Age-related changes in neuromuscular control of posture under unstable conditions

（不安定姿勢におけるバランス保持に関わる神経筋制御機構の加齢変化）

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リハビリテーション療法学専攻

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２０１９年度学位申請論文

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Abstract

[Background]

Prevention of falls is one of the major public health concerns in an aged society. It is suggested that an effective exercise intervention for preventing falls is a program consisting particularly of balance-challenging exercise, such as single-leg stance and forward postural leaning tasks. However, it is presumably unsafe and quite difficult for frail elderly adults to perform these high demanding tasks. The aim of this doctoral thesis was, therefore, to provide the foundational data that can be used to develop a safe and effective fall-prevention intervention for the frail elderly adults by examining the underlying neurophysiological mechanism of the age-related decline in those postural tasks.

In this doctoral thesis, we attempted to clarify the effect of aging on coordination of the plantar flexor muscles during the single-leg stance and forward leaning tasks, as these muscles are one of the main controllers of standing balance and strongly contribute to postural displacements. Specifically, we used an electromyography (EMG)-EMG coherence analysis that quantifies the common oscillatory input to motor neuron pools, and focused on delta-band (<5 Hz) coherence that reflects a similarity in the low-frequency discharge times between motoneuron pools, and beta-band (15-35 Hz) coherence that is suggested to be an indirect measure of the beta-band corticomuscular coherence and to reflect the corticospinal activity.

The delta-band coherence between the unilateral and bilateral homologous plantar flexor muscles can be observed during quiet standing and has been shown to be larger in the elderly than young adults. However, it is currently unknow how aging affects the coherence between

these muscles during the balance-challenging unstable postural tasks, in which the voluntary cortical control of posture is increased. To maintain balance during these postural tasks, it is necessary to hold the body in a steady state by voluntarily maintaining a steady force. Given that the larger delta-band coherence and smaller beta-band corticomuscular coherence could lead to the larger variability of force and thus possibly postural displacements, aging was hypothesized to influence the delta- and beta-band coherences between the plantar flexor muscles.

[Study 1: Single-leg stance]

Objective

The purpose was to investigate the effect of aging on the delta- and beta-band coherences between the unilateral plantar flexor muscles during single-leg stance.

Methods

Fourteen healthy young and seventeen healthy elderly adults performed bipedal and unipedal standing. We recorded center of pressure (COP) and EMGs unilaterally from the medial and lateral gastrocnemius (MG and LG) and soleus (SL) muscles. We analyzed variability of COP displacements and the EMG-EMG coherence for MG-LG, MG-SL, and LG-SL pairs in delta and beta bands.

Results

The delta-band coherence for the MG-SL pair and the beta-band coherences for all the pairs were larger during the unipedal than bipedal stance for both groups. The delta- and beta-band coherences for the MG-SL pair were larger for the elderly than young adults during the unipedal stance. The delta- and beta-band coherences were positively correlated to the variability of COP displacements in the elderly adults.

Summary

The coordination of the MG and SL muscles appears to be important for the control of single-leg stance, and aging can increase the delta- and beta-band coherences between these muscles during the single-leg stance. An increase in the delta- and beta-band coherences could be detrimental to the postural performance in the elderly adults. Alternatively, the increased beta-coherence may indicate a compensation for age-related decline in sensorimotor function.

[Study 2: Forward postural lean]

Objective

The low-frequency (<5 Hz) common input to the bilateral homologous plantar flexor muscles is reduced when increasing the cortical control by voluntarily leaning the body in young adults. However, it is unclear whether such modulation occurs in elderly adults. The purpose was to investigate the effect of aging on the delta-band EMG-EMG coherence between the bilateral homologous plantar flexor muscles and on the beta-band EMG-EMG coherence between the unilateral plantar flexor muscles during forward postural lean.

Methods

Fourteen healthy young and nineteen healthy elderly adults performed quiet standing task and tasks that required the subjects to lean their body forward to 35% and 75% of the maximum lean distance. We recorded EMGs from the bilateral MG and SL muscles. We analyzed the delta-band EMG-EMG coherence between the bilateral homologous muscles (MG-MG and SL-SL pairs) and also the beta-band EMG-EMG coherence between the unilateral muscles (MG-SL pair).

Results

The bilateral delta-band coherence was greater in the elderly than young adults in all the tasks. Importantly, the bilateral delta-band coherence for the MG-MG pair was smaller in the

75% forward lean than quiet standing and 35 % forward lean tasks for the young adults, and the bilateral delta-band coherence for the SL-SL pair was smaller in the 75% forward lean than 35% forward lean task for the young adults. Furthermore, the unilateral beta-band coherence was larger in the forward lean than quiet standing task for the young adults. Contrarily, significant changes were not observed in the elderly adults.

Summary

Aging may impair the ability to decrease the bilateral delta-band coherence and increase the unilateral beta-band coherence in the plantar flexor muscles during forward postural lean.

[Conclusion]

We found that aging influences the synchronous oscillations between the planar flexor muscles when attempting to maintain balance under unstable postural conditions. Although further investigations are necessary, it is possible that aging impairs the ability to functionally modulate the delta- and beta-band oscillations. Interventions that focus on these oscillations may, therefore, be effective in preventing the age-related decline in the ability to control posture under unstable conditions. This doctoral thesis provides the foundational data, potentially leading to future development of new fall-prevention intervention based on the neural oscillations.

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要旨

【背景】

超高齢社会を迎えた現在、高齢者の転倒は、非常に重要な社会問題となっている。そして、転倒予防には、片脚立位や前方へ重心を移動し傾斜姿勢を保持する課題など、バランス機能に負荷をかける運動が効果的とされる。しかし、これらの運動は、フレイル（虚弱）な高齢者にとっては負荷が高すぎる可能性があり、転倒や怪我の危険性を伴う。本研究は、片脚立位と前方傾斜姿勢を保持する能力が加齢により低下する神経生理学的機序の一端を解明することで、バランス機能に負荷をかける運動に取って代わる、安全で効果的な運動介入の開発につながる基礎的な情報を提供することを目的とした。

本研究では、立位姿勢の制御に強く関わる下腿三頭筋の協調的制御に加齢が与える影響の検証を試みた。具体的には、運動ニューロンプールへの共通の入力を定量化する筋電図間コヒーレンス解析法を用いて、運動ニューロンプールの同期的な活動を反映するデルタ帯（<5 Hz）コヒーレンス、及び皮質筋間の同期的神経活動を間接的に測定し皮質脊髄路の活動を反映するベータ帯（15-35 Hz）コヒーレンスに加齢が与える影響を検討した。

安静立位時には、一側、及び両側の下腿三頭筋間においてデルタ帯コヒーレンスが観察され、それは若年者に比べて高齢者の方が大きいことが報告されている。しかし、片脚立位や前傾姿勢など、皮質性制御の必要性が増大する不安定姿勢において、加齢が下腿三頭筋間の筋電図間コヒーレンスに与える影響は明らかとなっていない。不安定姿勢でバランスを保持するためには、随意的に発揮力を安定させることで姿勢の安定性を高める必要がある。デルタ帯コヒーレンスの増大と皮質筋間の同期的神経活動の低下が発揮力の安定性低下に関わることから、不安定姿勢における下腿三頭筋の筋電図間コヒーレンスは加齢により影響を受けると考えた。

【研究 1：片脚立位】

〈目的〉

　片脚立位時における立脚側の下腿三頭筋間のデルタ帯とベータ帯コヒーレンスに加齢が与える影響を明らかにすることを目的とした。

〈方法〉

　対象は、健常若年者 14 名と健常高齢者 17 名であった。課題は、両脚立位と片脚立位であった。足圧中心（center of pressure: COP）データを床反力計より取得し、その変動を解析した。また、筋電図信号を立脚側の内側腓腹筋（medial gastrocnemius: MG）、外側腓腹筋（lateral gastrocnemius: LG）、ヒラメ筋（soleus: SL）より取得し、デルタ帯とベータ帯の筋電図間コヒーレンスを MG-LG、MG-SL、LG-SL 間で解析した。

〈結果〉

　MG-SL 間のデルタ帯コヒーレンス、及び MG-LG、MG-SL、LG-SL 間のベータ帯コヒーレンスは、両脚立位に比べて片脚立位で大きかった。片脚立位において、MG-SL 間のデルタ帯及びベータ帯コヒーレンスは、若年者と比較して高齢者で大きかった。また、高齢者において、デルタ帯及びベータ帯コヒーレンスの増大は COP の変動増大と関連していた。

〈小括〉

　内側腓腹筋とヒラメ筋の協調的活動が片脚立位の制御に強く関わり、それらの筋間でのみデルタ帯及びベータ帯コヒーレンスが加齢により増大することが明らかとなった。また、高齢者において、デルタ帯、ベータ帯コヒーレンスの増大は姿勢の安定性低下につながる可能性が考えられた。ベータ帯コヒーレンスの増大は、加齢による感覚運動機能低下に対応するための代償的増大である可能性も考えられた。

【研究 2：前方傾斜姿勢】

〈目的〉

随意的な身体傾斜によって皮質性制御を増大する際、左右の下腿三頭筋の同期的活動（<5 Hz）は低下することが若年者において報告されている。しかし、高齢者においてこのような調整が生じるかは明らかでない。前方身体傾斜時における左右の下腿三頭筋間のデルタ帯コヒーレンスと、一側の下腿三頭筋間のベータ帯コヒーレンスに加齢が与える影響を明らかにすることを目的とした。

〈方法〉

対象は、健常若年者 14 名と健常高齢者 19 名であった。被験者は、両脚立位と身体を最大前方傾斜の 35%または 75%に傾ける課題（35%傾斜課題・75%傾斜課題）を行った。両側の MG と SL より筋電図信号を取得し、デルタ帯コヒーレンスを左右の MG-MG と SL-SL 間で、ベータ帯コヒーレンスを MG-SL 間で解析した。

〈結果〉

若年者において、MG-MG 間のデルタ帯コヒーレンスは、両脚立位と 35%傾斜課題に比べて、75%傾斜課題の方が小さかった。また、若年者において、SL-SL 間のデルタ帯コヒーレンスは、35%傾斜課題に比べて、75%傾斜課題の方が小さかった。さらに、若年者において、MG-SL 間のベータ帯コヒーレンスは、両脚立位に比べて傾斜課題で大きかった。一方、高齢者では、有意な変化は見られなかった。

〈小括〉

前方への身体傾斜姿勢の保持において、左右の下腿三頭筋間におけるデルタ帯コヒーレンスを低下させ、一側の下腿三頭筋間のベータ帯コヒーレンスを増大させる変調機能は加齢により低下する可能性が示唆された。

【結語】

本研究により、不安定姿勢においてバランスを保持する際に生じる下腿三頭筋の筋電図間コヒーレンスは加齢により変化することが明らかとなり、デルタ帯、ベータ帯コヒーレンスを機能的に変調する能力が加齢により低下する可能性が考えられた。より詳細な検証を必要とするが、これらの神経振動に着目した介入が不安定姿勢におけるバランス保持能力の改善に有効である可能性が示唆される。

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1. General Introduction

Owing to advances in medical care and improvements of living standards, the life expectancy has increased over the past decades, reaching a global average of 72 years in 2015 [1]. As a consequence, the elderly population has increased significantly and will continue to increase during the coming years: the percentage of the elderly aged 60 years or more is expected to double from 11 to 22 % between the years 2000 and 2050, and the number will be about 2 billion [2]. Although the increase in the life expectancy is desirable and a positive development, it is accompanied by increased risk of frailty and disability. Indeed, healthy life expectancy, which is defined as the period of life in good health, is reported to be a global average of 63 years [1], meaning that some sort of health care is needed for 9 years in average during life.

