The Interaction between motor fatigue and cognitive task performance
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Surface EMG Measurements during fMRI at 3T: Accurate EMG recordings after artifact correction

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Abstract
In this experiment we have measured surface EMG of the first dorsal interosseus during predefined submaximal isometric contractions (5, 15, 30, 50, and 70% of maximal force) of the index finger simultaneously with fMRI measurements. Since we have used sparse sampling fMRI (3 seconds scanning; 2 seconds non-scanning), we were able to compare the mean amplitude of the undisturbed EMG (non-scanning) intervals with the mean amplitude of the EMG intervals during scanning, after MRI-artifact correction. The agreement between the mean amplitudes of the corrected and the undisturbed EMG was excellent and the mean difference between the two amplitudes was not significantly different. Furthermore, there was no significant difference between the corrected and undisturbed amplitude at different force levels. In conclusion, we have shown that it is feasible to record surface-EMG during scanning and that, after MRI artifact correction, the EMG recordings can be used to quantify isometric muscle activity, even at very low activation intensities.

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Introduction
Functional magnetic resonance imaging (fMRI) is a non-invasive method to study human brain function. Measuring muscle force and electromyography (EMG) concurrently with brain activity allows for a direct comparison of brain activity in motor areas and motor output. However, recording force and EMG in the MR-scanner is challenging. Ferrometals are attracted to the scanner. In addition, these metals disturb the magnetic field and can cause artifacts in the MR-images. Furthermore, the static magnetic field, the changing gradient, and the radio frequency pulses can cause artifacts in force and EMG-signals. Some studies have resolved this problem by repeating the experiment outside the scanner in combination with EMG recordings (e.g., Ehrsson et al., 2001; Maclntosh et al., 2004), expecting that the muscle activation patterns inside and outside the scanner are similar. A few studies have recorded EMG simultaneously with fMRI. However, because of the problems described above, the EMG-signals were only analyzed during the short inter-scan-intervals (varying between 100 and 200ms; Dai et al., 2001; Liu et al., 2000, 2002a, 2002b, 2003, 2004) or EMG recordings were only used to determine the end of muscle activity and no quantitative measurements were made (Oga et al., 2002; Toma et al., 1999).

Recently, techniques have been developed to study electroencephalography (EEG) and fMRI simultaneously. In those studies, the EEG recordings contain large MRI scanning artifacts. However, since the MRI artifact is almost constant over time, it is possible to subtract the mean artifact from the EEG signal. This results in reasonably clean EEG recordings after further pulse artifact correction (e.g., Allen et al., 2000; Anami et al., 2003). EEG and surface EMG recordings are based on similar principles; however, we expected more problems because of the relatively long EMG wires that will unavoidably move with muscle contraction. It has been shown that artifact correction can be applied to EMG measured simultaneously with fMRI (Moosmann et al., 2004). However, in that study, the EMG recordings were only used to separate periods of rest from periods of muscle activation. Quantitative data of the EMG recordings have not yet been published, nor was EMG activity quantified at different force levels in combination with scanning experiments.

Therefore, it was the aim of our study to evaluate the quantitative accuracy of the EMG signal at different force levels during scanning, after artifact correction according to Allen and colleagues (2000). For this reason, we have recorded EMG during a sparse sampling fMRI protocol. In this protocol scanning periods are alternated with non-scanning periods, which enabled us to compare the EMG recordings of sustained
muscle contractions at predefined force levels during scanning and non-scanning periods.

**Methods**

**EMG measurements**

We used sintered silver/silver-chloride EMG-electrodes (Falk Minow Services) in combination with the BrainAmp MR plus system of Brain Products GmbH (Munich, Germany) for the EMG-recordings. This system is normally used for EEG-recordings in the MRI, but a special electrode input box made it possible to connect EMG-electrodes. Current-limiting resistors (2kOhm) were attached to the EMG-electrodes in order to prevent possible warming of the electrodes (see Lemieux, 1997).

