Title:

The presence of corticomuscular coherence during unipedal stance

Authors:

Sylmina Dalily Alkaff[1,2], Junichi Ushiyama[3,4]

Authors' Affiliations:

[1]Graduate School of Media and Governance, Keio University, 5322 Endo, Fujisawa, Kanagawa 252-0882, Japan; [2]Department of Robotics and Artificial Intelligence, Faculty of Advanced Technology and Multidiscipline, Universitas Airlangga, Surabaya 60115, Indonesia; [3]Faculty of Environment and Information Studies, Keio University, 5322 Endo, Fujisawa, Kanagawa 252-0882, Japan; [4]Department of Rehabilitation Medicine, Keio University School of Medicine, 35 Shinanomachi, Shinjuku, Tokyo 160-8582, Japan

Corresponding Authors:

Junichi Ushiyama, Ph.D.

Faculty of Environment and Information Studies, Keio University, 5322 Endo, Fujisawa, Kanagawa 252-0882, Japan

Tel & Fax: +81-466-49-3525

Email: ushiyama@sfc.keio.ac.jp

Acknowledgments

This work was supported by grants from the Grant-in-Aid for Scientific Research (B) (Japan Society for the Promotion of Science, JSPS) (grant number 20H04091) to JU, and a designated donation from Living Platform, Ltd, Japan to JU. We thank Ms. Tomomi Hamaoka, Ms. Kana Iijima, and Ms. Chieko Matsuda for their secretarial assistance. For all authors, no conflicts of interest were declared. We thank Benjamin Knight, MSc., from Edanz (https://jp.edanz.com/ac) for editing a draft of this manuscript.

Abstract

Objective: To elucidate cortical involvement in postural control during unipedal stance by observing corticomuscular coherence (CMC) between the sensorimotor cortex and ankle joint muscles.

Methods: Twenty-one participants performed three tasks: bipedal stance, unipedal stance, and isometric contraction. We measured the maximal peak of CMC (CMCmax) between electroencephalograms overlying the foot representation area and surface electromyograms from the tibialis anterior (TA), medial gastrocnemius (MG), lateral gastrocnemius (LG), and soleus (SOL), respectively, for each task. We measured the center of pressure (COP) during both stance tasks.

Results: Although there was no significant CMC during bipedal stance, significant CMC was observed for all muscles during unipedal stance, with larger COP fluctuation. The results revealed significant differences in CMCmax between unipedal and bipedal stance tasks (TA, $p = 0.002$; MG, $p = 0.016$; LG, $p = 0.003$; SOL, $p = 0.009$). Additionally, CMCmax was obtained in higher frequency bands during the unipedal stance task than during the isometric contraction task.

Conclusions: Significant CMC indicates direct involvement of the sensorimotor cortex in postural control during unipedal stance.

Significance: Greater postural demands due to narrow base-of-support during unipedal stance requires voluntary control of muscle activity by the sensorimotor cortex.

Highlights:

- Corticomuscular coherence (CMC) was measured in ankle muscles during bipedal stance, unipedal stance and isometric contraction.
- Significant CMC was observed during unipedal stance but not during bipedal stance.
- The peak frequency of CMC was higher during unipedal stance than during isometric contraction.

Keywords:

- Electroencephalography
- Electromyography
- Coherence
- Center of pressure
- Postural control

1. Introduction

Standing on one leg (unipedal stance) is performed in various daily activities, such as putting on pants, climbing stairs, and playing a range of sports. This stance challenges the ability to maintain postural balance, as the base-of-support (BOS) is relatively narrow, with a high risk of falling. Therefore, the unipedal stance is often implemented for fall risk assessment, balance training and rehabilitation programs (Hurvitz et al., 2000; Trojian and McKeag, 2006; Zech et al., 2010, 2009).

Unipedal stance requires rapid sensorimotor integration to maintain the center of mass (COM) within the BOS. Thus, it is important to understand the neural mechanisms that control postural balance in this stance. Previous studies reported that electromyogram (EMG) activity, primarily in the ankle joints, was increased during unipedal stance with a larger center of pressure (COP) displacement in young adults (Amiridis et al., 2003; Billot et al., 2010; Riemann et al., 2003). These findings suggest that increased muscle activity is required to compensate larger COP fluctuations caused by higher postural demand in unipedal stance. Additionally, increased theta-band power and decreased alpha-band power were observed in electroencephalogram (EEG) signals over the sensorimotor cortex as the standing task difficulty increased by changing the stance from bipedal to unipedal (Edwards et al., 2018). Thus, it is reasonable to assume that the sensorimotor cortex might be actively involved in postural control during unipedal stance, to compensate for increased task difficulty. However, there is no direct evidence supporting this assumption to date.

One approach for determining the involvement of the sensorimotor cortex in motor control is analyzing the functional connection between cortical and muscle activities. Coherence between EEG over the sensorimotor cortex and EMG from the contracting muscle, known as corticomuscular coherence (CMC), can provide such clarification. When performing isometric voluntary contraction, CMC is typically observed within the β-band

(15–35 Hz) (Baker et al., 1997; Conway et al., 1995; Ushiyama et al., 2017, 2011). However, only a few studies have investigated CMC modulation during standing. Masakado et al. (2008) investigated CMC in several postural tasks and reported that significant β-band CMC was obtained only during foot stomping tasks. Jacobs et al. (2015) reported changes in β-band CMC during bipedal stance depending on the stance width, surface, and vision. However, these studies measured EMGs from a limited number of ankle joint muscles. Because the functional roles played in postural control differ across synergistic and/or antagonistic muscles around the ankle joint, it is important to analyze these muscles simultaneously to obtain a more complete understanding of the mechanisms underlying postural control.

The aim of the present study was to elucidate cortical involvement in postural control by analyzing CMC during unipedal stance. To this end, we recorded EMGs from four major muscles acting on the ankle joints of both legs (tibialis anterior [TA], medial gastrocnemius [MG], lateral gastrocnemius [LG], and soleus [SOL]), and measured their CMC with EEG over sensorimotor cortex. We compared the CMC during unipedal stance with that during bipedal stance and isometric contraction tasks.

2. Methods

2.1 Ethical Approval

The experiment protocols and procedures were conducted following the Declaration of Helsinki and were approved by the ethical committee of the Faculty of Policy Management, Faculty of Environment and Information Studies, and Graduate School of Media and Governance, Keio University (receipt number 167). All participants received a detailed explanation of the experiment and provided informed consent prior to data collection.

2.2 Participants

Twenty-one healthy young adults (thirteen male and eight female, aged 18–25 years) participated in this study. All participants reported no history of neuromuscular and/or musculoskeletal disorders. The dominant foot of each participant was determined using three sets of questions taken from the Waterloo Footedness Questionnaire-Revised (WFQ-R) questionnaire (van Melick et al., 2017).