Falls are one of the major public health concerns among elderly adults, not only decreasing their quality of life but also increasing the medical and health care costs. One-third of community-dwelling elderly adults fall every year [3,4], and falls can be the cause of disability more often among the elderly aged 65 years [5,6]. Particularly, more than 90% of hip fractures are caused by falls [7], and 20 to 60 % of elderly adults who have experienced hip fracture require assistance for daily tasks for as long as 2 years after fracture [8]. This long-term care could also result in loss of independence and increased need of medical and health care. It is reported that fall-related injury in elderly adults are associated with admissions to nursing homes [9]. Therefore, preventing falls among elderly adults appears to be a key to reduce the gap between life expectancy and healthy life expectancy.

Interventions for preventing falls have been investigated in a number of previous studies. For example, vitamin D supplementation, with or without calcium, was suggested to reduce fall risk [10-12]. Also, home safety intervention (e.g., modification of home hazard) [13,14] and education about falls [15] appeared to decrease the occurrence rate of fall. Recent systematic reviews, however, concluded that these interventions do not decrease the fall risk [16,17], although the vitamin D supplementation and home safety intervention may be effective for elderly adults with low vitamin D level and visual impairments, respectively [17]. At present, an exercise program consisting of multiple components, particularly strength training and exercise that challenges balance control, appears to most effectively reduce the fall risk [17-21]. It is further concluded that the strength training alone does not prevent falls, and the balance-challenging exercise has to be included [17-21]. However, these studies involved relatively young and physically fit elderly subjects [20,21], and it is presumably unsafe and quite demanding for older and frail elderly adults to perform the balance-challenging exercise. Therefore, safe and effective fall-prevention interventions that can be used for the frail elderly population are required to be developed.

In clinical and community settings, single-leg stance task and forward leaning task during which the center of gravity is moved forward toward and held near the limits of stability are frequently used to challenge balance control [22,23]. Furthermore, impairments in performing these tasks are reported to be associated with future falls [24-27]. Indeed, falls are more likely to occur in daily activities where these movements are utilized (e.g., putting on a pair of pants and reaching out for something farther away). Thus, safe and effective fall prevention interventions for the frail elderly adults may be developed based on the underlying neurophysiological mechanism of the performance decline in those tasks. However, previous studies on the balance-challenging postural tasks have mainly examined biomechanical parameters, such as body sway amplitude and velocity, and to date, how aging affects the neurophysiological control of posture during the single-leg stance and forward leaning tasks has yet to be fully elucidated.

The plantar flexor muscles are one of the main controllers of standing balance and strongly contribute to postural displacements [28-30]. In this doctoral thesis, therefore, we attempted to clarify how aging affects coordination of these muscles during the single-leg stance and forward leaning tasks. Particularly, we focused on a coherence analysis of the electromyographic (EMG) signals recorded from these muscles. The EMG-EMG coherence analysis quantifies the common oscillatory input to two different parts of the same muscle (i.e., intramuscular coherence) or to two different muscles (i.e., intermuscular coherence) [31], and the synchronized oscillations in different motoneuron pools are suggested to originate from a common synaptic input to the spinal motoneurons [32]. The coherence in different frequency bands can reflect different physiological mechanisms. For example, the delta-band coherence (< 5 Hz) reflects a similarity in the low-frequency discharge times between motoneuron pools (i.e., low-frequency common drive to motoneuron pools) [33,34]. Alpha-band (5-15 Hz) coherence is associated with multiple factors (e.g., physiological tremor and spinal reflex) [35]. Beta-band (15-35 Hz) coherence is proposed to be an indirect measure of synchronous oscillations between the motor cortex and contracting muscle (i.e., corticomuscular coherence) and to reflect the corticospinal activity [36-39].

The coherence between the unilateral and bilateral homologous plantar flexor muscles has been consistently demonstrated in delta band in young and elderly adults during quiet standing [40,41]. Furthermore, this delta-band coherence has been shown to be larger in the elderly than young adults [41]. However, it is currently unknown how aging affects the coherence between these muscles during the unstable balance-challenging postural tasks, where the cortical control of posture is increased [42,43]. To maintain balance during these postural tasks, it is necessary to hold the body by voluntarily maintaining a steady force. Given that the variability of force during steady isometric contraction is attributed to the low-frequency (< 5Hz) oscillatory input to the motor neuron pool [44-46], it can be hypothesized that aging would increase the delta-band coherence between the plantar flexor muscles, consequently increasing the variability of postural displacements, even during the unstable balance-challenging postural tasks. Moreover, the beta-band synchronous oscillations between the motor cortex and contracting muscle are proposed to be associated with a steady and precise force output [44,47-50]. Thus, in addition to an increase in the beta-band coherence due to an increase in the cortical contribution, age-related change in the beta-band coherence would be expected as the variability of postural displacements is assumed to be larger in the elderly than young adults.

The aim of this doctoral thesis was to provide the foundational neurophysiological data and information that can eventually be used to develop a safe and effective fall-prevention intervention for the frail elderly adults who would not be able to perform the balance-challenging exercise. To this end, we examined the effect of aging on the delta- and beta-band synchronous oscillations between the plantar flexor muscles during single-leg stance and forward postural lean using the EMG-EMG coherence analysis. Results of this study will add to our understanding of the impact of aging on voluntary neuromuscular control of posture under unstable conditions, and this doctoral thesis has the potential to contribute to development of new fall prevention intervention.

2. Study 1: Single-leg stance

Main part of this study was published in Experimental Brain Research, entitled "Coordination of plantar flexor muscles during bipedal and unipedal stances in young and elderly adults" (2018, 236(5):1229-1239) [51].

2.1. Introduction

Even though bipedal stance has been largely focused on in most previous studies of postural control, many daily activities, such as putting on a pair of pants and striding over a large obstacle, require an ability to stand on one leg. Transferring from bipedal to unipedal stance reduces the base of support and consequently requires complicated motor skills to keep the center of mass (COM) within the base of support [52]. As the unipedal stance could sufficiently challenge postural control system, it may be more appropriate for assessment of balance function and a better predictor of future falls for the elderly [24], compared with the bipedal stance which has some limitations [53]. In spite of the clinical usefulness, however, relatively few studies have investigated the effects of aging on the motor control and strategies utilized during the unipedal stance [52].

Considering the human body as an inverted pendulum rotating at the ankle joint with the COM located in front of the joint [54], postural displacements during the bipedal stance occurs mainly in the anteroposterior (AP) direction, which is controlled primarily by the plantar flexor muscles [28,29]. On the other hand, postural displacements during the unipedal stance occurs in both AP and mediolateral (ML) directions, and it has been reported that the plantar flexor muscles are all potentially involved in AP and ML postural displacements [52]. To support this argument, several previous studies have identified moments in the direction other than plantar/dorsi flexion in these muscles: while medial gastrocnemius (MG), lateral gastrocnemius (LG), and SL muscles are all associated with inversion, the LG muscle also produces an eversion moment when ankle is everted [55,56]. Notwithstanding the importance of plantar flexors in postural control, however, aging has been reported not only to change their mechanical properties [57] but also to impair an ability to control force output using these muscles and increase the force variability [58], which could be related to deteriorated postural control [30]. Given that coordination of the plantar flexor muscles would be crucial for successful control of AP and ML postural displacements during standing, therefore, whether aging influences a way in which these muscles are coordinated to control a posture during the bipedal and unipedal stances needs to be clarified.

In the present study, we asked, using the EMG-EMG coherence analysis, how differently the unipedal stance is coordinated from the bipedal stance in terms of the shared neural inputs to the plantar flexor muscles and how aging affects the coordination strategy. We first identified how each of the plantar flexor muscles would be related to AP and ML postural displacements during bipedal and unipedal stances in young and elderly adults. Secondly, we calculated the coherence to evaluate the shared neural input between the plantar flexor muscles. Our hypothesis was that, considering the position of COM with respect to the base of support during the unipedal stance (i.e., anterolaterally located), the MG, LG, and SL muscles would all be involved in the ML postural displacements, but differently, in such a stance position. We also hypothesized that delta- and beta-band coherences would be larger during the unipedal than bipedal stance, as the greater variability of postural displacements and higher corticospinal drive would be expected during the unipedal stance. Considering the age effect on the variability of postural displacements, the delta-band coherence was further hypothesized to be larger in the elderly than young adults. Moreover, although the beta-band coherence is reported to be associated with the steadier force [49,50,59], we hypothesized that the beta-band coherence would be larger in the elderly than young adults based on a previous study showing an age-related increase in the beta-band corticomuscular coherence during a high demanding force matching task [60].

2.2. Methods

2.2.1. Subjects

Fourteen young (six females, mean age ± standard deviation (SD) = 22.7 ± 0.9, mean height ± SD = 163.1 ± 6.1) and seventeen elderly adults (eight females, mean age ± SD = 70.2 ± 3.4, mean height ± SD = 157.6 ± 8.9) participated in the present study. The young subjects were recruited from Nagoya University, and elderly subjects were recruited from the local community. The inclusion criteria for the elderly subjects were the following: 1) physically active and independent, 2) no history of any neurological, orthopedic, cognitive, or psychiatric problems that interfere the postural balance. None of the young subjects had the problems. All subjects gave written informed consent before participating the experiment. The institutional review board of Nagoya University approved the study (approval number: 16-518).

2.2.2. Experimental protocol

Each subject performed two tasks: bipedal and unipedal stance tasks. The order of the tasks was randomized among the subjects. In the bipedal stance task, the subjects were asked to stand quietly on a force plate for 40 s with their feet parallel and their arm along the sides. The feet were separated by 15 cm, and their positions were marked by a pen. In the unipedal stance task, the subjects stood on a force plate for 40 s with their dominant foot that was placed in the same position as the bipedal stance task. The foot was required to be kept on the same position during the whole task. The arms were along the sides, and the knee and hip angles were maintained as straight as possible. In both tasks, the subjects were asked to look at a fixation 1 m in front of them, and the examiners stood close to the subject in order to prevent falls and injuries during the task. 40 s of recording was started after we confirmed that the subject was in a physically stable state. When the task duration of 40 s was not achieved for the unipedal stance task, the task was repeated after a short break.

2.2.3. Recordings

Surface EMGs were recorded unilaterally from the MG, LG, and SL muscles of the dominant side using wireless EMG sensors (Tringo EMG sensors, DELSYS, Boston, MA, USA). The sensors were placed, according to the SENIAM recommendations (http://www.seniam.org/), as far as anatomically possible from each other in order to minimize the potential risk of cross-talk between the EMG recordings [61]. The subject's skin was gently abrased and cleaned with alcohol before the sensor placements. The EMG signals were amplified and filtered (band pass filter of 20-450 Hz) using a bio-amplifier (Tringo Wireless System, DELSYS, Boston, MA, USA). The sampling rate was set at 2000 Hz. Along with the EMG signals, the force signals were recorded from the force plate (Tec Gihan, Kyoto, Japan) at a sampling rate of 1000 Hz, digitized with an A/D converter, and stored on a personal computer for offline analysis, using a customized LabVIEW program (National Instruments, Austin, TX, USA). The EMG and force signals were synchronized (Trigger Module, DELSYS, Boston, MA, USA).