Muscle activity of first dorsal interosseus (FDI) of both hands was measured. Therefore, the skin above the FDI was cleaned with alcohol and scrubbed. One electrode was placed on the belly of the FDI and a second electrode on the metacarpophalangeal joint. A reference electrode was positioned on the processus styloideus radius of the right arm, and a ground electrode was placed on the processus styloideus radius of the left arm. The two electrode wires per muscle were twisted to maximally equalize the effect of the magnetic field on the EMG-recordings.

All electrodes were attached to the electrode input box, which was placed as far away from the scanner as possible (i.e., between the legs of the subject). The electrode input box was connected to the BrainAmp amplifier. The amplifier converted the neurophysiological signals to digital signals, which were transmitted via an optical cable to a PC outside the MR-room. The PC was equipped with Brain Vision Recorder software; the sampling rate of the signals was 5000 Hz/channel. The MRI-scanner sent a marker to Brain Vision Recorder every time a scan was started.

Offline, we used Brain Vision Analyzer software (version 1.05.0001) to correct the data for scanner artifacts according to the method described by Allen (Allen et al., 2000). In the MRI-Artifact Correction a low-pass filter and down sampling were included. Since we studied EMG recordings we used a low-pass filter of 400Hz, which implied that less of the scanner artifact was filtered out of the signal. The data was down sampled to 2500 Hz and, after artifact correction, bipolar derivations were calculated for each muscle. Thereafter, a high-pass filter of 10Hz was applied to remove possible movement artifacts. Finally, the data was exported (ASCII-format) and imported in Spike2 for Windows (version 5.08; Cambridge Electronic Design, Cambridge, UK) for further analysis.
**Force recordings**

The index finger abduction force was measured with a custom-made force transducer. This force transducer made it possible to measure force in the MRI-environment. Magnetic-compensated strain gauges (TML® MFLA-5.350-1L; Tokyo Sokki Kenkyujo Co., Japan), were placed in a full bridge configuration on an epoxy glass laminate bar (Tufnol® 10G/40 20mm diameter; RS-components number: 771-314), and registered the force applied to the bar. This resulted in electrical signals of a few millivolts, which were amplified and converted to digital signals. The digital signals left the scanner room via an optical cable. The optical cable was connected to a receiver in the operator room, by which the digital signals were adapted to voltage signals. These signals were sampled by a PC equipped with a data-acquisition interface and the accompanying software (1401 micro and Spike 2 version 5.08, Cambridge Electronic Design, Cambridge, UK). The sampling rate was 500 Hz.

**MRI**

Functional images were acquired using a 3T Philips MRI scanner (Best, the Netherlands), equipped with echo planar imaging (EPI) capability and a standard transient/receive (TR) head coil. The following pulse sequence parameters were used: FFE single shot EPI; 46 slices; slice thickness 3.5 mm; no gap; field of view 224 mm; scanning matrix 64x64; transverse slice orientation; repetition time (TR) = 5 s; echo time (TE) = 35 ms; minimal temporal slice timing (2884 ms); flip angle 90°.

In addition, T1-weighted anatomical images of the entire brain were obtained with the following pulse sequence parameters: field of view 256 mm; scanning matrix 256x256; 160 slices; slice thickness 1 mm; transverse slice orientation; TE = 4.6 ms; TR = 25 ms; flip angle 30°.

Additional data were acquired to compare the fMRI images with and without the presence of the EMG equipment. The pulse sequence parameters were the same as above (functional images), except that the repetition time was 3 seconds instead of 5 seconds.

**Tasks**

Before the experiment started, all subjects had practiced the tasks in the lab. The tasks consisted of several isometric index finger abductions. First, all subjects had to perform three maximal voluntary isometric contractions (MVCs) for ten seconds; each contraction was followed by 50 seconds rest. The contractions were performed for both left and right
hand separately. The highest force was determined to be the ‘control MVC level’ (cMVC; for left and right separately). Thereafter, subjects performed sub-maximal isometric contractions for 20 seconds at levels of 5, 15, 30, 50, and 70% cMVC; each contraction was followed by 40 seconds rest. All contractions were repeated three times in a semi-random order. To investigate the effect of the EMG recording on the fMRI images, additional scanning experiments were performed in which this series of contractions was repeated without EMG-recording. In this experiment, the first three series of contractions were performed with the EMG electrodes placed on the muscle, followed by three series of contractions without EMG montage. During all contractions subjects got visual feedback of the force that was projected by a beamer at the back-end of the scanner. Subjects were able to see this screen via a mirror on the head coil. During scanning the lights in the MR-room were off.