2.3 Experimental Protocol

Participants performed an isometric contraction task and two standing tasks (i.e., bipedal and unipedal stance tasks). In the bipedal stance task, participants were instructed to stand with their feet separated by 10 cm and to keep their arms at their sides. Foot positions were marked with white tape to ensure that the same position was maintained throughout measurements. In the unipedal stance task, participants were instructed to stand on one leg and keep their arms at their sides, in the same manner as in the bipedal stance task. The unipedal stance task was conducted with each leg. Both standing tasks required participants to maintain an upright position and focus their eyes on a fixed target on a screen positioned approximately 2 m in front of them. When participants were unable to keep their posture straight, the task was stopped and repeated after a brief rest of approximately 5 minutes. Two trials were conducted for each task, with a duration of approximately 70 s for each trial. The order of the tasks was randomized for each participant.

The isometric contraction task was performed approximately 5 minutes after the standing tasks. Participants sat on a chair with each foot fitted to an ankle dynamometer by attached straps. Participants' knee joints were fully extended, while the ankle joints were set in a neutral position. Before the isometric contraction task, participants performed maximal voluntary contraction (MVC) of ankle dorsiflexion and plantarflexion with each leg for 3 s while remaining as stable as possible. We determined the MVC value as the peak torque from

this measurement. After a sufficient rest period of approximately 60 s, participants performed isometric ankle dorsiflexion and plantarflexion on 10% MVC for each leg. Each contraction was performed for approximately 70 s, and repeated twice. The intensity level for this task was determined from the average EMG activity level from all muscles while performing a bipedal stance task in preliminary experiments. Visual feedback was provided on a display placed approximately 2 m in front of the participants, consisting of target torque level (10% MVC) represented by a blue line and an exerted torque level represented by a red line. Participants were instructed to keep the red line as closely as possible to the blue line.

2.4 Data Collection

We recorded scalp EEG signals over the foot representation area of the sensorimotor cortex. Five passive Ag/AgCl electrodes with a diameter of 18 mm were placed at Cz and its surrounding positions (FCz, CPz, C1, and C2) defined by the international 10-20 system on an EEG cap (g.GAMMAcap 1027; Guger Technologies, Graz, Austria). Reference and ground electrodes were placed on the right and left earlobes, respectively. Surface EMG signals were recorded from TA, MG, LG, and SOL muscles from both legs. Two passive bipolar Ag/AgCl disposable electrodes with a diameter of 10 mm were placed over each muscle belly with an inter-electrode distance of 20 mm. EMG ground electrodes were placed on the kneecap on the right leg of the participant. Participants' skin was gently abrased and cleaned with alcohol before sensor placement. The torque signal measured during isometric contraction task was recorded using an ankle dynamometer (TCF100N; Takei Scientific Instruments Co., Ltd., Niigata, Japan).

Both EEG and EMG signals were amplified and band-passed filtered with a cut-off frequency of 0.5–100 Hz for EEG and 2–1000 Hz for EMG, using a 16-channel analog biosignal amplifier (g.BSamp 0201a; Guger Technologies, Graz, Austria). The torque signal was low-pass filtered using a second-order Butterworth filter at a cut-off frequency of 50 Hz

and amplified using an amplifier (Kyowa Electronics Instruments Co., Ltd., Tokyo, Japan). All signals were then converted to digital signals at a sample rate of 1000 Hz by an analog-to-digital converter with 16-bit resolution (NI USB-6259 BNC, National Instruments, Austin, TX, USA) controlled by data-logger software using MATLAB (The Mathworks, Natick, MA, USA).

The COP signals were recorded using a Wii Balance Board (RVL-021; Nintendo, Kyoto, Japan). Previous studies reported that Wii Balance Board provided a reliable, cost-effective alternative to an industrial force plate (Clark et al., 2014, 2010; Holmes et al., 2013; Imaizumi et al., 2018). Signals were collected via Bluetooth at a sample rate of 30 Hz by another data-logger software program using MATLAB.

2.5 Data Analyses

The EEG signal from Cz was first Laplacian-filtered by subtracting the average potential from surrounding electrodes, then filtered again using a second-order Butterworth low-pass filter at 50 Hz (Hjorth, 1975). The EMG signal was full-wave rectified because grouped discharge patterns could be detected, and was thus suitable for CMC analysis (Dakin et al., 2014; Myers et al., 2003; Ward et al., 2013).

The last 60 s period of each trial was extracted from EEG and EMG signals and combined into one 120 s data-set for further analysis. Prior to coherence analysis, both EEG and rectified EMG signals from each muscle were segmented into 1 s data windows, resulting in 120 data epochs without overlap. Each data epoch was then Hanning-windowed to reduce spectral leakage (Baker et al., 1997; Farmer et al., 1993; Gross et al., 2000) before we calculated the coherence between EEG and EMG of each signal. Coherence was calculated using the following formula (Halliday et al., 1995):

$$\left|C^2{}_{xy}(f)\right| = \frac{\left|\overline{P_{xy}(f)}\right|^2}{\overline{P_{xx}(f)} \cdot \overline{P_{yy}(f)}} \#(1)$$

where $\overline{P_{xy}(f)}$ is the averaged cross-spectral density between EEG and rectified EMG signals, and $\overline{P_{xx}(f)}$ and $\overline{P_{yy}(f)}$ are the averaged power spectral densities of EEG and rectified EMG signals, respectively, at any given frequency *f*.

The calculated coherence value is between 0 and 1, where 1 indicates that two signals are perfectly correlated. A 95% confidence limit was determined as the significance level (SL) of the coherence and the frequency range of interest was set between 3 and 50 Hz. In accordance with our previous studies (e.g., Ushiyama et al., 2017, 2011), we applied Bonferroni correction to the equation determining a 95% confidence limit to eliminate potential statistical errors caused by multiple comparisons, resulting in the following equation:

$$SL(\alpha) = 1 - \left[\frac{1}{N} \cdot \left(1 - \frac{\alpha}{100}\right)\right]^{1/(L-1)} \#(2)$$

where L is the number of data segments, N is the number of frequency bins, and α is the 95% confidence limit. As L and N were set as 120 and 48, respectively, the SL in this study was determined to be 0.0565. The calculated coherence was considered to be significant when the value exceeded the SL.

In this study, we calculated the maximal peak of coherence (CMCmax) and the peak frequency at which CMCmax was observed (PF) within the β-band (15–35 Hz) or γ-band (35–50 Hz). If the observed CMCmax value was above SL, we also calculated the frequencies where the coherence curve first met the SL when traced backward (F1) and forward (F2) from the PF (Ushiyama et al., 2010), as shown in Figure 1. Among all participants who showed significant CMCmax, the PF was observed within either band depending on the task. We examined the mean EMG activity level from each muscle during all tasks. We first calculated the average value of rectified EMG within a 1 s window with maximal averaged torque output during MVC trials (EMGmax). We then calculated the

average value of rectified EMG data for 120 s during each task and normalized it by the above-mentioned EMGmax.

