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Muscle fatigue-induced enhancement of corticomuscular coherence following sustained submaximal isometric contraction of the tibialis anterior muscle

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Ushiyama J, Katsu M, Masakado Y, Kimura A, Liu M, Ushiba J. Muscle fatigue-induced enhancement of corticomuscular coherence following sustained submaximal isometric contraction of the tibialis anterior muscle. *J Appl Physiol* 110: 1233–1240, 2011. First published March 10, 2011; doi:10.1152/japplphysiol.01194.2010.—Oscillatory activity of the sensorimotor cortex shows coherence with muscle activity within the 15- to 35-Hz frequency band (β-band) during weak to moderate sustained isometric contraction. We aimed to examine the acute changes in this corticomuscular coupling due to muscle fatigue and its effect on the steadiness of the exerted force. We quantified the coherence between the electroencephalogram (EEG) recorded over the sensorimotor cortex and the rectified surface electromyogram (EMG) of the tibialis anterior muscle as well as the coefficient of variance of the dorsiflexion force ($Force_{CV}$) and sum of the auto-power spectral density function of the force within the β-band ($Force_{\beta\text{-}PSD}$) during 30% of maximal voluntary contraction (MVC) for 60 s before (prefatiguing task) and after (postfatiguing task) muscle fatigue induced by sustained isometric contraction at 50% of MVC until exhaustion in seven healthy male subjects. The magnitude of the EEG-EMG coherence increased in the postfatiguing task in six of seven subjects. The maximal peak of EEG-EMG coherence stayed within the β-band in both pre- and postfatiguing tasks. Interestingly, two subjects, who had no significant EEG-EMG coherence in the prefatiguing task, showed significant coherence in the postfatiguing task. Additionally, $Force_{CV}$ and $Force_{\beta\text{-}PSD}$ significantly increased after muscle fatigue. These data suggest that when muscle fatigue develops, the central nervous system enhances oscillatory muscular activity in the β-band stronger coupled with the sensorimotor cortex activity accomplishing the sustained isometric contraction at lower performance levels.

electroencephalogram; electromyogram; electroencephalogram-electromyogram coherence; force steadiness; central fatigue

MUSCLE FATIGUE is defined as a progressive exercise-induced decline in the force generation capacity (4, 14), and processes contributing to muscle fatigue can be divided into central and peripheral fatigue (11, 14, 43). Peripheral fatigue refers to an impairment of the processes located at or distal to the neuromuscular junction, whereas central fatigue refers to a reduction in the net excitatory input to α-motoneurons (14, 43). To compensate for the progressive loss of force due to peripheral fatigue during a prolonged submaximal contraction, the nervous system usually progressively modulates excitatory input to the motoneuron pool until task failure. Indeed, to quantify central fatigue, many studies (1, 17, 29–31, 36) have demonstrated an exercise-induced failure of voluntary muscle activation, i.e., changes in the amplitudes and/or spectrum of electromyogram (EMG) signals.

Acute changes in the interaction between the sensorimotor cortex and peripheral neurons due to muscle fatigue have been widely studied, and there are several reports of muscle fatigue-induced changes in corticomuscular coherence (44, 49, 50). Coherence is a measure used to quantify the linear correlation between two signals in the frequency domain (21). Furthermore, by determining the coherence between the sensorimotor cortex activity as measured by electroencephalogram (EEG) or magnetoencephalogram (MEG) and the muscle activity as measured by EMG, significant corticomuscular coupling during weak to moderate intensity of sustained isometric contraction within the 15- to 35-Hz frequency band (β band) has been shown (12, 16, 19, 25, 27, 35, 41, 48). Potential binding mechanisms of corticomuscular coupling include modulation of corticomuscular coherence after visuomotor skill learning (38), immobilization (32), and development (15, 23), suggesting changes in sensorimotor integration processes between the cortex and peripheral neurons as part of the motor adaptation process. However, the acute changes in corticomuscular coupling due to muscle fatigue remain highly controversial. For example, Tecchio et al. (44) demonstrated an increase in the magnitude of MEG-EMG coherence after sustained maximal voluntary contraction (MVC) of the extensor digitorum communis until exhaustion. In contrast, weakening of EEG-EMG coherence during sustained isometric elbow flexion at 30% of MVC (49) and during 200 maximal intermittent handgrip contractions (50) have also been reported. In addition, it remains unclear how muscle fatigue-induced changes in corticomuscular coupling actually influence motor performance.

The aim of the present study was to examine muscle fatigue-induced changes in the magnitudes of EEG-EMG coherence after sustained submaximal isometric contraction until exhaustion and to determine the effects on motor performance as assessed by the steadiness of the exerted force. We used the tibialis anterior (TA) muscle as we have previously reported (46) that the distally located lower limb muscles, including the TA and soleus muscles, showed the greatest EEG-EMG coherence among various upper and lower limb muscles. Furthermore, the TA is widely used to determine fatigue-induced modulation of neuromuscular activation patterns such as the EMG amplitude, EMG spectrum, or firing rate of the motor unit (5, 6, 26, 28, 45).

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MATERIALS AND METHODS

Subjects. Seven healthy male volunteers (age: 22.7 ± 1.6 yr) participated in this study. All subjects had no history of neurological disorders. All participants provided informed consent after receiving a detailed explanation of the purpose and potential benefits and risks involved in the study. All experimental procedures were approved by the local ethics committee of the Faculty of Science and Technology of Keio University.