2.2.4. Data analysis

We analyzed the data during the middle 30 s of the collection period using a customized Matlab script (MathWorks, Natick, MA, USA). The AP and ML center of pressure (COP) signals, calculated from the forces and moments, were low-pass filtered at 15 Hz with a fourth-order zero phase lag Butterworth filter. We then calculated the SD and mean speed of

AP and ML COP postural displacements. The SD reflects the variability of postural displacements.

Initially, we preprocessed the EMG signals with a fourth-order zero phase lag Butterworth band-pass filter at 10-500 Hz to remove electrical noise and motion artifacts. To investigate the relationship between COP displacements and EMG activity of each of the plantar flexor muscles in the frequency domain, we applied the coherence analysis [62]. We further calculated the cumulant density that provides the time domain information analogous to cross correlation analysis [63]. Moreover, the coherence analysis was performed between the EMG recordings of the plantar flexor muscles to investigate the shared neural input to these muscles. The phase was also calculated to provide the time domain information.

Coherence analysis of the data was performed based on methods provided by Halliday and colleagues [64]. For the calculation of coherence between the COP displacements and EMG activity, we down-sampled the surface EMG signals to 1000 Hz before further analysis and applied the filtered COP signals. The surface EMG signals were initially full-wave rectified, as full-wave rectification has been shown to enhance the information about the temporal pattern of grouped firing motor units [64,65]. For two signals x and y (i.e., either COP signal and EMG signal or EMG and EMG signals), the coherence function was calculated using the following equation [66]:

$$|C_{xy}(f)|^2 = \frac{|P_{xy}(f)|^2}{P_{xx}(f) \cdot P_{yy}(f)} ,$$

where $P_{xy}(f)$ is the cross-spectra and $P_{xx}(f)$ and $P_{yy}(f)$ are the auto-spectra of the signals x and y, respectively, for a given frequency f, which were calculated with a discrete Fourier transform of non-overlapping segments of 1024 data points. The coherence function ranges from zero to one, with zero indicating that two signals are completely independent, and one indicating that two signals are identical.

In the present study, COP-EMG coherence was estimated between AP COP displacements and either MG, LG, or SL EMG activity, and between ML COP displacements and either MG, LG, or SL EMG activity. The EMG-EMG coherence was estimated in the following EMG pairs: MG-LG, MG-SL, and LG-SL.

To summarize and visualize the average differences in the coherence between tasks and age groups, we calculated the pooled coherence function (C_{pooled}) across k records (the number of subjects) at a given frequency f using the following equation [66]:

$$C_{pooled} = \left| \frac{\sum_{i=1}^{k} L_i C_{xy}^{i}(f)}{\sum_{i=1}^{k} L_i} \right|^2 ,$$

where C_{xy}^{i} is the individual coherence function and L is the number of segments.

2.2.5. Statistical analysis

The effects of task and age on the SD and mean speed of AP and ML postural displacements were assessed with a two-way (task × age group) repeated measures analysis of variance (ANOVA). For each coherence analysis, 95% confidence limit was applied to identify the significance of coherence. Similar to previous studies [40,41,67], we averaged z transformed EMG-EMG coherence over the frequency ranges of 0-5 Hz (delta) and 15-35 Hz (beta) to quantitatively compare the EMG-EMG coherence among tasks, EMG pairs, and age groups, as these frequency ranges were of our interests. The effects of task, EMG pair (MG-LG, MG-SL, and LG-SL), and age on the mean EMG-EMG coherence values were assessed with a three-way (task × EMG pair × age group) repeated measure ANOVA. We further computed the correlation between each of the averaged EMG-EMG coherence value and postural performance (i.e., SD and mean speed of AP and ML COP displacements) using the Pearson's correlation coefficients. A Greenhouse-Geisser correction was applied for sphericity, and post hoc comparisons were performed with Bonferroni's correction. The statistical analyses were conducted using SPSS (IBM, Armonk, NY, USA), and the significant level was set at 0.05.

2.3. Results

2.3.1. COP parameters

The SD and mean speed of AP and ML COP displacements for each group are given in Figure 2.1. A two-way repeated measure ANOVA on the SD of AP COP displacements revealed the main effects of task ($F_{1,29} = 130.6$, $p < 0.001$). Furthermore, the analysis on the SD of ML COP postural displacements indicated the main effects of task ($F_{1,29} = 386.3$, $p < 0.001$) and group ($F_{1,29} = 24.1$, $p < 0.001$), as well as their interaction ($F_{1,29} = 17.5$, $p < 0.001$). Post hoc analysis showed that the SD of ML COP displacements was greater for the elderly than young adults in the unipedal stance task ($p < 0.001$) but not in the bipedal stance task ($p = 0.161$). The SD was greater in the unipedal than bipedal stance task for both groups ($p < 0.001$).

A two-way repeated measure ANOVA on the mean AP COP speed revealed the main effects of task ($F_{1,29} = 161.1$, $p < 0.001$) and group ($F_{1,29} = 13.3$, $p = 0.001$), as well as their interaction ($F_{1,29} = 9.2$, $p = 0.005$). The interaction can be explained by a finding that the magnitude of difference between tasks was greater in the elderly than young adults although the value was significantly larger in the unipedal than bipedal stance task for both groups ($p < 0.001$). The analysis on the mean ML COP speed also demonstrated the main effects of task ($F_{1,29} = 175.3$, $p < 0.001$) and group ($F_{1,29} = 15.7$, $p < 0.001$), and their interaction ($F_{1,29} =$

15.9, $p < 0.001$). Post hoc analysis revealed that the mean ML COP speed was higher for the elderly than young adults in the unipedal stance task ($p < 0.001$) but not in the bipedal stance task ($p = 0.701$). The mean ML COP speed was higher in the unipedal than bipedal stance task for both groups ($p < 0.001$).

2.3.2. COP-EMG coherence

The pooled COP-EMG coherence spectra are presented in Figure 2.2. For the bipedal stance task, a significant coherence between the AP COP displacements and EMG activity of each of plantar flexor muscles was found up to about 10 Hz in the elderly adults, while it was demonstrated up to about 2 to 3 Hz in the young adults. The time shifts between signals, as calculated with the cumulant density analysis, are also presented in the Figure 2.2. The positive peak was identified with a negative time shift in all AP COP-EMG pairs. The positivity of the peak indicates that COP moves forward/backward as EMG activity increase/decrease; a negative time shift indicates that EMG activity occurs in advance of COP displacements. There was no clear coherence between the ML COP displacements and EMG activity of each of plantar flexors.

For the unipedal stance task, a significant coherence between the AP COP displacements and EMG activity of each of plantar flexor muscles was found up to about 10 Hz in the elderly adults, while it was revealed up to about 2 to 3 Hz for the MG and SL muscles and up to about 5 Hz for the LG muscle in the young adults. The cumulant density analysis further demonstrated a positive peak with a negative time shift in all AP COP-EMG pairs. Regarding the relationship between the ML COP displacements and EMG activity, a significant coherence was found up to about 5-10 Hz in all COP-EMG pairs in both groups. The cumulant density analysis indicated a positive peak with negative time shift in the MG and SL muscles. On the other hand, there was a negative peak with a negative time shift for the

LG muscle. The positivity of the peak indicates that COP moves lateral/medial as EMG activity increase/decrease; a negative time shift indicates that EMG activity occurs in advance of COP displacements.

2.3.3. EMG-EMG coherence

The pooled EMG-EMG coherence spectra are presented in Figure 2.3 to visualize the average differences, and the mean z-transformed delta- and beta-band coherence values are shown in Figure 2.4. A three-way repeated measure ANOVA on the delta-band coherence revealed the main effects of EMG pair ($F_{1.4,40.3} = 70.0$, $p < 0.001$) and group ($F_{1,29} = 14.2$, $p = 0.001$). There were also interactions between task and EMG pair ($F_{1.5,43.4} = 28.7$, $p < 0.001$) and between EMG pair and group ($F_{1.4,40.3} = 4.0$, $p = 0.040$). Post hoc analysis demonstrated that the delta-band coherence for the MG-SL pair was larger in the unipedal than bipedal stance task for both young ($p = 0.002$) and elderly ($p = 0.001$) adults. Furthermore, the delta-band coherence for the MG-SL pair was larger than the other two EMG pairs (MG-LG and LG-SL) in both tasks for the elderly adults ($p < 0.05$) and in the unipedal stance task for the young adults ($p < 0.001$). The delta-band coherences for all the pairs were larger for the elderly than young adults ($p < 0.05$) in the bipedal stance task. In the unipedal stance task, the coherence for the MG-SL pair was revealed to be larger for the elderly than young adults ($p = 0.025$), but not for the other two EMG pairs.

A three-way repeated measure ANOVA on the beta-band coherence revealed the main effects of task ($F_{1,29} = 69.1$, $p < 0.001$) and EMG pair ($F_{2,57.4} = 10.5$, $p < 0.001$). There was also significant interaction between task and EMG pair ($F_{1.7,50.7} = 13.8$, $p < 0.001$). Post hoc analysis showed that the beta-band coherences for all the EMG pairs were larger in the unipedal than bipedal stance task for both groups ($p < 0.05$), except for the LG-SL pair for the young adults ($p = 0.058$). Furthermore, the beta-band coherence for the LG-SL pair was larger than the MG-LG pair in the bipedal stance task for the young adults ($p = 0.017$), and the beta-band coherence for the MG-SL pair was larger than the MG-LG pair in the bipeadal stance task ($p = 0.009$), and was larger than the other two EMG pairs in the unipedal stance task for the elderly adults ($p = 0.002$ for MG-LG and $p < 0.001$ for LG-SL). Moreover, the beta-band coherence for the MG-SL pair was marginally larger for the elderly than young adults in the unipedal stance task ($p = 0.063$). There were no other differences between age groups.

Results of correlations between the mean coherence value and postural performance parameters were summarized in Table 2.1.

2.4. Discussion

The present study investigated how plantar flexor muscles were coordinated during bipedal and unipedal stances in young and elderly adults. The main findings were as follows. All the plantar flexor muscles contributed, but differently, to the ML COP displacements during the unipedal stance. Furthermore, the beta-band coherences for most of the EMG pairs were larger in the unipedal than bipedal stance task in both age groups. Interestingly, there was a feature specific to the MG-SL pair: the delta-band coherence for this pair was larger in the unipedal than bipedal stance task for both age groups and was also larger for the elderly than young adults in the unipedal stance task, and the beta-band coherence for this pair was larger than the other pairs in the unipedal stance task for the elderly adults. It appears that the oscillatory activity between the MG and SL muscles was strongly involved in the control of unipedal stance, and that aging selectively increased the delta- and beta-band oscillatory inputs to these muscles.