COP data in the anterior-posterior (AP) and medio-lateral (ML) direction signals were low-pass filtered at 15 Hz with a fourth-order Butterworth filter. The last 60 s period of each trial was then extracted for further analysis. Based on previous studies, we subtracted the mean value from COP signals to remove the offset (Prieto et al., 2002; Ushiyama and Masani, 2011). We also calculated the resultant distance (RD) data series as the vector distance from mean COP to each data pair point in AP and ML data series, using the following equation:

$$RD[n] = \sqrt{(AP[n])^2 + (ML[n])^2}, n = 1,2, \ldots, N \ \#(3)$$

where N is the length of COP data included for the analysis. Various COP parameters can be used as measures of postural steadiness. We decided to calculate several parameters that have been most often used in posture and balance studies, such as standard deviation (SD), mean velocity (MVELO), mean displacement (MDIST), and mean frequency (MFREQ) for the RD direction. We chose to only include COP parameters in the RD direction as the result of COP fluctuation because the obtained results were similar to those in both the AP and ML directions. COP parameter results from each trial were then averaged for statistical analysis.

2.6 Statistics

Only participants showing significant CMC in at least one muscle during standing tasks were included for further analysis. To analyze the effects of standing tasks on each COP parameter in RD direction, we performed one-way repeated measures analysis of variance (ANOVA) to compare the average values of each COP parameter among stance tasks (i.e., bipedal stance, unipedal stance with dominant leg, and unipedal stance with the non-dominant leg). Two-way repeated measures ANOVA was performed to analyze the effects of leg dominance (dominant and non-dominant leg) and task (bipedal stance, unipedal stance,

and isometric contraction tasks) on CMCmax obtained from each muscle. In addition, two-way repeated measures ANOVA was performed to determine whether leg dominance and task (bipedal and unipedal stance) also had an effect on mean EMG activity level from each muscle. Greenhouse-Geisser correction was applied for sphericity. If a significant interaction was obtained, one-way repeated measures ANOVA was performed to determine whether the effect of tasks was significant on each muscle from each leg. If the result was significant, post-hoc pairwise comparisons with Bonferroni corrections were performed to determine whether there was a significant difference between tasks. However, if the interaction was not significant and at least one of the factors was significant, post-hoc pairwise comparisons with Bonferroni correction were executed to determine whether there was a significant difference within the factor for each muscle. The significance level was set at $\alpha = 0.05$ in all statistical tests. All statistical analyses were performed using SPSS software (IBM SPSS Statistics 26, IBM Developer Works, Tokyo, Japan).

3. Results

The results of the WFQ-R questionnaire revealed that all participants were right leg dominant. Fourteen participants showed significant CMCmax in at least one muscle during the unipedal stance task.

3.1 Typical example of EEG, EMG, and COP raw data in standing tasks

To depict the difference between bipedal and unipedal stance tasks, we showed typical EEG and rectified EMG signals during each standing task from a participant (Figure 2). The MG and SOL muscles were active during both bipedal and unipedal stance tasks. The TA and LG muscles showed little activity during the bipedal stance task, while they showed increased activity during the unipedal stance task. Moreover, we observed clearer oscillation in all muscles during the unipedal stance task than during the bipedal stance task.

The results revealed an increase in COP fluctuation during both unipedal stance tasks compared with the bipedal stance task (Figure 3). This finding indicates that participants were least stable during both unipedal stance tasks than during the bipedal stance task.

3.2 Differences in COP parameters between bipedal and unipedal standing tasks

Figure 4 represents the group data from all COP parameters on each standing task. One-way repeated measures ANOVA test revealed significant differences between tasks on all COP parameters (MDIST, $F_{2,26} = 51.995$, $p < 0.001$; SD, $F_{2,26} = 51.516$, $p < 0.001$; MVELO, $F_{2,26} = 131.829$, $p < 0.001$; MFREQ, $F_{2,26} = 29.097$; $p < 0.001$). Post-hoc analysis confirmed that all parameters during both unipedal stance tasks with the dominant leg (MDIST, 0.836 ± 0.134 cm; SD, 0.959 ± 0.151 cm; MVELO, 1.551 ± 0.314 cm/s; MFREQ, 0.301 ± 0.059 Hz) and with the non-dominant leg (MDIST, 0.763 ± 0.123 cm; SD, 0.870 ± 0.141 cm; MVELO, 1.612 ± 0.374 cm/s; MFREQ, 0.343 ± 0.072 Hz) were significantly higher than those during the bipedal stance task (MDIST, 0.393 ± 0.122 cm; SD, 0.463 ± 0.140 cm; MVELO, 0.459 ± 0.073 cm/s; MFREQ, 0.200 ± 0.041 Hz) (all, $p < 0.001$). This finding indicates that humans are inherently unstable during unipedal stance. In addition, post-hoc analysis revealed that there was no significant difference in COP parameters between unipedal stance using the dominant and non-dominant legs, indicating that leg dominance did not affect instability of participants (MDIST, $p = 0.421$; SD, $p = 0.333$; MVELO, $p = 0.497$; MFREQ, $p = 0.168$).

3.3 Differences in CMC and EMG activity between standing and isometric contraction tasks

Figure 5 shows group data for CMCmax from all muscles on each task. Two-way repeated measures ANOVA revealed that the main effect of task on CMCmax was significant for all muscles (TA, $F_{1.17,15.17} = 14.419$, $p = 0.001$; MG, $F_{1.24,16.1} = 7.733$, $p = 0.01$; LG, $F_{2,26} = 5.876$, $p = 0.008$; SOL, $F_{2,26} = 4.528$, $p = 0.021$). However, there was no significant effect of

leg dominance (TA, $F_{1,13} = 0.135$, $p = 0.719$; MG, $F_{1,13} = 3.741$, $p = 0.075$; LG, $F_{1,13} = 0.100$, $p = 0.757$; SOL, $F_{1,13} = 0.000$, $p = 0.986$) and no significant interaction between leg dominance and task for all muscles (TA, $F_{1.35,17.53} = 0.161$, $p = 0.767$; MG, $F_{1.24,16.07} = 0.875$, $p = 0.386$; LG, $F_{2,26} = 1.657$, $p = 0.210$; SOL, $F_{1.40,18.27} = 0.048$, $p = 0.902$). For further statistical analysis, CMCmax data from both legs were combined and post-hoc tests were performed for comparing differences in CMCmax among tasks for each muscle. The results revealed that all muscles showed significantly higher CMCmax during the unipedal stance task (TA, 0.069 ± 0.033; MG, 0.060 ± 0.034; LG, 0.067 ± 0.030; SOL, 0.074 ± 0.038) compared with the bipedal stance task (TA, 0.037 ± 0.011; MG, 0.033 ± 0.011; LG, 0.037 ± 0.011; SOL, 0.041 ± 0.015) (TA, $p = 0.002$; MG, $p = 0.016$; LG, $p = 0.003$; SOL, $p = 0.009$). In addition, both the TA and MG muscles showed significantly higher CMCmax during the isometric contraction task (TA, 0.132 ± 0.090; MG, 0.101 ± 0.080) compared with the bipedal stance task (TA, $p = 0.003$; MG, $p = 0.015$). Only the TA muscle showed significantly higher CMC during the isometric contraction task compared with the unipedal stance task ($p = 0.028$). Although we could not perform statistical comparison of CMCmax frequency distribution between the unipedal stance and isometric contraction tasks because of inconsistency of numbers of data-points with significant CMCmax between tasks, we found a tendency for the PF to shift to high β- or γ-band during the unipedal stance task in all muscles (Figure 6).