Force, EMG, and EEG recordings. Subjects were comfortably seated on an armchair in a semireclining position, and the right foot was attached to a footplate with the knee at 90° from full extension and the ankle at 0° (neutral position). Dorsiflexion force was recorded with a force transducer (TU-BR, TEAK, Tokyo, Japan) attached to the footplate. Surface EMG recordings were made from the TA of the right leg, over the muscle belly, using bipolar Ag/AgCl electrodes with a diameter of 10 mm and an interelectrode distance of 20 mm. Differential EEG recordings were made from the sensorimotor cortex using two Ag/AgCl surface electrodes with a diameter of 10 mm, placed at Cz (defined by the international 10-20 system) and its 20 mm left lateral position, according to our previous report (46). The reference electrode was placed at A2 (right earlobe). An additional electrode was placed at A1 (left earlobe) as a ground electrode. Impedances of the EEG and EMG electrodes were kept below 10 and 20 kΩ, respectively, during the recording. All analog EEG and EMG signals were amplified and bandpass filtered (EEG: 0.1–200 Hz and EMG: 0.5–1,000 Hz) using a standard EEG or EMG recording system (Neuropack MEB-4308, Nihon Kohden, Tokyo, Japan).

All analog signals were converted to digital signals at a sample frequency of 10 kHz by an analog-to-digital converter with 12-bit resolution (NI-6071E, National Instruments, Austin, TX) controlled by data logger software originally designed using MATLAB software (The Mathworks, Natick, MA). Digital data were stored on the hard disk of a personal computer.

Experimental protocol. Before experimentation, dorsiflexion force was determined for each subject when performing a MVC. After the MVC measurement, subjects performed a prefatiguing task of sustained isometric contraction at 30% of MVC for 60 s. After a sufficient rest period of 60 s, subjects performed sustained isometric dorsiflexion at 50% of MVC for as long as possible until the limit of endurance (fatiguing task), which was defined as the point at which the exerted force dropped below 40% of MVC for 5 s. Soon after the fatiguing task, subjects again performed a postfatiguing task of sustained isometric contraction at 30% of MVC for 60 s, similar to the prefatiguing task. In all tasks, dorsiflexion force was visualized on the computer screen positioned in front of the subjects, and subjects were instructed to maintain their exerted force as close as possible to the line corresponding to the target force level. At the end of the experiment, MVC force was measured again to evaluate the decrease in maximal force generation capacity due to muscle fatigue.

Data analysis. EMG and force signals were segmented into 8,192-point nonoverlapping epochs with the Hanning window function (73 epochs), and the power spectral densities (PSDs) of both signals were computed by Welch's method. To examine the muscle fatigue-induced changes in the characteristic spectral frequency of EMG, the mean power frequency of EMG (EMG_{MPF}) was then calculated for the frequency range as the ratio between the spectral moments of *orders 1* and *0* (1, 17, 29, 36). The first two frequency points past the direct current component (0 and 1 Hz) were not included in the analysis of EMG_{MPF}. The root mean square of raw EMG signals (EMG_{RMS}) was also calculated to examine muscle fatigue-induced changes in the amplitudes of the muscle activity (1, 31, 36). To evaluate the effects of muscle fatigue on the steadiness of the exerted force, the coefficient of variance of the force ($Force_{CV}$) was determined. Furthermore, since significant EEG-EMG coherence was observed in the β-band, as previously reported (3, 9, 12, 16, 19, 25, 27, 35, 37, 41, 48), we examined the muscle fatigue-induced changes in same frequency component of the force signal by determining the sum of the auto-PSD function of the force between 15 and 35 Hz ($Force_{\beta\text{-}PSD}$).

To evaluate the muscle fatigue-induced changes in corticomuscular coupling, we estimated EEG-EMG coherence in both pre- and postfatiguing tasks. When we estimated EEG-EMG coherence, EMG signals were rectified, as full-wave rectification is known to provide the temporal pattern of grouped firing motor units (20, 21). Raw EEG and rectified EMG signals were segmented into artifact-free 8,192-point epochs with no overlap. Each 8,192-point data segment was Hanning windowed to reduce spectral leakage. Correlations between EEG and rectified EMG [$C_{xy}(f)$] were calculated by coherence using the following equation:

$$|C_{xy}(f)| = \frac{\left|\overline{P_{xy}(f)}\right|^2}{\overline{P_{xx}(f)} \times \overline{P_{yy}(f)}} \quad (1)$$

where $\overline{P_{xx}(f)}$ and $\overline{P_{yy}(f)}$ are the averaged autospectra of the EEG and rectified EMG signals throughout the epochs for a given frequency f, respectively. $\overline{P_{xy}(f)}$ is the averaged cross-spectrum between these two parameters throughout the segments. The coherence function provides a normative measure of linear correlation on a scale of *0* to *1*, where *1* indicates a perfect linear correlation. According to Halliday et al. (21), we also estimated the phase spectrum, defined as the argument of cross-spectrum, to investigate the timing of information between EEG and EMG in the frequency domain.

We set the frequency range for the latter quantitative analyses at 3–50 Hz (including α-, β-, and γ-bands) and then defined the 95% confidence limit of EEG-EMG coherence according to Halliday et al. (21) and Rosenberg et al. (40). To eliminate the potential risk that the coherence value was judged significant due to statistical error, the 95% confidence limit was Bonferroni corrected for multiple comparisons throughout the frequency bins within 3–50 Hz (39 bins) (25). Thus, the EEG-EMG coherence was considered significant when the value exceeded the estimated significant level (SL) at 0.088.