An analysis of a relationship between the COP displacements and EMG activity of each of plantar flexor muscles revealed that the MG, LG, and SL muscles were all involved in the

AP COP displacements during the bipedal and unipedal stances, and also in the ML COP displacements during the unipedal stance. In agreement with previous studies [28,29], the muscle activity was found to precede the *forward* COP displacement. Furthermore, it was demonstrated that the muscle activity precedes the ML COP displacement during the unipedal stance. Anticipatory activity of plantar flexor muscles [68] with reference to COP position appears to be necessitated in not only AP but also ML COP displacement. It is interesting to note here that the LG muscle was shown to precede the medial COP displacement, while the MG and SL muscles preceded the lateral COP displacement (Fig. 2.2). The MG and LG muscles, that are innervated by different nerves [7], could, thus, have distinct actions in the frontal plane [56], similarly to other muscles with multiple heads that generate asymmetrical activities [69].

The relationship revealed between the ML COP displacements and activity of plantar flexor muscles might also explain the significant increase in the delta-band coherence for the MG-SL pair and not for the MG-LG and LG-SL pairs when transferring from bipedal to unipedal stance. Given the differences in postural displacement direction that each muscle works for, the MG and SL muscles have likely worked cooperatively to produce force, while the LG muscle has worked less cooperatively and somewhat independently from the other two muscles. Furthermore, the transfer to the unipedal stance that assumedly requires the higher cortical demands than bipedal stance appears to have increased the corticospinal drive, reflected by the larger beta-band coherence [36]. As the beta-band coherence has been shown to potentially reflect not only descending but also ascending pathways [70], it is also possible that the sensory information from the peripheral has been integrated more extensively during the unipedal than bipedal stance. More importantly, it has been suggested that the ratio of the low-frequency (< 5Hz) common input that creates noisy force oscillations to the common input that is important for force control, possibly carried through the beta-band oscillations, could determine the force variability [45,46,71]. To strengthen this view, the beta-band oscillation or coherence has been reported to be associated with maintenance of a constant motor state or force output [48,72]. Considering the amount of increase in each of the delta- and beta-band coherences, it appears, therefore, that the common inputs to the cooperatively working MG and SL muscles consisting of a relatively greater amount of low-frequency noisy inputs have led to an increase in the variability of COP displacements during the unipedal stance.

The findings regarding the differences in the delta- and beta-band coherences between age groups are consistent with previous studies in which motor unit coherence and corticomuscular coherences were demonstrated to be larger in elderly adults during a force-matching task [60,73]. Although the greater beta-band coherence has been reported to result in the steadier force [49,50,59], an increase in the beta-band coherence has been shown to not necessarily lead to better force-matching performance in the elderly adults [60]. Their greater beta-band coherence during the unipedal stance may, thus, indicate a compensatory increase in the brain activity. They could have increased the corticospinal drive to the MG and SL muscles and/or have tried to integrate the sensory information from these muscles more extensively, to cope with the noisy forces generated cooperatively by these muscles, but the increase might have been dysfunctional (unsuccessful compensation [74]). Alternatively, the elderly adults could have been unable to disinhibit the corticospinal drive, given that smaller intracortical inhibition is associated with poorer performance in the elderly adults during a challenging postural task [75]. On the other hand, Jaiser and colleagues [76] have recently reported an insignificant difference in beta-band coherence between young and older adults using a task that required subjects to perform weak phasic voluntary contraction. Because the age-related difference in beta-band corticomuscular coherence has been observed predominantly during the force-matching task performed concurrently with a cognitive task [60], the types of task seem to play a role here. It may be likely that the age-related difference in the beta-band oscillation can be evident only during tasks imposing heavy demands, including the unipedal standing. This needs to be, however, clarified in future studies.

Considering the presumed contribution of motor unit and corticomuscular coherences to force variability [45,46], it was expected that the coherence would correlate with the variability of COP displacements. Interestingly, however, they were found to be correlated with each other primarily in the elderly adults (Table 2.1). It appears that the association of delta-band coherence with the variability has depended on the postural displacement direction. Also, the elderly adults with the greater variability have shown the larger beta-band coherence. As mentioned previously, the poor performance may have forced the elderly adults to increase the cortical activity, or they could have been unable to disinhibit the cortical activity that could cause unnecessary force production. The effect of aging on the beta-band coherence in relation to postural control seems not to be straightforward [60], and thus the additional investigation will be necessary. An important and interesting question here is why the correlation was present primarily in the elderly. One of the reasons would be ascribed to the overall generally smaller coherence and postural performance parameters in the young adults; the small variability could have made the correlation unlikely, given that the correlation was revealed in the analysis with two age groups combined. It is also possible that posture is controlled more passively by tissues around the ankle joint in young adults, as passive stiffness is reported to account for about 90 % of the ankle stabilization, although active control is certainly necessary [77]. The delta- and beta-band coherences, that supposedly reflect active cortical control of movement, therefore, were less directly associated with the postural performance parameters in the healthy young adults. A finding of apparently smaller coherence between the COP and EMG activity in the young adults would support this view (Fig 2.2).

There are some limitations to be mentioned. When analyzing EMG-EMG coherence, it is important to recognize a potential risk of cross-talk. There is one previous study proposing a novel method to mitigate the cross-talk [78], but it was based on an incorrect assumption that the cross-talk always enlarges the correlation between EMG signals (refer to Lowery et al. 2003 [79]). As stated by Farina and colleagues [80,81], no procedures to eliminate the cross-talk are currently available. In this study, we confirmed the absence of extremely high coherence over a wide frequency range (0-400Hz), that would be expected when cross-talk exists [61]; thus, we believe that the cross-talk was not significant. The study using the indwelling EMG, however, reported the absence of the LG muscle involvement in the control of bipedal stance [82], unlike the present study and the others using the surface EMG[28,29], which indicates the potential need for further studies. Moreover, we focused on the plantar flexor muscles in spite of a fact that more distal and proximal muscles also contribute to the postural control, especially during the unipedal stance. Similarly, we did not measure the angle of knee or hip. Although we asked the subjects to keep the knee and hip as straight as possible and confirmed that the subjects were physically stable without large body sways, the small angular movements at these joints as well as foot deformities [83] could have affected the postural performance parameters. In addition, given that Hoffman reflex is reduced during unstable posture [43], potentially through the Ia presynaptic inhibition controlled by descending commands, assessment of spinal circuitry may have strengthened our discussion about peripheral factors. Finally, the soleus muscle can be divided into several compartments; thus, the finding of the present study may be specific to the compartment that we have chosen to analyze.

In conclusion, we confirm the strong involvement of the oscillatory activity between the MG and SL muscles in the control of unipedal stance. Elderly adults increase the beta-band oscillatory input to these muscles, possibly to handle the postural sway that could be affected by forces generated cooperatively by them, or they have a reduced capability to disinhibit the beta-band oscillatory input. It is suggested that effective coordination of those two muscles is one of the keys for successful control of unipedal stance. We believe that this study would advance the understanding of the neuromuscular coordination strategies used during bipedal and unipedal stances and the effects of aging on the postural control.

2.5. Figures
2.5.1. Figure 2.1

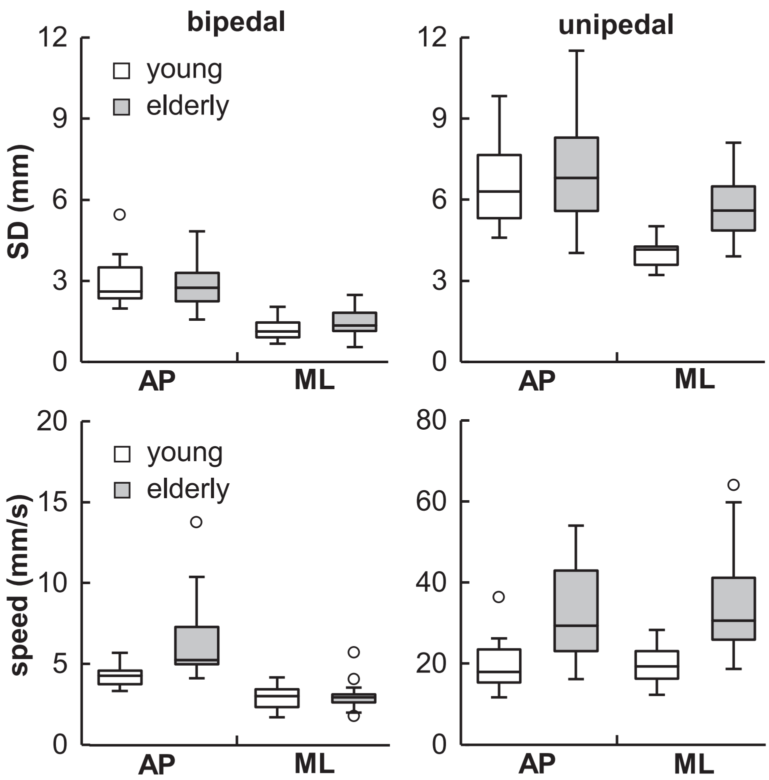

Fig. 2.1. Effects of stance and age group on center of pressure (COP) parameters. Data are shown separately for bipedal and unipedal stances and for each age group. The standard deviations (SDs) of anteroposterior (AP) and mediolateral (ML) COP displacements are presented on the top. The mean speeds of the AP and ML COP displacements are presented on the bottom. Small circles indicate outliners.