Figure 7 shows the group data for mean EMG activity level from all muscles between tasks. Two-way repeated measures ANOVA revealed that the main effect of task on the mean EMG activity level in all muscles was significant (TA, $F_{1,13} = 37.014$, $p < 0.001$; MG, $F_{1,13} = 35.525$, $p < 0.001$; LG, $F_{1,13} = 16.850$, $p = 0.001$; SOL, $F_{1,13} = 18.939$, $p < 0.001$). However, there was no significant effect of leg dominance (TA, $F_{1,13} = 2.046$, $p = 0.176$; MG, $F_{1,13} = 0.544$, $p = 0.474$; LG, $F_{1,13} = 4.614$, $p = 0.051$, SOL, $F_{1,13} = 3.455$, $p = 0.086$) and no

significant interaction between leg dominance and task for all muscles (TA, $F_{1,13} = 0.148$, $p = 0.707$; MG, $F_{1,13} = 0.282$, $p = 0.683$; LG, $F_{1,13} = 0.993$, $p = 0.337$; SOL, $F_{1,13} = 0.159$, $p = 0.697$). Similar to CMCmax, mean EMG activity level data from both legs were combined and post-hoc tests were performed to compare the difference among tasks for each muscle. The results revealed that all muscles exhibited significantly higher mean EMG activity level in the unipedal stance task (TA, 11.031 ± 5.547 %EMGmax; MG, 29.746 ± 15.668 %EMGmax; LG, 18.503 ± 11.293 %EMGmax; SOL, 35.618 ± 16.774 %EMGmax) compared with the bipedal stance task (TA, 5.166 ± 3.945 %EMGmax; MG, 7.821 ± 5.231 %EMGmax; LG, 11.264 ± 6.802 %EMGmax; SOL, 22.986 ± 13.570 %EMGmax) (TA, $p < 0.001$; MG, $p < 0.001$; LG, $p = 0.001$; SOL, $p < 0.001$).

4. Discussion

The present study investigated the involvement of the cortex in postural control during unipedal stance by measuring CMC between EEG over the sensorimotor cortex and EMG from ankle extensor and flexor muscles and conducting comparisons with CMC values recorded during bipedal stance. We found significant CMC for all muscles during the unipedal stance task, with larger COP fluctuation and increased muscle activation level, whereas no significant CMC was observed during the bipedal stance task.

4.1 Stronger CMC during unipedal stance indicated increased direct cortical involvement in postural control

The current findings are in accordance with previous reports (Jacobs et al., 2015; Masakado et al., 2008), revealing no significant CMC from any muscles during bipedal stance. Because postural balance is mainly regulated by the cerebellum (Thach and Bastian, 2004), basal ganglia, and brainstem (Takakusaki et al., 2004), less or no involvement of the

sensorimotor cortex in postural balance control during bipedal stance would be expected to result in a lack of significant CMC in the bipedal stance task. However, in the present study, significant CMC was observed during the unipedal stance. Indeed, CMCmax was significantly stronger in the unipedal stance task compared with the bipedal stance task for all muscles. Therefore, the sensorimotor cortex appears to be directly involved in the control of muscle activity around the ankle joint during unipedal stance.

Biomechanically, the human body during standing can be represented as an inverted pendulum rotating around the ankle joint with COM located in front of the joint (Winter, 1995). During bipedal stance, postural sway in response to counteracting gravitational force occurred mainly in the AP direction and was found to be coherent with ankle extensor muscle activity (Gatev et al., 1999; Loram et al., 2005). However, as COP fluctuation becomes larger and faster because of narrow BOS, postural control strategies would be expected to be modified during unipedal stance.

As a biarticular muscle crossing the ankle and knee joints, motor unit recruitment in the MG is known to be triggered by forward sway occurring during bipedal stance, resulting in a phasic muscular activity pattern (Vieira et al., 2011). Consequently, MG muscle activity was reported to be coherent with body angle during bipedal stance (Sasagawa et al., 2009). This suggests that MG is responsible for regulating body fluctuation in postural control. Regarding unipedal stance, greater effort would be expected to be required for MG to compensate larger body fluctuation occurring because of narrow BOS, as reflected by the increased muscle activity level. Increased CMC during the unipedal stance suggests that MG requires additional involvement from the sensorimotor cortex to regulate body fluctuation.

As a monoarticular muscle, SOL muscle activity was found to be linearly correlated with ankle angle and ankle torque during bipedal stance (Fitzpatrick et al., 1992; Mochizuki et al., 2006). This suggests that SOL is responsible for maintaining ankle joint tone during

quiet standing. In the case of unipedal stance, narrow BOS caused destabilization of postural balance, requiring the SOL muscle to increase ankle joint tone. Increased CMC during unipedal stance suggests that the SOL muscle requires additional involvement from the sensorimotor cortex to increase ankle joint tone.

Research on the functional role of the LG muscle in postural control has been less extensive compared with the MG and SOL muscles (Gatev et al., 1999; Héroux et al., 2014; Watanabe et al., 2018). This is because postural sway during the bipedal stance occurs mainly in the AP direction, resulting in less contribution from the LG muscle to postural control. However, during unipedal stance, the BOS area was significantly decreased in the frontal plane, resulting in imbalance in ML postural sway. Increased CMC in the unipedal stance indicated that increased LG muscle activity level required additional involvement from the sensorimotor cortex to maintain postural balance in the ML direction.

As an antagonist muscle from the ankle extensor muscles, TA muscle activation is not required during the bipedal stance because the focus of postural control during the bipedal stance is to counteract forward sway caused by gravitational pull. However, the contribution of TA muscle activity to postural control is known to be increased when greater backward sway occurs, either because of narrow BOS (Lemos et al., 2015; Sozzi et al., 2013) or backward external perturbation (Hwang et al., 2009). In the current study, CMC obtained from the TA muscle was increased during unipedal stance, where BOS is relatively narrow. This indicates that increased TA muscle activation requires additional involvement from the sensorimotor cortex to provide forward sway to prevent backward falling.

As such, each ankle joint muscle would be expected to play different functional roles for postural control depending on the requirement from the postural response. During bipedal stance, the postural response was sufficient to maintain postural balance within BOS with less to no contribution from the sensorimotor cortex to postural control. However, the decreased

BOS area when standing in the unipedal stance induces an imbalance in postural sway in both the AP and ML directions, as reflected by larger and faster COP fluctuation. To compensate for the greater postural demand during the unipedal stance, each muscle must increase the effort to maintain postural balance by increasing the activity level, requiring additional involvement from the sensorimotor cortex to provide more voluntary postural adjustment.