Coherence was normalized with the use of the arc hyperbolic tangent transformation for the latter statistical analysis (21). As shown in Fig. 1, within the range of 3–50 Hz, we calculated the maximal peak of the coherence (Coh_{max}) and the frequencies where Coh_{max}

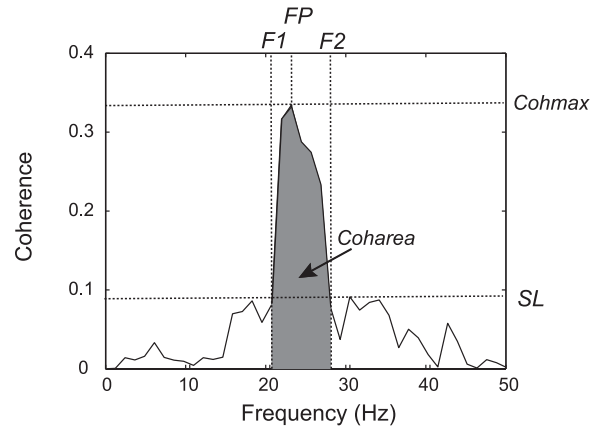

Fig. 1. Coherence spectrum between the electroencephalogram (EEG) of the sensorimotor cortex and the rectified electromyogram (EMG) during sustained isometric contraction of the tibialis anterior (TA) muscle at 30% of the maximal voluntary contraction (MVC). We calculated *1*) the maximal value of the EEG-EMG coherence (Coh_{max}), *2*) the frequency where Coh_{max} was observed (FP), *3*) the frequency where the coherence spectrum first met the estimated significance level of $P < 0.05$ (SL) when traced backward from FP (F_1), *4*) the frequency where the coherence spectrum first met the SL when traced forward from FP (F_2), and *5*) the area of the coherence curve in the frequency range of F_1 and F_2 (Coh_{area}). The gray area represents Coh_{area}.

values were observed (FP). When Coh_{max} was greater than the SL, we also calculated the frequencies where the coherence spectrum first met the SL when traced backward from FP (F_1) and where the coherence spectrum first met the SL when traced forward from FP (F_2). We then estimated the area of the coherence curve in the frequency range of F_1 and F_2 (Coh_{area}). When Coh_{max} was below SL, Coh_{area} was estimated as zero.

As for the data for the fatiguing task, we validated the time course of changes in EMG_{RMS}, EMG_{MPF}, $Force_{CV}$, and EEG-EMG coherence, which were calculated over a sliding short-time (10-s long) window, and the 10-s window was moved through the task in 1-s steps. When estimating the EEG-EMG coherence, each 10-s-long window was segmented into artifact-free 8,192-point epochs with half overlap (23 epochs).

Statistical analysis. Values are presented as means ± SD. To examine the effect of muscle fatigue on Coh_{max}, Coh_{area}, EMG_{RMS}, EMG_{MPF}, $Force_{CV}$, and $Force_{\beta\text{-}PSD}$ during sustained isometric contraction at 30% of MVC, paired-Student's *t*-tests were conducted. An α-level of 5% was chosen for all statistical analyses. All statistical analyses were performed using PASW statistics software (SPSS Japan, Tokyo, Japan).

RESULTS

Examples of the raw EEG signals, raw EMG signals, and raw force signals, auto-PSDs for the EEG, rectified EMG, and force signals, and coherence and phase spectra between the EEG and rectified EMG signals in the pre- and postfatiguing tasks are shown in Fig. 2. The amplitude of oscillatory EMG activity in the β-band increased due to muscle fatigue. Furthermore, EMG activity fluctuated more synchronously with the simultaneously recorded EEG activity in the postfatiguing task than in the prefatiguing task. Indeed, both EEG and EMG PSDs showed more distinct peaks in the β-band, and, as a result, the magnitude of EEG-EMG coherence within the β-band was greater in the postfatiguing task. Within the frequency range where the coherence spectra exceeded the SL, the phase spectra showed a constant positive slope, indicating that the EEG and EMG signals were phase locked and that EEG precedes EMG. In addition, the increase in the magnitude of EEG-EMG coherence due to muscle fatigue was clearly accompanied by increased force fluctuation. Indeed, both the low-frequency component (≤10 Hz) and high-frequency component, corresponding to the frequency range where significant EEG-EMG coherence was observed (β-band), of the force signal showed larger oscillations in the postfatiguing task.

Examples of progressive changes in raw EMG signals, raw force signals, EMG_{RMS}, EMG_{MPF}, $Force_{CV}$, and coherence between EEG and rectified EMG signals during the fatiguing