Statistics
To evaluate the accuracy of the EMG recordings after MRI-Artifact Correction, we determined the mean rectified EMG over 2 seconds during scanning (corrected-EMG) and non-scanning intervals (undisturbed-EMG) within a contraction. To correct for the physiological variation in EMG amplitudes during prolonged contractions, we calculated the average of the mean EMG amplitudes of the undisturbed EMG intervals preceding (see Fig. 1: B&D) and following (see Fig. 1: D&F) the corrected EMG interval (see Fig. 1: C&E). To analyze the difference between the corrected and undisturbed EMG values, we plotted the mean EMG of the corrected interval against the average of the undisturbed EMG intervals (see Fig. 1: C versus [(B+D)/2], and E versus [(D+F)/2]). As a measure of the physiological variation in EMG, we have calculated the mean value of the first (B) and third (F) undisturbed EMG-interval, and we have compared this mean to the second (D) undisturbed EMG-interval (see Fig. 1: D versus [(B+F)/2]). For the comparison between the undisturbed and corrected EMG, we have used the Bland and Altman method (1986). This method is based on a graphical analysis of the mean of the corrected and undisturbed EMG versus the difference between the corrected and undisturbed EMG. Limits of agreement (mean difference ± 2 standard deviations of the difference; Bland and Altman, 1986) were calculated for the comparison of corrected versus undisturbed EMG and also for the repeated measurements of undisturbed EMG within a contraction. Furthermore, the differences between corrected and undisturbed EMG values were determined by using an ANOVA for repeated measurements. Within-
subject factors were EMG (2 levels: corrected and undisturbed) and force (5 levels: 5, 15, 30, 50, and 70% cMVC).

In addition, we have performed a power-spectrum analysis of the corrected and undisturbed EMG. Therefore, we first segmented the data for the different force-levels in segments of 2048 data points, starting 500 ms (corrected) and 3500 ms (undisturbed) after the scanner marker. We used the Fast Fourier Transform to estimate the power ($\mu V^2$) of the segments (full spectrum used; Hanning window 10%). The spectra were stabilized by averaging over 10-12 segments per force level. Then, we calculated the median power for the different force levels.

![rs EMG schematic](image)

**Figure 1.** Schematic illustration of rectified and smoothed EMG (rsEMG) during a submaximal isometric contraction. The scanner markers show the start of the acquisition of a brain volume. The thick lines of the contraction (A, C, E) represent EMG-data that is collected during MR-scanning (3 s), while the thin lines (B, D, F) represent EMG-data that is collected during non-scanning intervals (2 s). The EMG-data during scanning is off-line corrected for the scanner artifact. Due to muscle fatigability, the mean amplitude of the EMG increases with time. To compare corrected with undisturbed EMG values we have, therefore, calculated the mean of intervals preceding (e.g. B) and following (D) a corrected interval and compared this value with the mean corrected EMG (C).

In addition, we have estimated the overall variance of the EMG by taking the average of the undisturbed intervals B & F and compared this average with the EMG value during the undisturbed interval D (see "Methods" for further details).

To account for the effect of the EMG equipment on the EPI-slices, we compared slices that were obtained in the presence and absence of EMG equipment for structural artifacts (Fig. 6) and functional artifacts (signal-to-noise-ratio). The signal-to-noise-ratio was determined on realigned data; the images were masked and within the mask the mean standard deviation over time (arbitrary units) was determined.
Results

In this report we focus on the EMG recordings to evaluate the accuracy of the MRI Artifact Correction method. Figure 2 shows a representative graph of the produced force in combination with the raw and rectified plus smoothed EMG-signal, both before and after MRI-artifact correction. Note the relatively clean EMG recordings after the artifact correction. Since our subjects have made contractions at different force levels we could assess different levels of EMG activity.