4.2 CMC peak frequency tends to shift towards higher β-band or γ-band during unipedal stance

The current results revealed a tendency for significant CMCmax during unipedal stance to be observed within higher frequency bands for all muscles. Previous studies reported a frequency shift toward the γ-band during a dynamic contraction task (Gwin and Ferris, 2012; Omlor et al., 2007) or during stronger intensity of an isometric contraction task (Brown et al., 1998; Mima et al., 1999; Ushiyama et al., 2012). However, it should be noted that a frequency shift toward high β-band or γ-band during unipedal stance occurred in the present study despite the low to moderate muscle activity level (less than 60% EMGmax). Masakado et al. (2008) also reported a similar frequency shift toward higher frequency bands during unipedal stance in one participant with significant CMC, despite all of the other participants showing no significant CMC. A possible explanation for this phenomenon is that larger and faster COP fluctuation during unipedal stance demands faster postural correction, thus requiring the sensorimotor system to integrate the relevant sensory information with motor commands at a higher rate. By increasing oscillation frequency (i.e., γ-band), more and faster information could be exchanged within the corticospinal loop. Therefore, frequency shift tendency in CMC may have occurred to facilitate the demand of rapid sensorimotor integration because of the difficulty in maintaining postural balance caused by narrow BOS in the unipedal stance task.

4.3 Technical limitations of the study

Several potential limitations of the current study should be considered. First, the presence of an anatomical buffer, such as skull thickness, cerebrospinal fluid, and location and orientation of the relevant cortical neuron, may have influenced the signals measured by scalp EEG (Malmivuo et al., 1997; Nunez, 1989; Yan et al., 1991). Similar anatomical issues, such as orientation, size, and depth of muscle fibers, can also affect surface EMG recordings (Farina et al., 2002; Mesin et al., 2009). However, while these limitations might affect the individual amplitude of EEG and EMG signals, CMC is a measure of the constancy of the amplitude ratio and phase difference between two signals (Halliday et al., 1995). Therefore, we believe that these technical limitations were not likely to have had a substantial influence on the present results.

Second, the present study focused only on muscles around the ankle joint. However, it is possible that other proximal muscles (i.e., muscles acting on the hip and trunk joint) could have contributed to postural control, particularly during unipedal stance. Older adults, for example, were found to exhibit increased hip angle movement accompanied by greater hip muscle EMG activity compared with younger adults during unipedal stance (Amiridis et al., 2003; Riemann et al., 2003). If a similar study was expanded to include older adults, it would be necessary to record EMG signals from proximal muscles and evaluate CMC. However, younger adults are known to mainly adopt ankle strategies for postural control with little contribution from the hip joint during both bipedal and unipedal stance (Amiridis et al., 2003; Billot et al., 2010; Riemann et al., 2003). Therefore, although the contribution of hip movements to postural control cannot be ignored, we believe that the present approach, which focused on CMC in ankle muscles, was effective for elucidating the neural mechanisms involved in postural control in young adults.

Finally, it is known that individual differences exist within CMC (Ushiyama et al., 2017, 2011), suggesting that some individuals or populations may not exhibit significant

CMC. Although we rejected participants who did not show significant CMC for all tasks in the current study, this does not imply that they do not utilize neural interactions between the sensorimotor cortex and contracting muscle during the task. Further studies using other measurements would be needed if participants who did not show significant CMC were included.

5. Conclusion

The involvement of the sensorimotor cortex in postural control during unipedal stance was investigated by measuring coherence between EEG over the sensorimotor cortex and EMG from the ankle flexor and extensor muscles. We found significant CMC during unipedal stance for all muscles, while no significant CMC was observed in the bipedal stance task. In addition, there was a tendency for the CMC PF to shift toward a higher frequency band during unipedal stance compared with isometric contraction. These findings suggest that the sensorimotor cortex is directly involved in postural control, providing more voluntary postural adjustment during unipedal stance because of narrow BOS.

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Figure Captions

Figure 1. An example of a coherence spectrum between electroencephalogram (EEG) over the sensorimotor cortex and rectified electromyogram (EMG). The figure shows data from the tibialis anterior muscle (TA) during the isometric contraction task. We calculated the maximum value of CMC (CMCmax), the peak frequency at which CMCmax was observed (PF), the frequency at which coherence first met the significance level (SL) when traced backwards (F1) and forward (F2) from PF. CMCmax was considered statistically significant when it was greater than SL.

Figure 2. Examples of raw EEG over the sensorimotor cortex and rectified EMG from the four muscles (TA; Medial Gastrocnemius, MG; Lateral Gastrocnemius, LG; Soleus, Sol) on each leg during the bipedal stance and unipedal stance tasks. Data were taken from a participant who showed significant CMC only during the unipedal stance task. Note that the unipedal stance task data were collected separately from each leg, while the bipedal stance task data were collected simultaneously from both legs.

Figure 3. Typical examples of center of pressure (COP) displacement signals in anterior-posterior (AP) and medio-lateral (ML) directions from the same participant as in Figure 2. The figure shows data from the bipedal stance task (A), the unipedal stance task performed by the dominant leg (B), and the unipedal stance task performed by the non-dominant leg (C). During the unipedal stance task, COP fluctuation was larger compared with during the bipedal stance task.