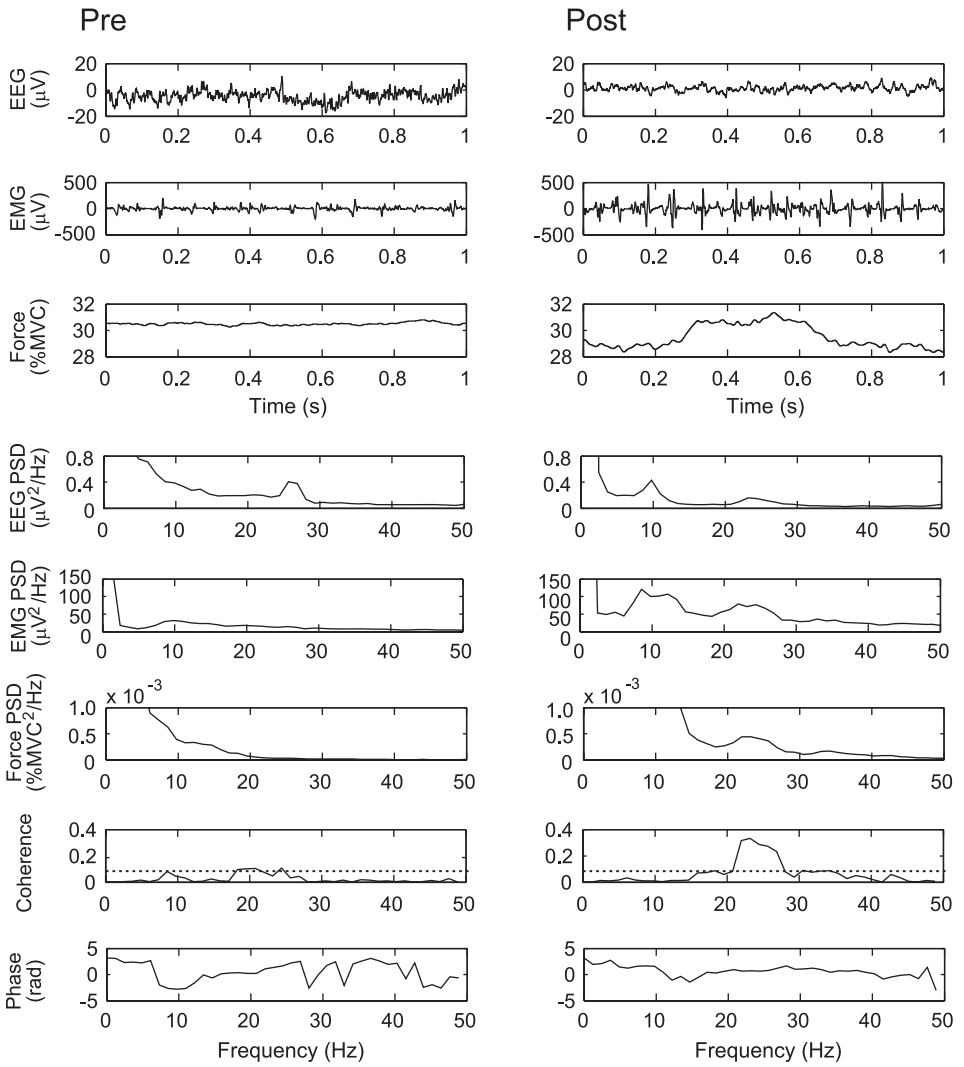

Fig. 2. Typical examples of raw EEG signals, raw EMG signals, and raw force signals, power spectral density functions (PSDs) for the EEG, rectified EMG, and force signals, and coherence and phase spectra between EEG and rectified EMG signals during sustained isometric contraction of the TA muscle at 30% of MVC before (Pre) and after (Post) muscle fatigue. In the coherence spectra, the estimated SL is shown by the horizontal dashed lines.

task are shown in Fig. 3. The mean time to task failure in the seven subjects was 174 ± 23 s. As observed in the raw EMG signal and EMG_{RMS}, the amplitude of EMG gradually increased, whereas EMG_{MPF} gradually decreased from the beginning of the fatiguing task. Both EMG_{RMS} and EMG_{MPF} reached a plateau near the midpoint of the task, after which the magnitude of EEG-EMG coherence dramatically increased. Force fluctuation remained constant during the first half of the task, but $Force_{CV}$ clearly increased with the increased magnitude of EEG-EMG coherence after the midpoint of the task. Although the progressive time course of these changes differed between subjects, almost all subjects showed a trend of increased EEG-EMG coherence and $Force_{CV}$ after EMG_{RMS} and EMG_{MPF} reached plateau.

The PSDs for raw EEG and rectified EMG signals and coherence spectra between the two signals in both the pre- and postfatiguing tasks for all subjects are shown in Fig. 4*A*. Increments in the magnitudes of EEG-EMG coherence were observed in six of seven subjects. For the majority of the subjects showing a dramatic increase in EEG-EMG coherence due to muscle fatigue, such as *subjects 1*, *3*, *5*, and *7*, the peak within the β-band appeared much more prominent in both EEG and EMG PSD in the postfatiguing task. Interestingly, two subjects (*subject 2* and *4*) who had no significant EEG-EMG coherence in the prefatiguing task showed significant coherence in the postfatiguing task. The frequency range where Coh_{max} occurred stayed within the β-band in both the pre- and postfatiguing tasks. Paired-Student's *t*-tests showed a significant increase in Coh_{max} (0.114 ± 0.005 to 0.344 ± 0.091, $P = 0.004$) and Coh_{area} (0.247 ± 0.279 to 0.802 ± 0.531, $P = 0.032$) due to muscle fatigue (Fig. 4*B*).

Group data for EMG_{RMS}, EMG_{MPF}, $Force_{CV}$, and $Force_{\beta\text{-}PSD}$ in the pre- and postfatiguing tasks are shown in Fig. 5. EMG_{RMS} significantly increased after muscle fatigue (59.9 ± 29.9 to 88.3 ± 38.1 mV, $P = 0.0107$), whereas EMG_{MPF} significantly decreased after muscle fatigue (83.8 ± 15.4 to 63.8 ± 9.0 Hz, $P = 0.0090$). In addition, there was a significant increase in $Force_{CV}$ (3.24 ± 1.18 to 6.42 ± 3.36%, $P = 0.0$) and $Force_{\beta\text{-}PSD}$ (0.00048 ± 0.00043 to 0.00298 ± 0.00351 $\%MVC^2$, $P = 0.038$).