2.5.2. Figure 2.2

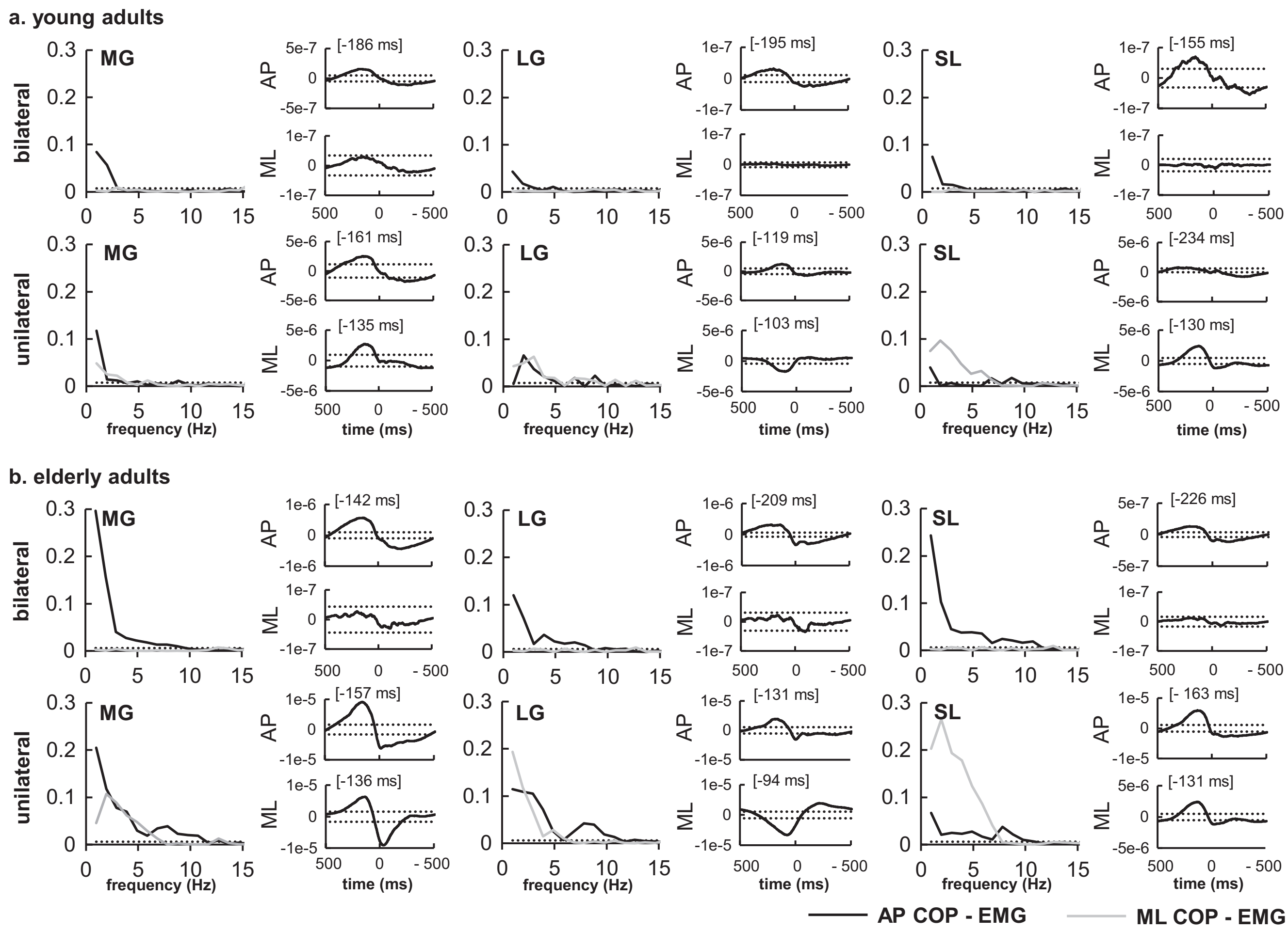

Fig. 2.2. Pooled coherence and cumulant density computed between center of pressure (COP) and electromyogram (EMG) of plantar flexor muscles. Data are presented separately for bipedal and unipedal stances and for each age group (a: young, and b: elderly). In the coherence spectra, black lines indicate the coherence between anteroposterior (AP) COP and EMG of either medical gastrocnemius (MG), lateral gastrocnemius (LG), or soleus (SL) muscle, and grey lines represent the coherence between mediolateral (ML) COP and EMG of either MG, LG, or SL muscle. Horizontal dashed lines indicate the 95% confidence limit. The cumulant density is presented next to the corresponding coherence spectrum separately for the AP and ML data. The numbers shown in brackets indicate lag times. Horizontal dashed lines represent the 95% confidence limit.

2.5.3. Figure 2.3

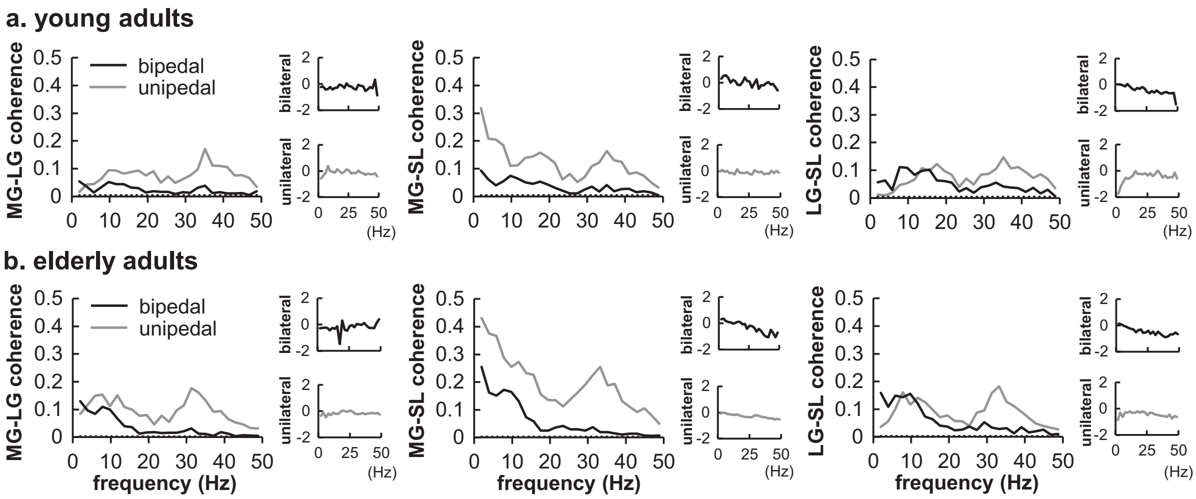

Fig. 2.3. Pooled coherence spectra during bipedal and unipedal stances. Data are presented separately for each age group (a: young, and b: elderly) and each electromyogram pair (medical gastrocnemius (MG) and lateral gastrocnemius (LG), MG and soleus (SL), and LG and SL). Black lines indicate the data for bipedal stance, and grey lines represent the data for unipedal stance. Horizontal dashed lines indicate the 95% confidence limit. The phase in radian was calculated to show the time domain information between the signals and illustrated next to the coherence spectrum.

2.5.4. Figure 2.4

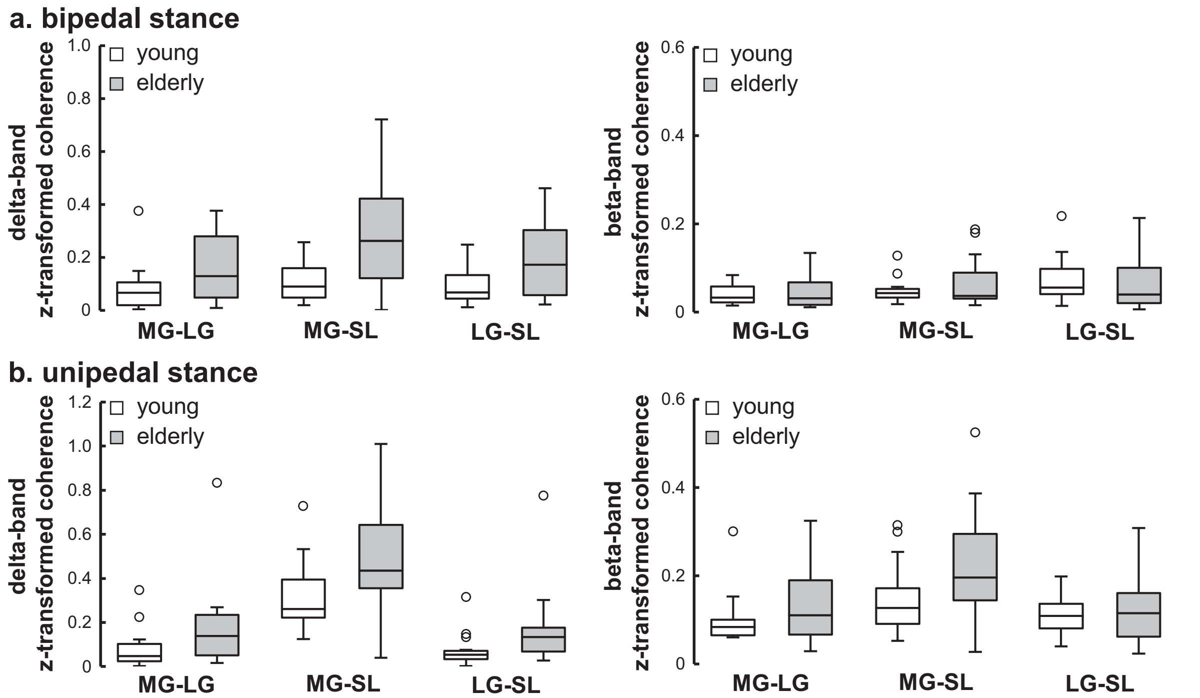

Fig. 2.4. Effects of stance and age group on mean z-transformed coherence values in delta and beta frequency bands. Data are shown separately for bipedal (a) and unipedal stances (b), for each age group, and for each electromyogram pair (from left to right: medical gastrocnemius (MG) and lateral gastrocnemius (LG), MG and soleus (SL), and LG and SL). Small circles indicate outliners.

2.6. Table

2.6.1. Table 2.1

Correlation coefficient (*r*) between EMG-EMG coherence and postural performance

bipedal

		young				elderly			
		AP		ML		AP		ML	
		SD	speed	SD	speed	SD	speed	SD	speed
delta	MG-LG	-0.44	-0.10	-0.38	-0.11	0.73*	0.39	-0.14	0.12
	MG-SL	0.09	-0.57*	-0.48	-0.27	0.84*	0.61*	-0.25	0.29
	LG-SL	-0.21	-0.10	-0.65*	-0.35	0.72*	0.31	-0.14	0.15
beta	MG-LG	0.09	-0.12	-0.51	-0.11	0.71*	0.51*	-0.24	0.26
	MG-SL	0.23	-0.32	-0.55*	-0.31	0.75*	0.56*	-0.22	0.26
	LG-SL	0.12	-0.10	-0.65*	-0.26	0.63*	0.59*	-0.36	0.54*

unipedal

		young				elderly			
		AP		ML		AP		ML	
		SD	speed	SD	speed	SD	speed	SD	speed
delta	MG-LG	0.35	0.52	0.10	0.66*	0.50*	0.55*	0.53*	0.71*
	MG-SL	-0.45	-0.14	0.16	0.24	0.68*	0.53*	0.56*	0.47
	LG-SL	0.37	0.50	-0.08	0.71*	0.75*	0.63*	0.56*	0.69*
beta	MG-LG	0.07	0.19	0.67*	0.41	0.42	0.53*	0.53*	0.46
	MG-SL	0.13	-0.29	0.48	-0.18	0.33	0.32	0.40	0.16
	LG-SL	0.24	0.08	-0.20	-0.16	0.45	0.52*	0.61*	0.50*

young and elderly combined

		bipedal				unipedal			
		AP		ML		AP		ML	
		SD	speed	SD	speed	SD	speed	SD	speed
delta	MG-LG	0.19	0.43*	-0.10	0.06	0.45*	0.60*	0.53*	0.71*
	MG-SL	0.44*	0.63*	-0.09	0.17	0.31	0.49*	0.59*	0.54*
	LG-SL	0.33	0.43*	-0.11	0.05	0.63*	0.65*	0.53*	0.71*
beta	MG-LG	0.45*	0.42*	-0.27	0.16	0.32*	0.49*	0.54*	0.48*
	MG-SL	0.51*	0.49*	-0.22	0.11	0.24	0.32	0.51*	0.28
	LG-SL	0.40*	0.37*	-0.46*	0.23	0.39	0.42*	0.41*	0.38*

* significant correlation ($p < 0.05$).
AP: anteroposterior; ML: mediolateral; MG: medial gastrocnemius; SL: soleus; LG: lateral gastrocnemius.

3. Study 2: Forward postural lean

Main part of this study was published in Frontiers in Human Neuroscience, entitled "Age-related declines in the ability to modulate common input to bilateral and unilateral plantar flexors during forward postural lean" (2018, 12:254) [84].