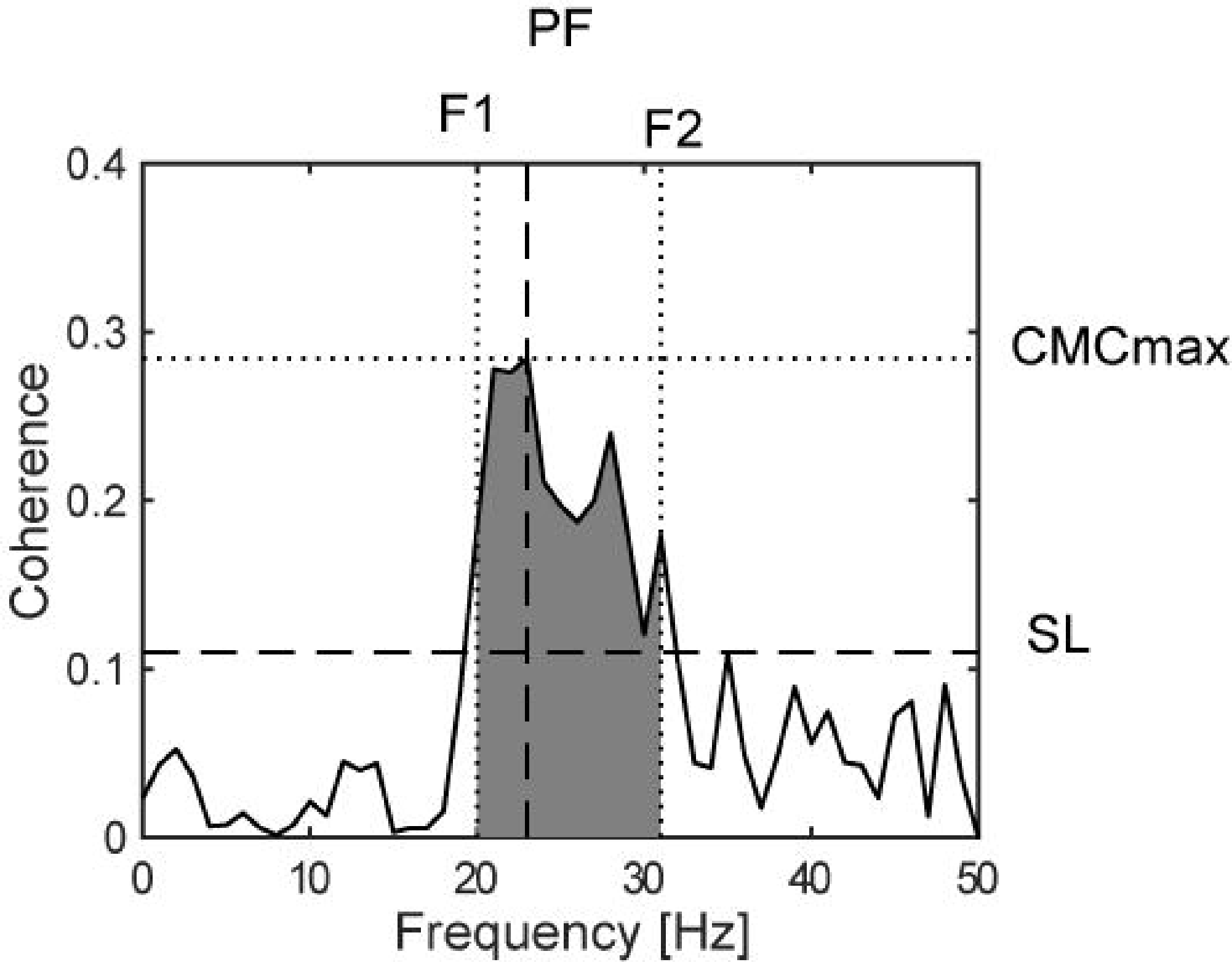
Figure 4. Group data (mean ± SD) for COP parameters on three tasks (bipedal stance, unipedal stance performed by the dominant leg, and unipedal stance performed by the non-dominant leg). Black bars indicate the average COP parameter value while the black line on the bar indicates standard deviation (SD) across participants. Significant differences between tasks are denoted as follows: *$p < 0.05$, **$p < 0.01$, ***$p < 0.001$.

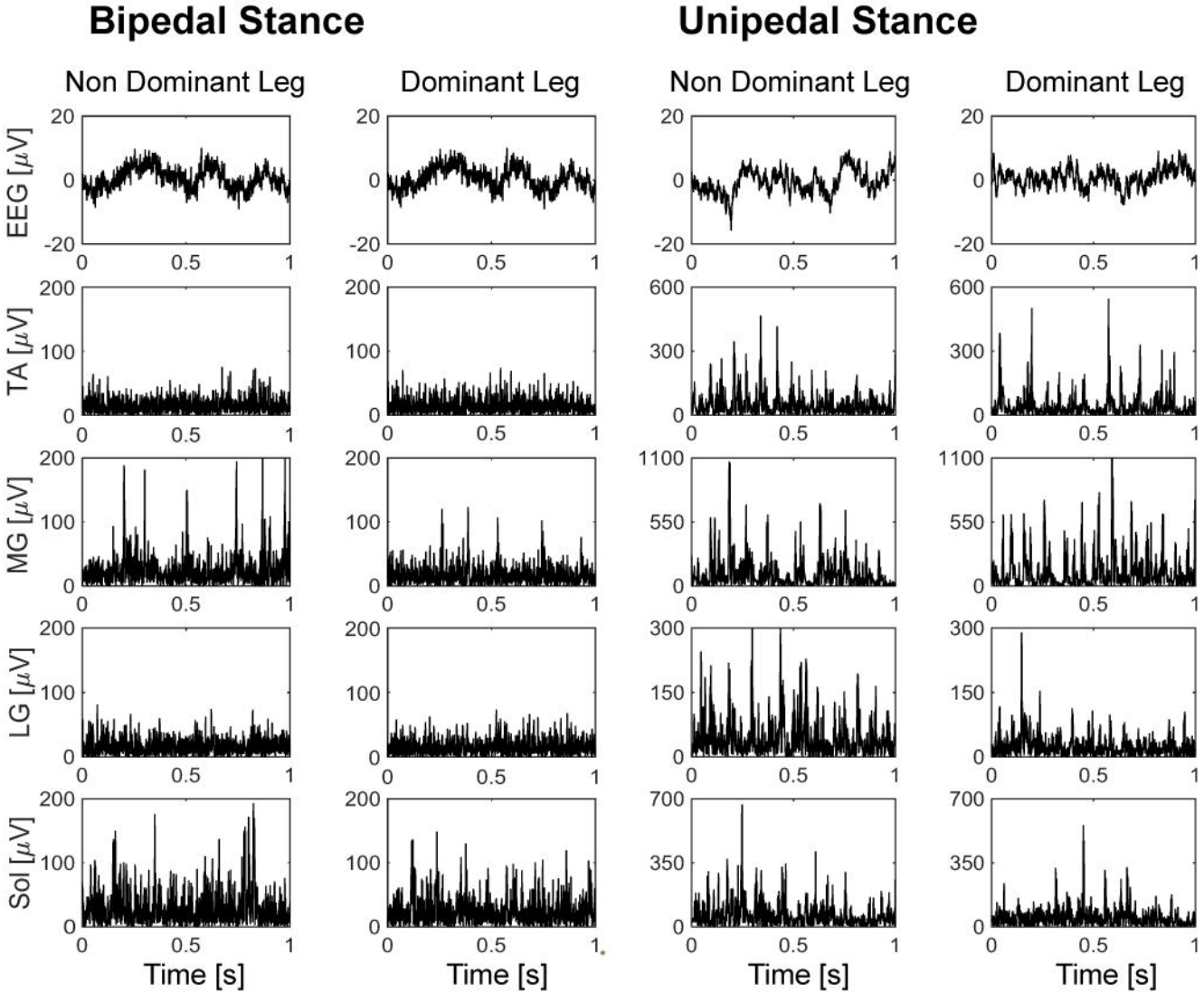
Figure 5. Group data (mean ± SD) for CMCmax from four muscles on three tasks (bipedal stance, unipedal stance, and isometric contraction). Note that the data from the dominant and non-dominant legs were combined into one figure for each muscle as the statistical analysis showed that there was no significant effect of leg dominance on CMCmax. Black bars indicate average CMCmax, while black lines on the bar indicate standard deviation (SD) across participants. The SL is represented as horizontal dashed lines. CMCmax was higher during the unipedal and isometric contraction tasks compared with the bipedal stance task. Significant differences between tasks are denoted as follows: *$p < 0.05$, **$p < 0.01$, ***$p < 0.001$.