DISCUSSION

The main finding of the present study was that the magnitude of EEG-EMG coherence significantly increased after muscle fatigue. The frequency range where Coh_{max} occurred stayed within the β-band in both pre- and postfatiguing tasks. Interestingly, two subjects, both of whom had no significant EEG-EMG coherence in the prefatiguing task, exhibited significant coherence in the postfatiguing task. In addition, $Force_{CV}$ and $Force_{\beta\text{-}PSD}$ significantly increased after muscle fatigue. These data suggest that oscillatory coupling between the sensorimotor cortex and the contracting muscle is enhanced due to muscle fatigue and that this enhancement in corticomuscular coupling impairs the control ability of muscle force during sustained submaximal isometric contraction.

Potential mechanisms of the enhanced corticomuscular coherence after muscle fatigue. In humans, corticomuscular coherence was initially assumed to be mediated by fast corticospinal axons and their monosynaptic connections to spinal motoneurons (8). This assumption suggests that oscillatory coupling between the sensorimotor cortex and the contracting muscle in the β-band may reflect discharge of corticospinal cells in this frequency range (8, 12). In addition, several studies (3, 7, 8, 19, 35) have demonstrated a phase delay between the synchronized EEG and EMG oscillations, indicating that the two signals are phase locked and that EEG precedes EMG. On the basis of these data, EMG oscillations are considered to

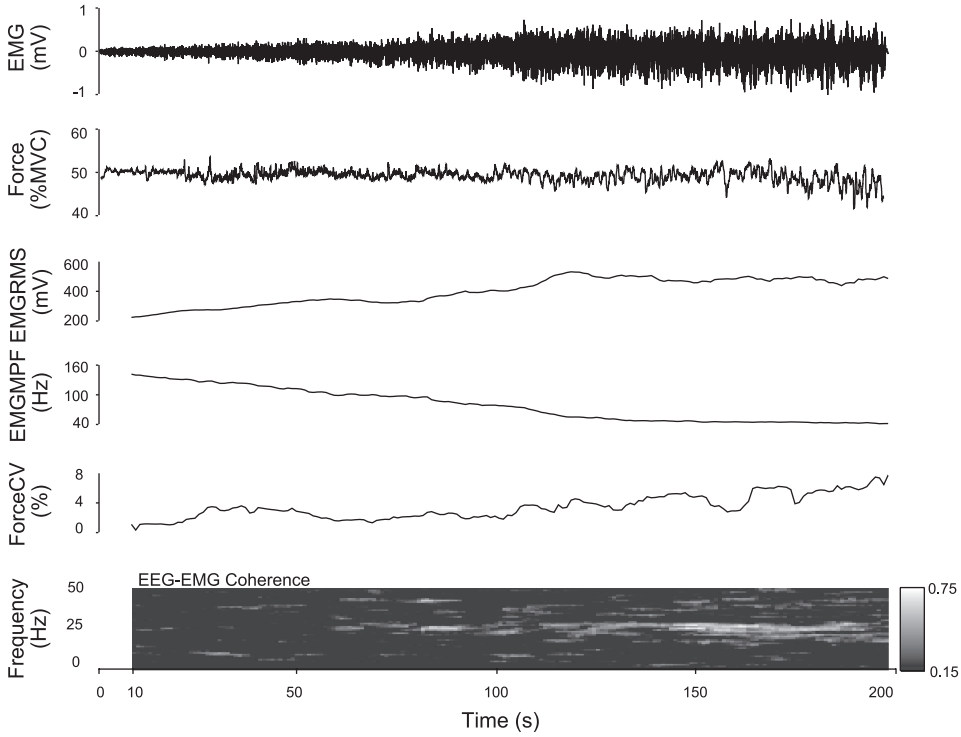

Fig. 3. Typical example of progressive changes in raw EMG, force, root mean square of raw EMG (EMG_{RMS}), mean power frequency of raw EMG (EMG_{MPF}), coefficient of variance of force output ($Force_{CV}$), and EEG-EMG coherence during sustained isometric contraction of the TA muscle at 50% of MVC until the limit of endurance (fatiguing task). We calculated EMG_{RMS}, EMG_{MPF}, $Force_{CV}$, and EEG-EMG coherence by sliding the 10-s-long window along the signal in 1-s steps. As for the time-frequency map of EEG-EMG coherence, the monochrome bar represents the magnitude of EEG-EMG coherence.