3.1. Introduction

Functional tasks in daily activities, such as reaching out for an item, often times require shifting of the weight toward the edge of the base of support. Leaning of the body imposes higher demands on the postural control system, and age-related general declines in physical function commonly lead to difficulty in performing tasks involving weight shifting. It has been reported that elderly adults show less controlled lean path and greater sway at the maximal lean position, when compared to young adults [85]. Furthermore, falls frequently occur during leaning or reaching [86], and a reduced ability to lean the body forward is associated with a future fall risk [25]. Despite the widespread use of the leaning task as an assessment tool of balance dysfunction in clinical and community settings [25,26], the underlying neurophysiological mechanism of the age-related impairments has not been fully elucidated yet.

Control of posture is a complicated neuromuscular mechanism requiring effective and efficient activation of postural muscles, and a number of previous studies have explored how plantar flexor muscles are coordinated during standing as the COM locates in front of the ankle joint [54]. It has been reported that the bilateral SL muscles receive greater common input during standing than voluntary contraction [87,88]. Furthermore, recent studies have demonstrated coherence of the EMG signals between bilateral homologous plantar flexor muscles in delta band (0-5 Hz) during quiet standing [40,41]. The delta-band coherence can reflect comodulation of muscle activity [33,34], and the origin of this bilateral comodulation during standing is suggested to be the subcortical system [40,87]. In contrast to quiet standing that requires relatively little cortical control [89], challenging postural tasks can induce the cortical activity [42]. Indeed, cortical control of posture is greater during standing on a foam than rigid surface [90] and also during unsupported than supported forward postural lean [91]. Moreover, increasing the cortical contribution to postural control by voluntarily swaying the body forward and backward can result in a reduction in the comodulation of bilateral plantar flexor muscles [87]. It can be, therefore, expected that increasing the cortical contribution by leaning the body forward would similarly decrease the bilateral comodulation. Although it is well-recognized that elderly adults have impairments in reducing the synchronous bilateral activation of limb muscles [92,93], how aging impacts a way in which bilateral plantar flexor muscles are coordinated during the forward postural lean has not been examined previously.

Accordingly, the purpose of the present study was to investigate the effect of aging on the modulation of common input to the bilateral and unilateral plantar flexor muscles when changing the postural position between quiet standing and forward leaning, using the coherence analysis. More specifically, we examined the previously identified delta-band coherence between the bilateral homologous plantar flexor muscles (bilateral coherence) [40,41], and also assessed the beta-band (15-35 Hz) coherence within the unilateral plantar flexor muscles (unilateral coherence), that reflects the corticospinal drive to the contracting muscles [36]. We hypothesized the bilateral delta-band coherence to be smaller and the unilateral beta-band coherence to be larger during the forward postural lean than quiet standing, and that the modulation would be smaller in the elderly than young adults.

3.2. Methods

3.2.1. Subjects

Fourteen young (six females, mean age ± SD = 22.6 ± 0.9) and nineteen elderly adults (eight females, mean age ± SD = 70.1 ± 3.3) participated in the present study. We recruited the young adults from Nagoya University and the elderly adults from the local community. None of the subjects had neurological, orthopedic, cognitive, or psychiatric problems that influence the postural balance. They also had normal or corrected-to-normal vision, and were all right-foot dominant, which was determined by asking the predominant foot used for kicking a ball. The ethics committee of Nagoya University approved this study (approval number: 16-518), and all subjects gave written informed consent before participating the experiment. The experiment was conducted in accordance with the Declaration of Helsinki.

3.2.2. Tasks

Figure 3.1 depicts an illustration of the experimental setup and a flow chart of this experiment. There were three tasks to be performed for each subject: quiet standing and two tasks requiring the subjects to lean their body forward to 35% and 75% of the maximum lean distance (35% and 75% forward lean tasks). Before beginning the tasks, we assessed the maximum forward lean distance. The subjects were asked to stand on a force plate with their feet parallel to each other (heel-to-heel distance of 15 cm), and the feet positions were marked and kept constant during the whole experiment. They were further instructed to lean their body forward by dorsiflexing the ankle joints while maintaining the rest of their body straight. The greatest lean distance, that was determined based on the COP displacement, out of three trials was adopted as the maximum lean distance and used to calculate a distance to lean in the 35% and 75% forward lean tasks. In the quiet standing task, the subjects were asked to stand quietly and look at a fixation sign on a PC monitor set 1 m in front of them. In the 35% and 75% forward lean tasks, the subjects were instructed to lean as when the maximum lean distance was determined. A green horizontal target line representing a 35% or 75% of the maximum forward lean and two yellow horizontal lines at +5 % and -5% of the target line were displayed on the PC monitor. The subject's COP position was also presented on the PC monitor as a red line progressing from left to right, which moved upward/downward as the subject leaned forward/backward. The subjects were asked to keep their COP on the green target line as accurately and consistently as possible. The task duration was 40 s, and the order of the tasks was randomized among the subjects. Two practice trials were performed before each forward lean task to familiarize the subjects with the tasks.

3.2.3. Data acquisition

Wireless EMG sensors (Trigno EMG sensors, DELSYS, Boston, MA, USA) were placed on the bilateral MG and SL muscles according to the SENIAM recommendations (http://www.seniam.org/), after the skin was gently abrased and cleaned with alcohol, as these two muscles are mainly involved in postural control [29,82]. The sensors were placed as far as anatomically possible from each other to minimize the potential risk of cross-talk between the EMG recordings [61]. EMG signals were amplified and filtered (band pass filter of 20-450 Hz) using a bio-amplifier (Trigno Wireless System, DELSYS, Boston, MA, USA), and sampled at 2000 Hz. Force signals were also recorded using a force plate (Tec Gihan, Kyoto, Japan) at a sampling rate of 1000 Hz to compute the COP position. A customized LabVIEW program (National Instruments, Austin, TX, USA) was used to display the horizontal lines and COP position on the PC monitor.

3.2.4. Data analysis

Examples of the anteroposterior COP displacement data are shown in Figure 3.2. The data during the middle 30 s of the collection period were analyzed using a customized Matlab script (MathWorks, Natick, MA, USA). After low-pass filtering the anteroposterior COP signals at 15 Hz with a fourth-order zero phase lag Butterworth filter, we calculated the SD and mean speed of anteroposterior COP displacement. The SD reflects the variability of postural displacement. For the forward lean tasks, we additionally calculated the average of absolute difference between the COP position and the target line to quantify the error from the target.

Coherence between the EMG recordings was computed based on methods provided by Halliday and colleagues [64] to quantify common input to the plantar flexor muscles. Reproducibility of coherence analysis was investigated in several studies [94,95], and the analysis was found to be quite reliable; however, large changes are needed to demonstrate a real difference [95]. We initially rectified the EMG signals, because rectification of EMG signals can increase the information about temporal firing pattern of motor unit pools [64,65,96]. We them computed the coherence function using the following equation.

$$|C_{xy}(f)|^2 = \frac{|P_{xy}(f)|^2}{P_{xx}(f) \cdot P_{yy}(f)}$$

$P_{xx}(f)$ and $P_{yy}(f)$ are the auto-spectra of the signals x and y, and $P_{xy}(f)$ is the cross-spectra at the frequency f. They were calculated with a discrete Fourier transform of non-overlapping segments of 1024 data points. The coherence function is a number ranging from zero to one: zero indicates that two signals are completely independent, and one indicates that two signals are identical. For each analysis, 95% confidence limit was applied to identify the significant coherence. In the present study, the coherence was estimated in the following muscle pairs: right MG and left MG (MG-MG), right SL and left SL (SL-SL), and right MG and right SL (MG-SL). We analyzed the unilateral coherence in the dominant right leg because it was expected to be used more for the control of forward lean posture than the non-dominant left leg.

We also calculated the pooled coherence function to summarize and visualize the average difference among tasks and age groups using the following equation [64,97].

$$C_{pooled} = \left| \frac{\sum_{i=1}^{k} L_i C_{xy}^{i}(f)}{\sum_{i=1}^{k} L_i} \right|^2$$

C_{xy}^{i} is the coherence at the frequency f from individual subject, L_i is the number of segments, and k is the number of subjects.

3.2.5. Statistical analysis

The effects of task and age on the COP parameters (SD and mean speed of COP displacement and error from the target) were assessed with a two-way (task × age) repeated measures analysis of variance (ANOVA). Similar to previous studies [40,41,67], we averaged z-transformed coherence over the frequency ranges of 0-5 Hz (delta) for the bilateral coherence and 15-35 Hz (beta) for the unilateral coherence, to quantitatively compare the coherence among the tasks and age group. The effects of task and age on each of the coherence values were assessed with a two-way (task × age) repeated measure ANOVA. A Greenhouse-Geisser correction was applied for sphericity, and post hoc analysis and planned comparison analysis between the tasks within each group were performed with Bonferroni's correction. We additionally analyzed correlation between either the SD or mean speed of COP displacement and each of the averaged coherence values using the Pearson's correlation coefficients. The statistical analyses were conducted using R (R Development Core Team, Vienna, Austria) at the significant level of 0.05.

3.3. Results

3.3.1. COP parameters

Results of the COP parameters are presented in Figure 3.3. A two-way repeated measure ANOVA on the SD of COP displacement revealed main effects of task ($F_{1.2,38.4} = 5.60$, $p = 0.017$) and age ($F_{1,31} = 6.0$, $p = 0.020$) as well as their interaction ($F_{1.2,38.4} = 8.1$, $p = 0.0044$). Post hoc analysis indicated that the SD was significantly smaller in the 35% ($p = 0.039$) and 75% ($p = 0.0012$) forward lean tasks than quiet standing task for the young adults. There were significant differences between age groups in the 35% ($p < 0.001$) and 75% ($p = 0.0024$) forward lean tasks. An analysis on the mean speed of anteroposterior COP displacement demonstrated main effects of task ($F_{1.2,37.1} = 76.5$, $p < 0.001$) and age ($F_{1,31} = 9.3$, $p = 0.0047$). Post hoc analysis showed that the mean speed was significantly higher in the 75% forward lean task than 35% forward lean and quiet standing tasks for both groups ($p < 0.001$). Also, it was significantly higher in the 35% forward lean than quiet standing task for both groups ($p < 0.001$). An analysis on the error from the target line revealed main effects of task ($F_{1,31} = 28.7$, $p < 0.001$) and age ($F_{1,31} = 28.8$, $p < 0.001$).