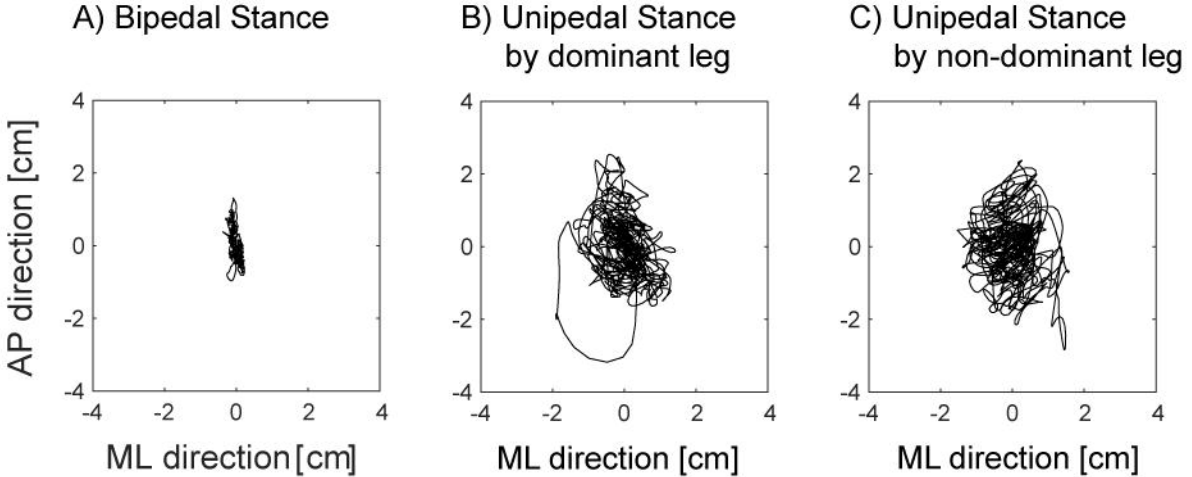
Figure 6. Distribution relationship between frequency width of significant CMC and CMCmax during isometric contraction and unipedal stance tasks. Data from dominant and non-dominant legs were merged into one figure for each muscle. Black squares indicate the PF value during the isometric contraction task, while asterisks indicate the PF value during the unipedal stance task. The error bar indicates the width of F1–F2.

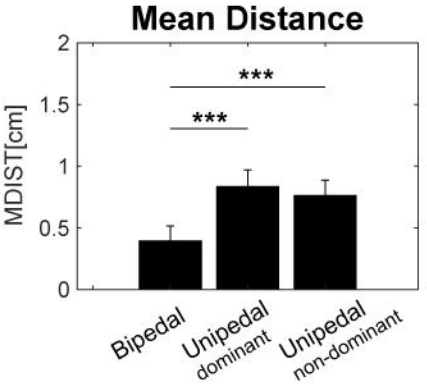
Figure 7. Group data (mean ± SD) for mean EMG activity level from four muscles on each task. Data from both dominant and non-dominant legs were merged into one figure for each

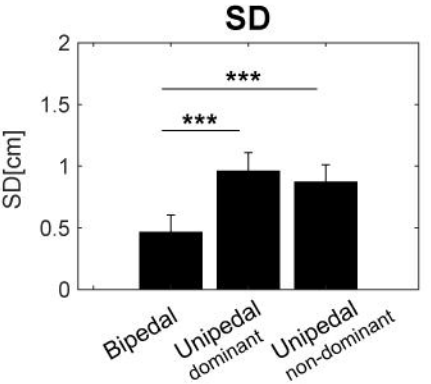
muscle as the statistical analysis showed that there was no significant effect of leg dominance on mean EMG activity level. Black bars indicate the average value of mean EMG activity level while the black line on the bar indicates SD across participants. Significant differences between tasks are denoted as follows: *$p < 0.05$, **$p < 0.01$, ***$p < 0.001$.

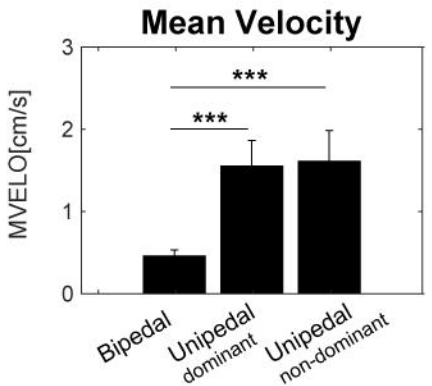

PF
F1
F2
CMCmax
SL
Coherence
0.4
0.3
0.2
0.1
0
0
10
20
30
40
50
Frequency [Hz]

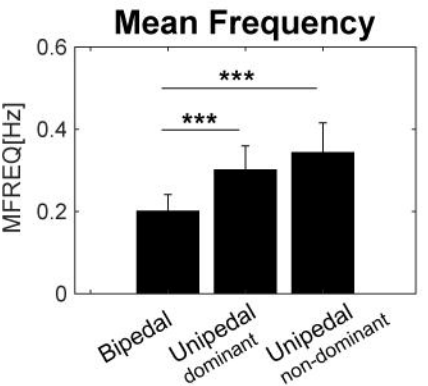

Bipedal Stance
Unipedal Stance
Non Dominant Leg
Dominant Leg
Non Dominant Leg
Dominant Leg
EEG [μV]
TA [μV]
MG [μV]
LG [μV]
Sol [μV]
Time [s]
Time [s]
Time [s]
Time [s]