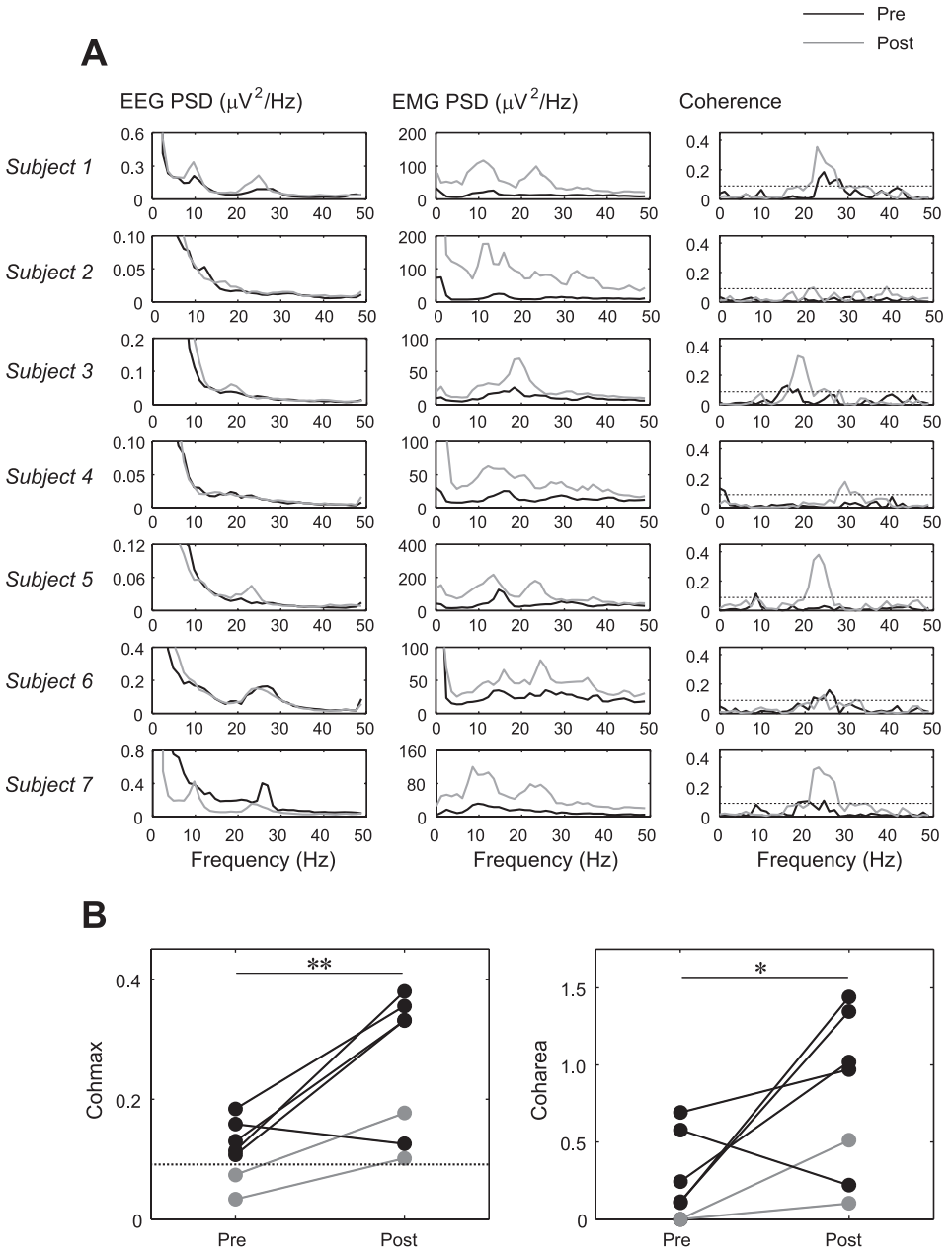

Fig. 4. Muscle fatigue-induced changes in the EEG PSD, EMG PSD, and EEG-EMG coherence during sustained isometric contraction of the TA muscle at 30% of MVC. *A*: PSDs for raw EEG and rectified EMG signals and coherence spectra between the EEG and rectified EMG in pre- and postfatiguing tasks for all subjects. Black and gray lines represent the data for pre- and postfatiguing tasks, respectively. The estimated SL is shown by the horizontal dashed lines. *B*: group data for Coh_{max} and Coh_{area} in the pre- and postfatiguing tasks. Shaded circles indicate the data for the subjects who did not show significant EEG-EMG coherence in the prefatiguing task (*subjects 2* and *4*). Significant differences between pre- and post-fatiguing tasks are shown ($*P < 0.05$; $**P < 0.01$).

largely result from sensorimotor cortex neural activity transmitted to the spinal motoneurons via corticospinal pathways. As such, the enhancement of the magnitude of EEG-EMG after muscle fatigue in our study may simply reflect an increase in the relative contribution of the descending command to the excitation of spinal motoneurons after muscle fatigue. However, it is also possible that the magnitude of EEG-EMG coherence is influenced by the tendency of the population of cortical neurons to discharge in synchrony (32), i.e., a larger magnitude of EEG-EMG coherence reflects a higher level of synchronization of corticospinal neurons, and these neuronal behaviors generate the oscillatory EMG activity in the β-band. Thus, it is likely that an increase in the tendency of the corticospinal cell population to discharge in synchrony may also account for the observed enhancement of the magnitude of EEG-EMG coherence after muscle fatigue.

Several recent studies have suggested that the generation mechanism of corticomuscular coherence may actually be more complex, i.e., that sensory feedback also contributes to the generation and/or modulation of corticomuscular coherence. For instance, Baker et al. (2) demonstrated that Ia afferent spiking showed coherence with oscillatory EMG activity over a wide frequency range, including the β-band, during isometric contraction. Thus, Ia afferent signals may be related to the generation and/or modulation of oscillatory EMG activity in the β-band. Furthermore, Pohja and Salenius (39) reported a decrease in corticomuscular coherence due to ischemia-induced deafferentation. As such, it is reasonable to assume that oscillatory signals from group Ia muscle spindle afferents return to the cortex via the spinal cord and contribute to the generation and/or modulation of synchronized activity of the cortical cell population that influences the oscillatory mus-