3.3.2. EMG-EMG coherence

The pooled coherence spectra for the bilateral and unilateral coherences are shown in Figure 3.4 to visualize the difference among tasks and age groups. The mean z-transformed delta- and beta-band coherence values are presented in Figure 3.5. A two-way repeated measure ANOVA on the delta-band coherence for the MG-MG pair revealed main effects of task ($F_{1.5,45.6} = 6.3$, $p = 0.008$) and age ($F_{1,31} = 13.4$, $p < 0.001$). Planned comparison analysis demonstrated that the delta-band coherence for the MG-MG pair was significantly smaller in the 75% forward lean task than 35% forward lean ($p = 0.0011$) and quiet standing tasks ($p = 0.036$) for the young adults. An analysis on the delta-band coherence for the SL-SL pair also revealed main effects of task ($F_{1.5,47.5} = 6.3$, $p = 0.008$) and age ($F_{1,31} = 11.5$, $p = 0.002$). Planned comparison analysis showed that the delta-band coherence for the SL-SL pair was significantly smaller in the 75% forward lean than 35% forward lean task ($p = 0.027$) for the young adults. A two-way repeated measure ANOVA on the beta-band coherence for the MG-SL pair indicated a main effect of task ($F_{1.7,51.9} = 17.1$, $p < 0.001$) and an interaction between task and age ($F_{1.7,51.9} = 4.9$, $p = 0.016$). Post-hoc analysis demonstrated that the beta-band coherence for the MG-SL pair was larger in the 35% and 75% forward lean than quiet standing task (35%: $p < 0.001$; 75%: $p = 0.029$) for the young adults. There was no statistically significant difference between the tasks for the elderly adults. Thus, we performed post hoc power analysis for detecting difference between the quiet standing and 75% forward lean task in the elderly adults using GPower [98] and found post hoc power estimates of 0.55 for the delta-band coherence for the MG-MG pair, 0.66 for the delta-band coherence for the SL-SL pair, and 0.61 for the beta-band coherence for the MG-SL pair.

There was positive correlation between the SD of COP displacement and the delta-band coherence for the SL-SL pair ($r = 0.56$, $p = 0.038$) in the 75% forward lean task for the young adults. For the elderly adults, the SD and speed of COP displacement were both positively correlated with the delta-band coherence for the SL-SL pair in the quiet standing task (SD: $r = 0.51$, $p = 0.027$; speed: $r = 0.51$, $p = 0.025$). The speed of COP displacement was also correlated positively with the beta-band coherence for the MG-SL pair in the 35% ($r = 0.53$, $p = 0.019$) and 75% forward lean tasks ($r = 0.54$, $p = 0.018$) for the elderly adults.

3.4. Discussion

The present study investigated, using the coherence analysis, the common input to the bilateral and unilateral plantar flexor muscles during quiet standing and forward postural lean in young and elderly adults. Main results indicated that the bilateral delta-band coherence was significantly smaller in the 75% forward lean task as compared to the other tasks for the young adults. Also, the unilateral beta-band coherence was found to be significantly larger in the 35% and 75% forward lean than quiet standing task for the young adults. Contrarily, such changes were not significant in the elderly adults. It is likely that elderly adults have difficulty in modulating the common inputs to bilateral and unilateral plantar flexor muscles when leaning the body forward.

Because corticomuscular coherence has been reported mainly in the beta band [36,99], intermuscular coherence within one limb in this frequency range has been suggested to reflect the corticospinal drive [37]. Furthermore, it was recently demonstrated that the intermuscular beta-band coherence could reflect not merely the corticospinal drive but cortical control over synergistically working muscles (i.e., muscle coordination) [67,100]. As leaning the body forward supposedly requires higher cortical demands, it was expected that the unilateral beta-band coherence would be larger during the forward lean than quiet standing task. It appears that the young adults increased cortical control of the MG and SL muscles during the forward lean tasks, and the increase consequently resulted in a decrease in the SD of COP displacement, as greater beta-band oscillation and coherence are associated with steadier force output [48,72]. On the other hand, the modulation by the elderly adults was not significant. In a study by Papegaaij and colleagues [91], they suggested that intracortical inhibitory activity could be lower during unsupported than supported forward leaning in both age groups. In addition, the thread/fear of losing balance was suggested to be an important factor modulating the cortical activity because the modulation was greater as the center of mass moved closer to the edge of the base of support [91]. Thus, it can be proposed from the present and previous findings that, although an increase in difficulty of postural task and the associated fear of falling can modulate the cortical activity in elderly adults, they have a reduced ability to cortically coordinate the synergistically working muscles as a functional unit [67,100]. Because aging has been reported to increase the beta-band corticomuscular coherences [60], this study's findings may be specific to tasks that involve the bilateral activation. The capability to increase the beta-band coherence may, hence, depend on the magnitude of bilateral comodulation of plantar flexor muscles, as described below.

In agreement with previous observations [40,41], we found the delta-band coherence between the bilateral homologous plantar flexor muscles during quiet standing, and this bilateral comodulation was greater in the elderly than young adults in all tasks. An important and interesting finding was the smaller delta-band coherence for the MG-MG and SL-SL pairs in the 75% forward lean task for the young adults. It appears that the young adults decreased the bilateral comodulation while increasing the cortical control of the plantar flexor muscles when leaning the body forward close to the edge of the base of support. The decrease in the bilateral comodulation could also indicate that the planter flexors were controlled more (or relatively) unilaterally. On the other hand, the elderly adults might have had difficulty in operating such a switch from bilateral to unilateral control, as evidenced by an age-related impairment in the ability to reduce in-phase coordination during a bilateral task [93,101]. Indeed, the bilateral coherence is larger during in-phase than antiphase coordination [102]. Similar to when performing tasks involving upper and lower limb coordination [103,104], the present age-related decline in the modulatory ability might be ascribed to inhibitory dysfunction. It is also possible that their relatively larger delta-band coherence made it harder to modulate the bilateral and unilateral coherences.

Correlation analysis between the COP parameter and the strength of coherence revealed different patterns of correlation among the age groups. The positive correlation for the bilateral delta-band coherence during quiet standing in the elderly adults may indicate that the bilateral low-frequency oscillations influenced the force output variability [46] and thus variability of postural displacements. More importantly, in the 75% forward lean task, there was positive correlation between the SD of COP displacement and the bilateral delta-band coherence for the SL-SL pair in the young adults. In a recent study by de Vries and colleagues [105], the bilateral coherence was reported to depend on the degree of bilateral coordination required for the task and become smaller with the slighter bilateral coordination. Furthermore, they demonstrated the larger beta-band corticomuscular coherence with the slighter bilateral coordination [105], suggesting an inverse relationship between the bilateral and unilateral coherences. The present results including the larger beta-band coherence in the forward lean than quiet standing task, thus, likely propose that the young adults employed a strategy shifting from the synchronous bilateral activation to unilateral cortical control of plantar flexor muscles during the 75 % forward lean task, in order to realize the better performance, supporting the above-mentioned argument. On the other hand, the mean COP speed was correlated positively with the unilateral beta-band coherence in the forward lean tasks for the elderly adults. As the larger beta-band coherence has been reported not to necessarily lead to the better force-matching performance in the elderly adults [60], their increase in the corticospinal drive to cope with difficult tasks might have been dysfunctional. The role of the beta-band oscillation during a bilateral coordination appears to differ between young and elderly adults but should be clarified in future studies with more sophisticated experimental manipulations [105].

Significant differences in the mean COP speed and the error from the target line between two forward lean tasks expectedly indicate that the difficulty of the task became greater as the lean distance became larger. With respect to the coherence values, the significant decrease in the delta-band coherence was observed primarily in the 75% forward lean task. In contrast, the magnitude of increase in the beta-band coherence was not different between two forward lean tasks. These two observations may imply that the modulation of cortical activity is a primary strategy employed during a slight body leaning, and the magnitude of bilateral comodulation becomes an important factor as the body moves close to the edge of the base of support. Not merely the task difficulty but the fear of falling, therefore, are likely associated with the bilateral comodultion of plantar flexor muscles.

This study has several limitations. First, this study included planned comparisons and was slightly underpowered to examine a difference in the coherence between the postural tasks in the elderly adults. Further studies with a larger sample are needed to confirm this study's findings. Second, we did not measure the angle of the hip, knee, or ankle joint, or activity of upper-leg or back muscles that could be involved in maintaining a forward postural lean position. As aging can reduce muscle strengths and cause joint degeneration, consideration of these factors would have strengthened our discussion. Third, visual feedback of the COP position that was provided in the forward lean tasks to keep the lean distance constant might have had some influence on the present results, although its effect on postural control still remains under debate [106,107]. How the visual feedback affects the common input to postural muscles should be confirmed in future research. Fourth, there is a possibility that the modulation of beta-band coherence would be smaller in the non-dominant than dominant leg. Finally, we estimated the corticospinal activity using the coherence analysis of surface EMG. Although there are numerous studies evaluating the corticospinal activity using this method [31,61,108-111], inclusion of intramuscular single-unit and electroencephalography recordings would have enhanced the study's conclusion.

In summary, this study demonstrates the importance of decreasing the delta-band coherence between bilateral homologous plantar flexor muscles and increasing the beta-band coherence among unilateral plantar flexor muscles when maintaining a forward postural lean position. Aging further appears to impair such a modulatory ability. High-quality forward leaning performance likely requires a smooth shift from more synchronous bilateral activation to more unilateral cortical control of plantar flexor muscles. Therefore, interventions focusing on this factor may be beneficial for improving voluntary control of forward lean posture.

3.5. Figures

3.5.1. Figure 3.1

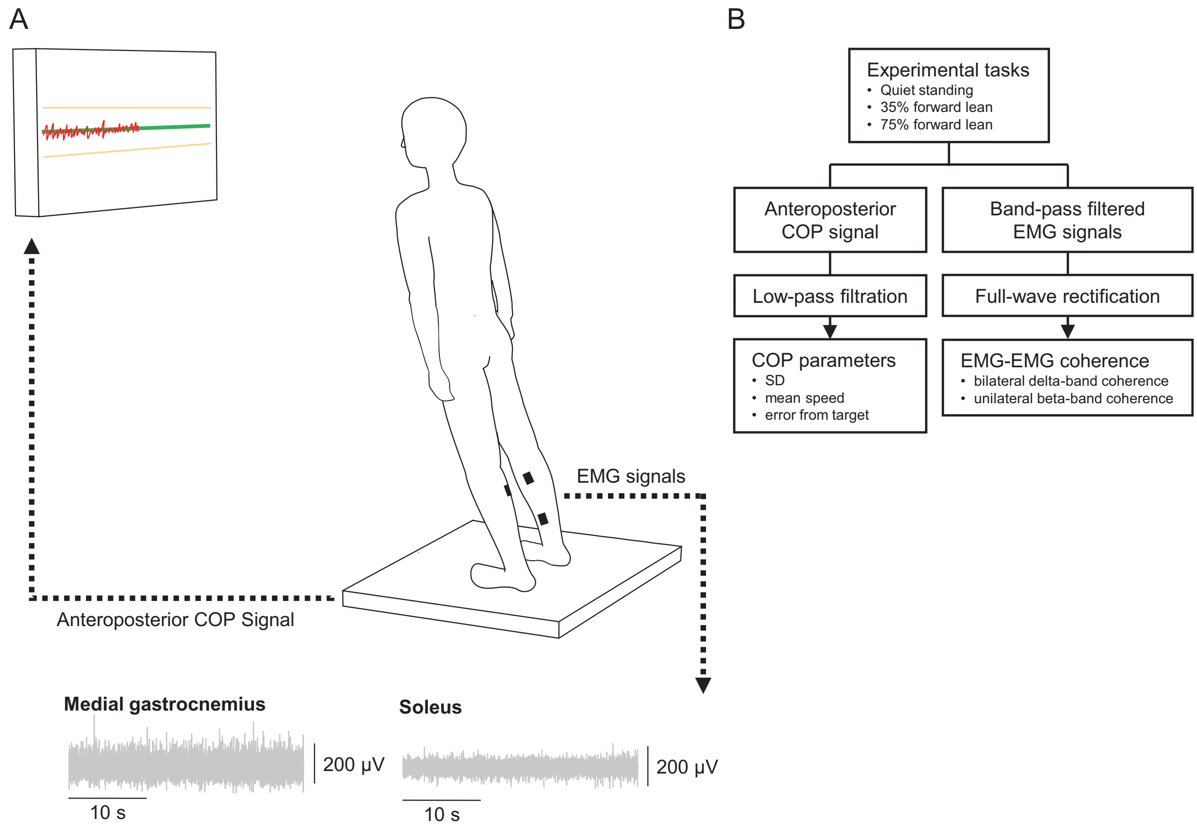

Fig. 3.1. Illustration of the experimental setup (A) and flow chart of the experiment (B). In forward lean tasks the subject stood on a force plate and was instructed to lean the body forward by dorsiflexing the ankle joint and keep the center of pressure (COP: red line) on the target line (green line) as accurately and consistently as possible. We recorded electromyograms (EMGs) from the bilateral medial gastrocnemius and soleus muscles using wireless sensors.