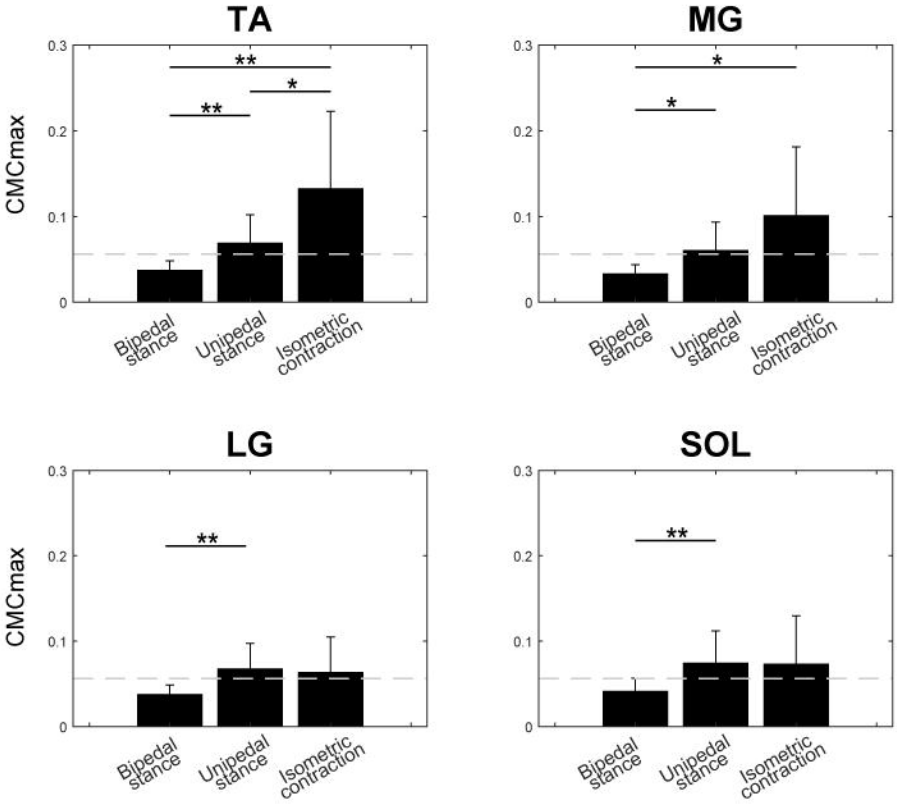

A) Bipedal Stance
B) Unipedal Stance by dominant leg
C) Unipedal Stance by non-dominant leg
AP direction [cm]
ML direction [cm]
ML direction [cm]
ML direction [cm]
4
2
0
-2
-4
-4
-2
0
2
4

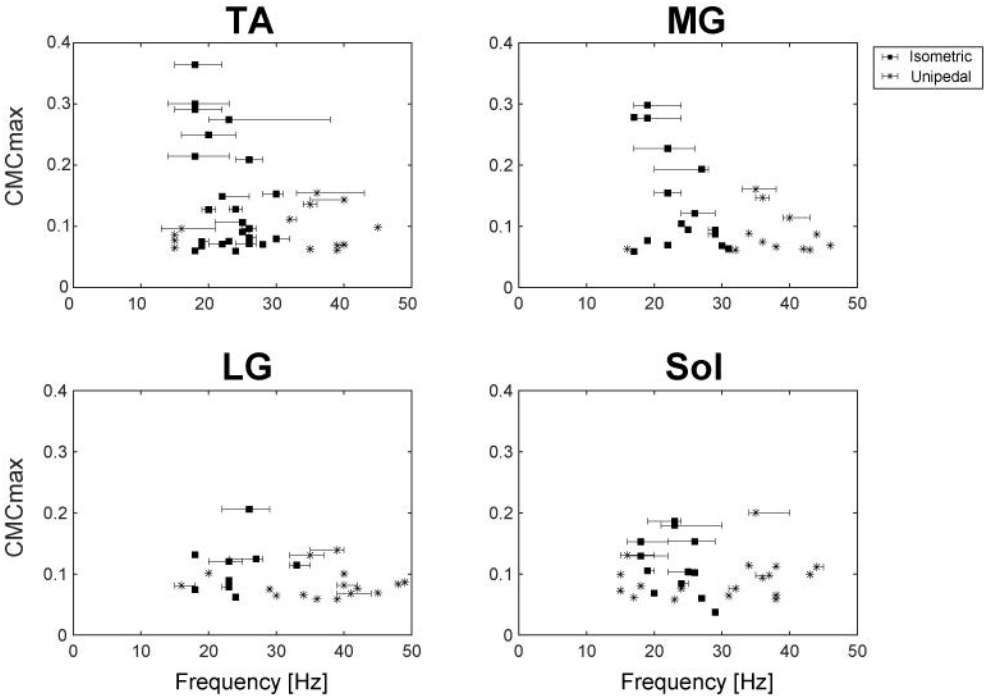

Mean Distance
MDIST[cm]
2
1.5
1
0.5
0
***
***
Bipedal
Unipedal dominant
Unipedal non-dominant

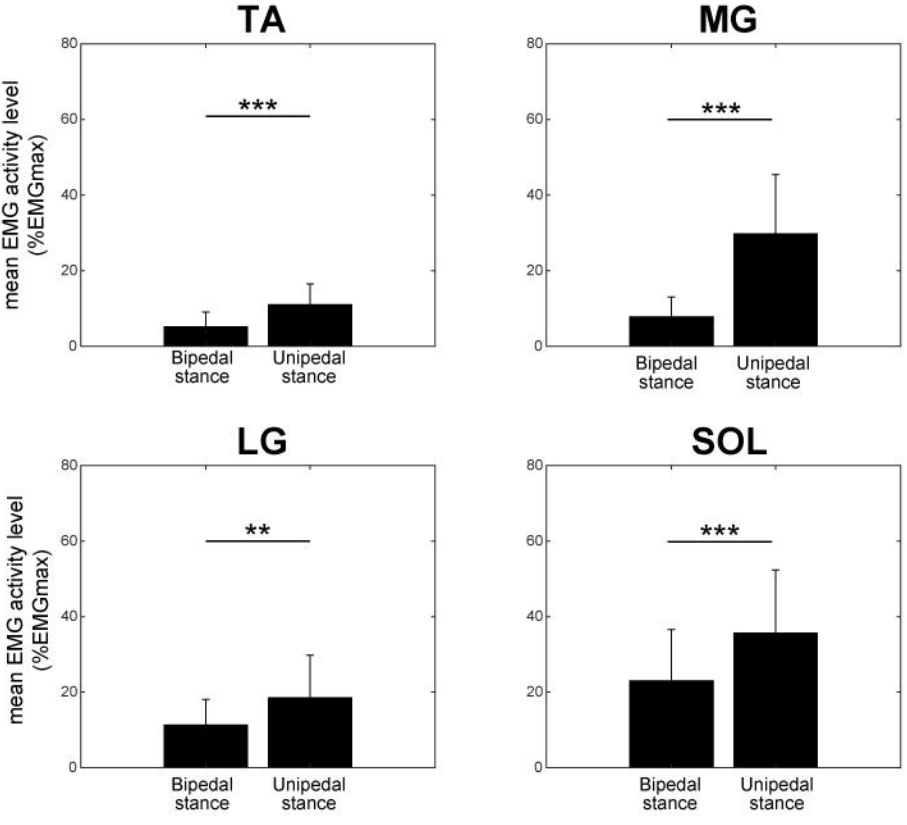

SD
SD[cm]
2
1.5
1
0.5
0
***
***
Bipedal
Unipedal dominant
Unipedal non-dominant


Mean Velocity
MVELO[cm/s]
3
2
1
0
***
***
Bipedal
Unipedal dominant
Unipedal non-dominant


Mean Frequency
MFREQ[Hz]
0.6
0.4
0.2
0
***
***
Bipedal
Unipedal dominant
Unipedal non-dominant


TA
MG
LG
SOL
CMCmax
0.3
0.2
0.1
0
**
*
Bipedal stance
Unipedal stance
Isometric contraction


TA
MG
LG
Sol
Isometric
Unipedal
CMCmax
CMCmax
Frequency [Hz]
Frequency [Hz]
0
0.1
0.2
0.3
0.4
0
10
20
30
40
50


TA
MG
LG
SOL
mean EMG activity level (%EMGmax)
0
20
40
60
80
***
**
Bipedal stance
Unipedal stance