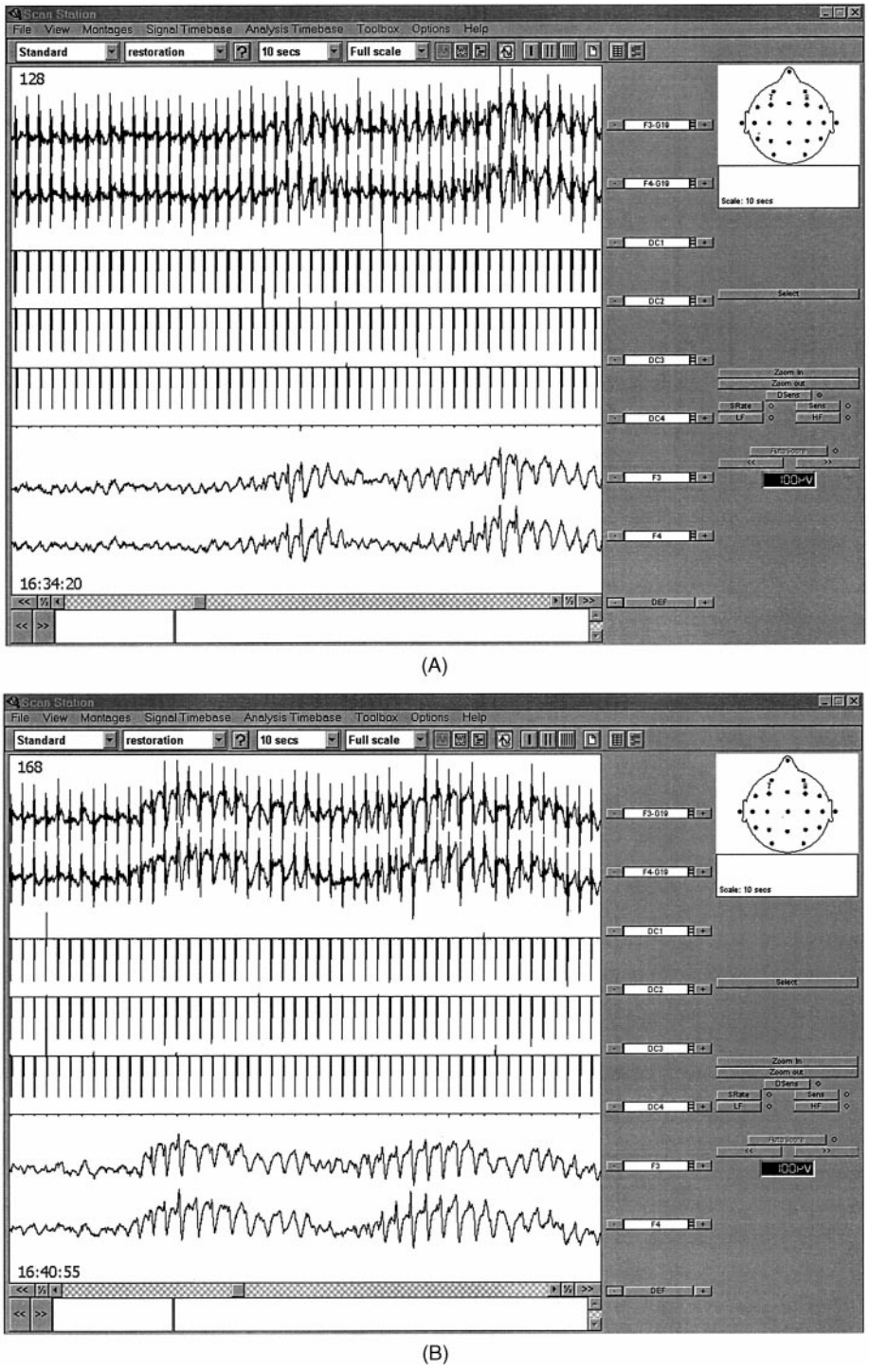


Fig. 4. Restoration results of MR gradient induced artifacts. The figure a simultaneous MRI/EEG acquisition of a rat showing epileptic characteristics. As can be observed, typical epileptic features are well retained after restoration.

signal is very difficult, if not impossible. In this work, we showed that, with properly encoding the MR imaging scheme into the EEG recordings, it is possible to construct a simple, robust, and highly efficient restoration scheme. The restoration scheme has proved to reduce the gradient related artifacts significantly during various MR acquisition schemes, while retaining high frequency information.

Acknowledgement—The work was financially supported by (grants of) the Flemish institute for the promotion of scientific and technological research in the industry (IWT), Brussels.

REFERENCES

1. Ives, J.R.; Warach, S.; Schmitt, F.; Edelman, R.R.; Schomer, D.L. Monitoring the patients' EEG during echo planar MRI. Clin. Neurol. 87:417–420; 1993.
2. Warach, S.; Ives, J.R.; Schlaug, G.; Patel, M.R.; Darby, D.G.; Thangaraj, V.; Edelman, R.R.; Schomer, D.L. EEG-triggered echo-planar functional MRI in epilepsy. Neurology 47:89–93; 1996.
3. Busch, E.; Hoehn-Berlage, M.; Eis, M.; Gyngell, M.L.; Hossman, K.A. Simultaneous recording of EEG, DC potential and diffusion-weighted NMR imaging during potassium induced cortical spreading depression in rats. NMR Biomed. 8:59–64; 1995.
4. Huang-Hellinger, F.R.; Heiter, H.C.; McGormack, G.; Cohen, M.S.; Kwong, K.K.; Sutton, J.P.; Savoy, R.L.; Weiskoff, R.M.; Davis, T.L.; Baker, J.R.; Belliveau, J.W.; Rosen, B.R. Simultaneous functional magnetic resonance imaging and electrophysiological recording. Human Brain Mapping 3:13–23; 1995.
5. Felblinger, J.; Müri, R.M.; Rösler, K.M.; Jung, B.; Hess, C.W.; Boesch, C. Misleading EEG recordings in the MR environment. In: EMRMB 14th Annual Meeting. Brussels, 1997: p. 108.
6. Krakow, K.; Allen, P.J.; Polizzi, G.; Lemieux, L.; Fish, D.R. The amplitude and distribution of EEG pulse artifact in the MR scanner: Its effect on the detection of EEG events. In: ISMRM 6th Scientific Meeting. Sydney, Australia, 1997: p. 1494.
7. Müri, R.M.; Felblinger, J.; Rösler, K.M.; Jung, B.; Hess, C.W.; Boesch, C. Recording of electrical brain activity in a magnetic resonance environment: distorting effects of the static magnetic field. Magn. Reson. Med. 39:18–22; 1998.
8. Allen, P.J.; Polizzi, G.; Krakow, K.; Fish, D.R.; Lemieux, L. Identification of EEG events in the MR scanner: The problem of pulse artifact and a method for its subtraction. NeuroImage 8:229–239; 1998.
9. Hill, R.A.; Chiappa, K.H.; Huang-Hellinger, F.; Jenkins, B.G. EEG during MR imaging: Differentation of movement artifact from paroxymal cortical activity. Neurology 45:1942–1943; 1995.
10. Felblinger, J.; Slotboom, J.; Kreis, R.; Jung, B.; Hess, C.W.; Boesch, C. Restoration of electrophysiological signals distorted by inductive effects of magnetic field gradients during MR sequences. Magn. Reson. Med. 41:715–721; 1999.