Figure 2. Force (A) and EMG (B-E) during contractions at different force levels. Panels B & D show the raw EMG data, panels C & E show rectified and smoothed (time-constant: 0.1 s) EMG recordings before (B & C) and after (D & E) MR-artifact correction. The task started with 2 dummy scans without markers. The MR-corrections was triggered by the MR-markers and started at the first dynamic san (t>48 s).

Correlation analysis

The correlation between the corrected and undisturbed EMG was strong and highly significant (mean $R^2=0.98$; all $p<0.001$ for both left and right hands (paired samples T-test); Fig. 3). However, a strong correlation does not tell us whether there is agreement
between the EMG values. There is only agreement when the slopes are close to 1 and the intercept close to 0. For the right hand the correlation is associated with slopes close to 1 (range: 0.97 to 1.03) and small intercept values (-0.0031 to +0.0067). For the left hand, the slopes are rather close to 1 (range: 1.06 to 1.13); the intercepts are more variable (range: -0.0797 to -0.0046).

To test whether the strong correlations also implied a good agreement, we visualized the agreement according to Bland and Altman (1986). Therefore, the average of the corrected and undisturbed EMG values was plotted against their difference (Fig. 4). A strong agreement is indicated by a mean difference that is close to 0. In our data the difference between the corrected and undisturbed EMG values did not deviate from 0 (one-sample T-test, \( t_{(5)}=0.953 \), n.s.). Thus, corrected and undisturbed EMG values showed a strong similarity. Furthermore, the ANOVA for repeated measurements revealed that there was no significant difference between the corrected and undisturbed EMG (\( F_{(1,5)}=0.78 \), n.s.). As expected, EMG values were significantly different for the different force levels (\( F_{(4,20)}=41.62, p=0.001 \)).

**EMG power spectrum**

The data of the power spectrum analysis showed no significant difference in total power between the corrected and undisturbed segments of EMG (\( F_{(1,2)}=3.63 \), n.s.). An example of the calculated power spectra for one force level (30% MVC) is shown in Fig. 5. In addition, no significant difference between the median power frequency of the corrected (68.1 ± 0.7) and undisturbed (66.1 ± 1.5) segments of EMG was found (\( F_{(1,2)}=4.92, \) n.s.).

**Brain activity**

The EMG equipment did not influence the EPI-images of the subjects as can be seen in Figure 6. Moreover, the standard deviation of the BOLD response recorded during contractions at different force levels did not deviate between measurements with or without EMG equipment (noise with equipment: 20.0 ± 2.7 arbitrary units; without equipment: 18.4 ± 0.9 a.u.; \( F_{(1,2)}=0.81, \) n.s.).
Figure 3. Relationships between corrected and undisturbed EMG recordings of three subjects for a right and left hand muscle (FDI). The mean amplitude of rectified EMG was determined for two second intervals of corrected EMG and undisturbed EMG intervals; see “Methods” and Fig. 1 for detailed description of the calculation of the mean values. In the graphs the linear regression line (solid line) and the identity line (dashed line) are plotted.
Figure 4. The mean ± 2SD of corrected and undisturbed EMG plotted against the difference between corrected and undisturbed EMG for each hand of three subjects (closed symbols and interrupted lines). The mean ± 2SD of the undisturbed EMG is plotted against the difference between the first and third versus the second interval of undisturbed EMG (open symbols and thin solid lines). The thick solid lines represent the mean of the differences between the corrected and undisturbed EMG. The asterisk is placed to clarify that the thin line is covered by the thick line.
Our data showed that it was feasible to perform surface EMG recordings during fMRI measurements. The EMG recordings were not free of MRI artifacts; however, it was possible to eliminate the MRI artifacts by using the Allen method (Allen et al., 2000) implemented in the Brain Vision Analyzer software. Moreover, our data showed that the corrected EMG recordings reflected muscle activity accurately during relatively weak contractions (5% cMVC) as well as during strong contractions (>70% cMVC).