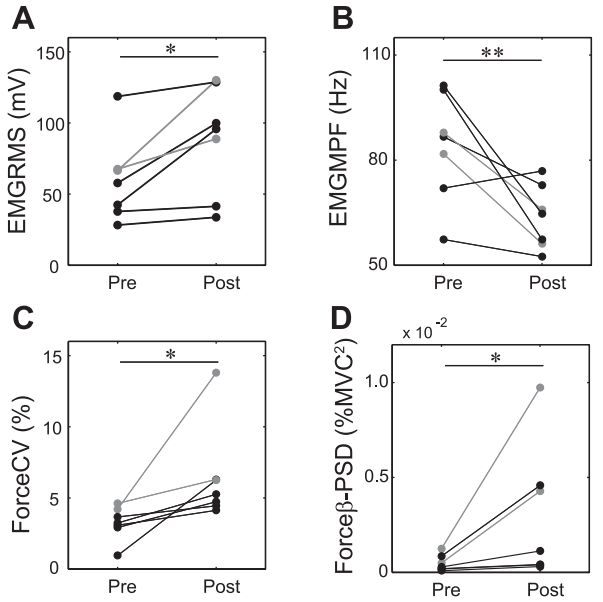

Fig. 5. Muscle fatigue-induced changes in the amplitude and characteristic spectral frequency of the EMG activity and force steadiness during sustained isometric contraction of the TA muscle at 30% of MVC. *A–D*: group data for EMG_{RMS} (*A*), EMG_{MPF} (*B*), $Force_{CV}$ (*C*), and sum of the power spectra of the force within the 15-to 35-Hz frequency band ($Force_{\beta\text{-}PSD}$; *D*) in the pre- and postfatiguing tasks. Shaded circles indicate the data for the subjects who did not show significant EEG-EMG coherence in the prefatiguing task (*subjects 2* and *4*). Significant differences between the pre- and postfatiguing tasks are shown (*$P < 0.05$; **$P < 0.01$).

cle activity. However, Macefield et al. (33) demonstrated that discharge frequency of the group Ia afferent decreases during sustained submaximal isometric contraction with time if muscle fatigue develops, resulting in a progressive decline in the motor unit firing rate. In the present study, as it is possible that excitatory input from group Ia afferents to the α-motoneuron pool was reduced due to muscle fatigue, a direct role for fatigue-induced changes in group Ia afferent feedback in the enhancement of corticomuscular coherence after muscle fatigue remains unclear. Nevertheless, there is evidence that during submaximal sustained isometric contraction, enhanced supraspinal drive compensates for a loss of excitation from group Ia afferents onto α-motoneurons (10, 31). Thus, we suggest that the fatigue-induced decline in group Ia afferent feedback reinforces cortical activity compensatively and indirectly leads to an enhancement of corticomuscular coupling.

Recurrent inhibition via Renshaw cells is considered to be an additional factor involved in generating and/or modulating the magnitude of corticomuscular coherence. Indeed, a recent computational modeling study (47) demonstrated a reduction of corticomuscular coherence by Renshaw cell inhibitory feedback. Furthermore, Löscher et al. (30) demonstrated a muscle fatigue-induced decrease in recurrent inhibition during sustained submaximal contraction, suggesting that modulation of Renshaw interneurons optimizes motor unit recruitment and firing rates of this muscle during sustained submaximal contraction. As we used a similar task to induce muscle fatigue in the present study, our experiment may have also induced a decline in recurrent inhibition. Thus, overall, it is possible that muscle fatigue-induced decline in recurrent inhibition contributed to the observed enhancement of the magnitude of EEG-EMG coherence in the postfatiguing task.

During strong voluntary contraction, the EMG demonstrates a tendency for rhythmic oscillations at ~40 Hz (the Piper rhythm), whereas EEG or MEG power in the 35- to 60-Hz frequency band (γ-band) shows coherence with this EMG Piper rhythm (7, 34). Since it is possible that the relative intensity of muscle contraction increased in the postfatiguing task because of the muscle fatigue-induced decrease of MVC force, it is likely that the peak of EEG-EMG coherence shifted to the γ-band in the postfatiguing task. In fact, we found that MVC force dropped by 16.9 ± 3.5% after muscle fatigue. However, the frequency range where Coh_{max} occurred stayed within the β-band in both the pre- and postfatiguing tasks for all subjects (Fig. 4*A*). Thus, muscle fatigue-induced enhancement of EEG-EMG coherence observed in the present study would be due to a different mechanism from the corticomuscular coherence in the γ-band during strong contraction without fatigue.

Functional significance of the enhanced corticomuscular coherence after muscle fatigue. Quantification of changes in the amplitude and spectrum of EMG signals is a well-established method for the evaluation of central fatigue. During sustained submaximal contraction, muscle fatigue induces an increase in EMG amplitude that is considered to reflect facilitated recruitment of new, unfatigued motor units (30, 31, 36) and a decrease in EMG_{MPF} that is considered to represent attenuated conduction velocity of action potentials and/or the firing rate of motor units (1, 29, 31, 36). Such changes in motor unit recruitment and firing rate would be likely to occur to compensate for the deficit in the developed force (30, 31, 36). In the present study, we also found that EMG_{RMS} was significantly larger and that the EMG_{MPF} was significantly smaller in the postfatiguing task than in the prefatiguing task. Furthermore, by validating the time course of changes in those measures during the fatiguing task, we found that the magnitude of EEG-EMG coherence was maintained constant when the EMG_{RMS} and EMG_{MPF} gradually increased in the early part of the fatiguing task and dramatically increased after EMG_{RMS} and EMG_{MPF} reached plateaus. These differences in the time ordering between the changes in EEG-EMG coherence and those in EMG_{RMS} and EMG_{MPF} during the fatiguing task suggest that a limit in the various compensatory strategies appears after muscle fatigue has developed completely. After this limit is reached, the descending command from the sensorimotor cortex becomes more rhythmic, leading to the enhancement of oscillatory synchronization of motor unit activities. Indeed, rhythmic grouped discharge in the EMG signals at ~20 Hz was clearer in the postfatiguing task than in the prefatiguing task. As such, the peak within the β-band was more prominent in EMG PSD in the postfatiguing task, with the majority of subjects showing a dramatic increase in EEG-EMG coherence due to muscle fatigue.

We also found that enhancement of the magnitude of EEG-EMG coherence in the postfatiguing task was accompanied by increased $Force_{CV}$, indicating attenuated force steadiness. The higher-frequency component of the force signal ($Force_{\beta\text{-}PSD}$), corresponding to the frequency range where significant EEG-EMG coherence was observed, also increased after muscle fatigue. Furthermore, by validating the time course of changes in these measures, we found that $Force_{CV}$ also dramatically increased after EMG_{RMS} and EMG_{MPF} reached plateau, as for the magnitude of EEG-EMG coherence. These findings sug-

gest that, particularly after the limit to recruit new unfatigued motor units and modulate the firing rate of motor units is reached, enhanced rhythmic drive from the sensorimotor cortex reinforces the rhythmic grouped discharge in EMG within the β-band. Thus, the exerted force by such a grouped discharge in EMG would be instantly increased and reach the target force level. The central nervous system may match the exerted force to the target force level by repeating this process at the final stage of muscle fatigue. This strategy would result in the attenuation of the steadiness of force in the postfatiguing state. Thus, the observed enhancement of corticomuscular coherence would be an effect due to muscle fatigue and reflect the additional strategy used by the central nervous system to accomplish the sustained contraction at the expense of force steadiness at the end of a fatiguing task or in a postfatiguing task.

We (46) recently reported that there is individual variation in the magnitude of EEG-EMG coherence, especially in the distally located lower limb muscles, and that the magnitude of EEG-EMG coherence is related to the degree of grouped discharge in EMG signals, i.e., subjects with greater EEG-EMG coherence show prominent oscillatory fluctuations in EMG signals within the β-band. In support of these findings, as mentioned above, the present study also demonstrated that fatigue-induced enhancement of EEG-EMG coherence led to larger oscillatory EMG activity in the β-band. In addition, the higher-frequency component of the force signal ($Force_{\beta\text{-}PSD}$), corresponding to the frequency range where significant EEG-EMG coherence was observed, increased after muscle fatigue. The major frequency range for the force fluctuations was ≤4–5 Hz, and this lower-frequency component of the force signal may result from a central loop that ordinarily keeps movements on target by comparing the motor command with the actual position of the body part by use of feedback information such as visual information (18). Although it is unclear how the higher-frequency component of force, which is directly influenced by oscillatory EMG activity in the β-band, is related to the generation of larger low-frequency force fluctuation, we suggest that the magnitude of corticomuscular coherence can regulate the extent to which the exerted force fluctuates during sustained submaximal isometric contraction. This provides an important functional role for corticomuscular coherence.

Recently, several studies reported acute changes in corticomuscular coherence due to muscle fatigue similar to those in the present study. Nevertheless, there remains some controversy. Tecchio et al. (44) demonstrated an increase in the magnitude of MEG-EMG coherence after sustained MVC of the extensor digitorum communis until exhaustion. In contrast, weakening of EEG-EMG coherence during sustained isometric elbow flexion at 30% of MVC (49) and during 200 maximal intermittent handgrip contractions (50) have also been reported. These differences may relate to differences in the task used to induce the muscle fatigue. For example, although there are some common features of central fatigue caused by submaximal or maximal efforts, central fatigue forms a larger proportion of attenuated force generation capacity when muscle fatigue was induced by sustained submaximal isometric contraction than by sustained or repetitive MVC, whereas peripheral fatigue is more prominent in MVC tasks (22, 42, 43). As in the present study, Yang et al. (49) used sustained submaximal isometric contraction to induce muscle fatigue. Nevertheless, they reported weakening of EEG-EMG coherence due to sustained isometric elbow flexion at 30% of MVC, in contrast to the fatigue-induced enhancement of EEG-EMG coherence in our study. It is possible that differences in the recorded muscle may also contribute to the acute changes of corticomuscular coupling due to muscle fatigue. We used the TA muscle to examine changes in EEG-EMG coherence with muscle fatigue, as we have previously reported (46) that the magnitude of EEG-EMG coherence was significantly greater in lower limb muscles than in upper limb muscles and was greater in distal muscles than in proximal muscles within each limb. Indeed, the magnitude of EEG-EMG coherence and the dynamic range of changes due to muscle fatigue in our study were much greater than those found by Yang et al. (49). In addition, based on the proportion of physiological cross-sectional area (PCSA), Kawakami et al. (24) demonstrated that several synergistic muscles contribute to the generation of elbow flexion force (biceps brachii: 34%, brachialis: 47%, and brachioradialis: 19%). Thus, in the study by Yang et al. (49), it is possible that compensatory activities were induced in synergistic muscles when muscle fatigue developed after elbow flexion. The TA accounts for ~60% of the PCSA of the total dorsiflexors (13) and is therefore regarded as a major agonist muscle for dorsiflexion. Thus, our research design represents the effect of muscle fatigue for one target muscle on corticomuscular coupling. Further investigation is required to clarify the task and/or muscle dependency of muscle fatigue-induced changes in corticomuscular coherence.

Conclusions. We demonstrated that the magnitude of EEG-EMG coherence in the β-band was significantly increased by muscle fatigue after sustained submaximal isometric contraction of the TA muscle until exhaustion. In addition, $Force_{CV}$ and $Force_{\beta\text{-}PSD}$ significantly increased after muscle fatigue. These data suggest that when muscle fatigue develops, the central nervous system enhances oscillatory muscular activity in the β-band stronger coupled with the sensorimotor cortex activity accomplishing the sustained isometric contraction at lower performance levels.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

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