The correlation between the corrected and the undisturbed EMG signals was excellent. In addition, the analysis according to Brand and Altman (1986) showed that the difference between the corrected and undisturbed EMG was smaller than the physiological variance in EMG amplitude. During prolonged contractions the central nervous system has to increase its drive to the spinal motoneurons to counteract a decline in force due to muscle fatigability (Bigland-Ritchie et al., 1986a, 1986b). This increased drive is shown in the EMG recordings as an increase in amplitude. Thus, during the production of a constant force a progressive increase in the EMG-amplitude

**Figure 5.** Typical example of a powerspectrum (μV²) for the undisturbed (solid line) and corrected (dashed line) EMG. The data were obtained during 30% MVC contractions and were averaged over three trials.

**Discussion**

Our data showed that it was feasible to perform surface EMG recordings during fMRI measurements. The EMG recordings were not free of MRI artifacts; however, it was possible to eliminate the MRI artifacts by using the Allen method (Allen et al., 2000) implemented in the Brain Vision Analyzer software. Moreover, our data showed that the corrected EMG recordings reflected muscle activity accurately during relatively weak contractions (5% cMVC) as well as during strong contractions (>70% cMVC).

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can be observed. Despite averaging EMG values of intervals preceding and following another interval, still variability was seen during a contraction. This physiological variance – determined by the EMG values of the undisturbed intervals – was larger than the variance induced by the artifact correction. Furthermore, the power spectra of the corrected and undisturbed EMG signals were in good agreement, and no effects on multiples of the slice frequency (16Hz) were found.

![Figure 6](image)

**Figure 6.** EPI-slices (31-35) of one subject obtained with (upper panel) and without (middle panel) the montage of the EMG equipment. The data was realigned to the very first slice to remove movement artifacts. In the grey panel below, the difference between the EPI-slices is shown.

The association between the mean EMG values obtained during the scanning and the non-scanning intervals showed that the slopes and intercept were closer to the identity-line for the right FDI than for the left FDI. Also, the variability of the difference between the corrected and undisturbed EMG versus the mean of the corrected and undisturbed EMG was larger for the left FDI. The dissimilarity between the recordings of the right and left FDI is probably caused by a difference in the distance between the reference and the recording electrodes. In an ideal situation, the noise from the MR-scanner that is recorded by the EMG electrodes is identical to the noise that is picked-up by the reference electrode. However, the static magnetic field and the changing fields are not distributed homogeneously and, therefore, electrodes will be influenced
differently. Since the reference electrode in this experiment was placed on the processus styloidius radius of the right arm, which was closer to the right FDI, this resulted in a “cleaner” signal of the right FDI than of the left FDI. Thus, it is recommended to position the reference-electrode close to the recording-electrodes. However, in future experiments we want to record activity of several muscles on different sides of the body simultaneously. A pilot study, in which we systematically changed the location of the reference electrode, showed that positioning the reference-electrode on the ankle gives an optimal solution for recording activity of several muscles. In this case the MR-artifact is similar for all our EMG recordings; for individual EMG recordings, however, it is still advisable to position the reference electrode close to the recording electrodes.

The average artifact is equal to the actual artifact plus a standard deviation due to noise and a small EMG component. Movement of the wires results in movement artifacts due to changes of the position of the wire in the magnetic field, thereby increasing the noise component of the average artifact. This could impose problems during dynamic contractions. However, the influence of the movement artifact can be reduced by calculating the mean artifact of the bipolar EMG recordings, instead of unipolar EMG recordings. In the current study, movement of the wires was virtually non-existent. Hence, MRI artifact correction was applied to unipolar EMG recordings only.

Unlike the studies that combined EEG with fMRI (e.g. Krakow et al., 2000; Lazeyras et al., 2001), we did not find artifacts in the MR images or a difference in signal to noise ratio. The images obtained during EEG recording showed that the artifacts induced by the EEG equipment were local and since the electrodes during EMG measurements are not placed in the vicinity of the TR head coil no disturbing effects of the EMG equipment were expected.

In conclusion, our data showed that after MRI artifact correction accurate EMG recordings of isometric contractions can be obtained during fMRI scanning, both at high and low force levels. Simultaneous recordings of fMRI and EMG are expected to provide new insights in the functioning of the healthy motor system and the pathophysiology of movement disorders.