3.5.2. Figure 3.2

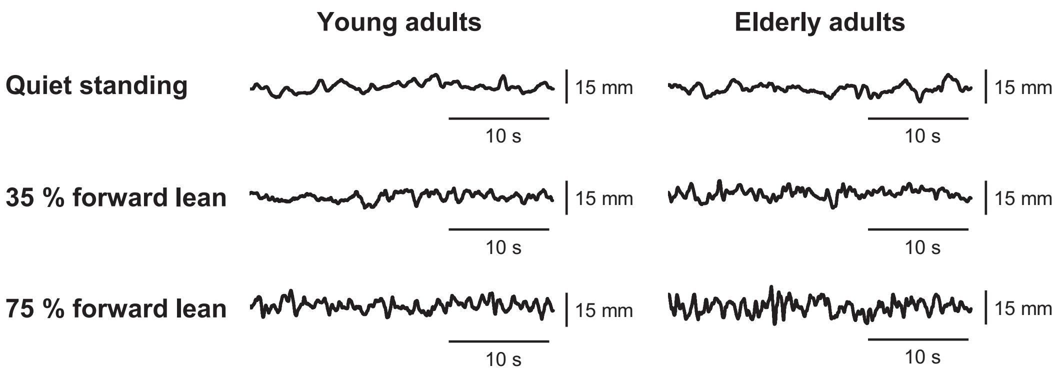

Fig. 3.2. Example data of anteroposterior center of pressure (COP) displacement from an individual young adult and an individual elderly adult.

3.5.3. Figure 3.3

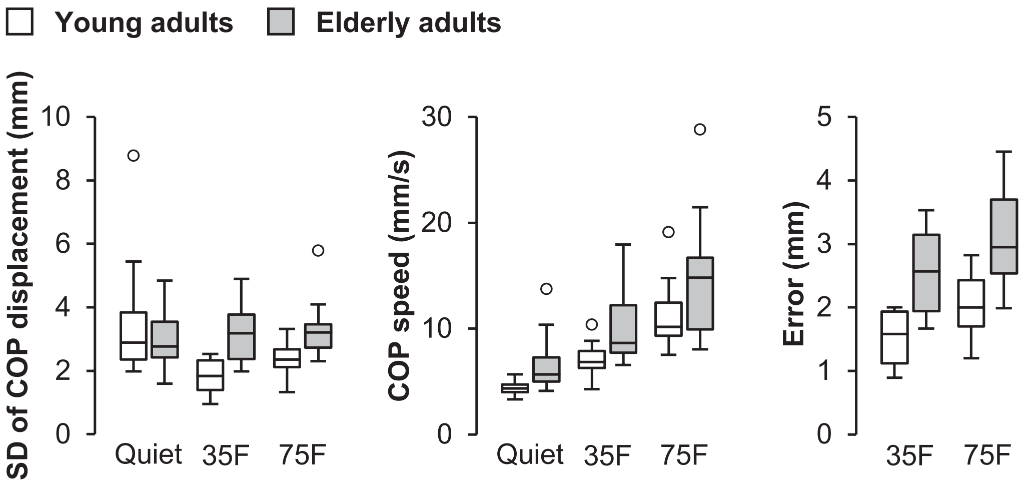

Fig. 3.3. Effects of task and age on center of pressure (COP) parameters. Data for standard deviation (SD) of COP displacement, mean COP speed, and error from target line were separately presented from left to right. Tasks were quiet standing (Quiet), 35% forward lean (35F), and 75% forward lean (75F). Small circles indicate outliners.

3.5.4. Figure 3.4

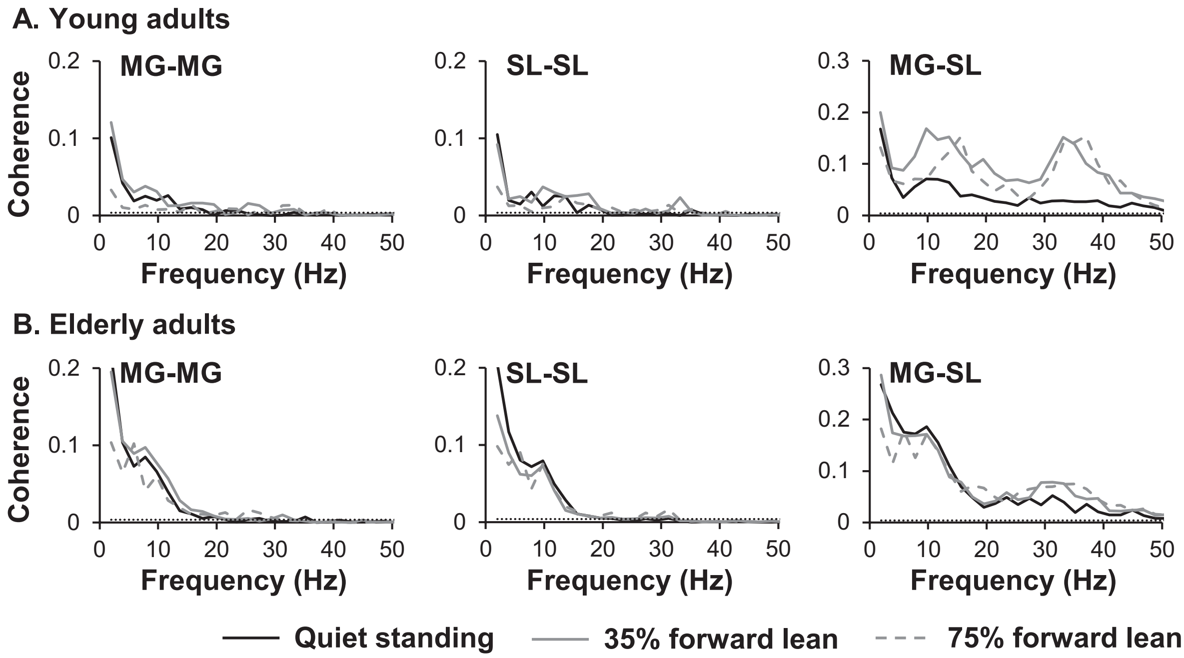

Fig. 3.4. Pooled coherence spectra for young (A) and elderly (B) adults. Data are presented separately for each task and muscle pair. Horizontal dashed line indicates the 95% confidence limit. MG: medial gastrocnemius; and SL: soleus.

3.5.5. Figure 3.5

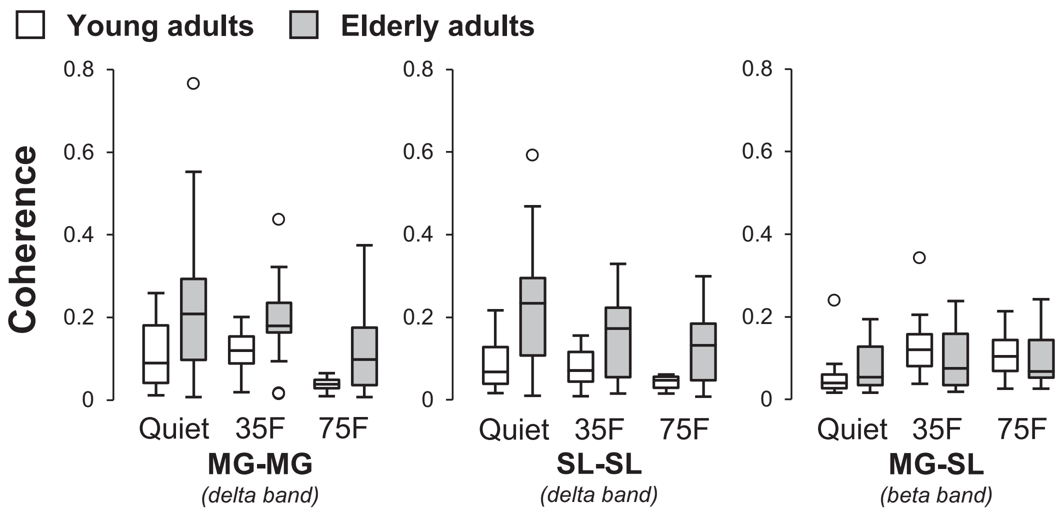

Fig. 3.5. Effects of task and age on z–transformed coherence. Tasks were quiet standing (Quiet), 35% forward lean (35F), and 75% forward lean (75F). Data are presented separately for each muscle pair. Small circles indicate outliners. MG: medial gastrocnemius; and SL: soleus.

4. General Discussion

The effective exercise intervention for preventing falls among the elderly population is a program composing strength training and exercise that challenges balance control [17-21]. However, this conclusion was made based on studies with younger, physically fit elderly adults [20,21], and it is presumably unsafe and, in some case, extremely difficult for older, frail elderly adults to perform especially the balance-challenging exercise. The aim of this doctoral thesis was to provide the foundational data that will eventually contribute to the development of an alternative, safe and effective fall-prevention intervention by investigating the underlying neurophysiological mechanism of age-related decline in voluntary control of unstable postures. We examined how aging affects delta- (<5 Hz) and beta-band (15-35 Hz) coherences between the plantar flexor muscles during single-leg stance and forward postural lean. We found that aging can increase these coherences during single-leg stance, and also impair the ability to modulate these coherences during forward postural lean.

The delta-band coherence between the plantar flexor muscles has been demonstrated to be larger in the elderly than young adults during quiet standing [41]. This doctoral thesis revealed that it was greater in the elderly than young adults during the quiet stance, single-leg stance, and forward postural lean tasks. The greater delta-band coherence was further related to the greater variability of postural displacements during quiet and single-leg stances particularly in the elderly adults. It appears that an increase in the low-frequency common input to the plantar flexor muscles increased the variability of force [44-47] and resultantly postural displacements particularly in the elderly adults. We also found the age-induced reduction in the ability to decrease the delta-band coherence between the bilateral homologous plantar flexor muscles during the forward postural lean. Therefore, it is likely that aging not only increase the delta-band coherence but also impairs the ability to decrease it, consequently affecting the postural performance.

During the steady isometric contractions, the beta-band synchronous oscillations are evident between the sensorimotor cortex and the contracting muscle (i.e., corticomuscular coherence) [36,99], and also between motoneuron pools within the same muscle or of synergistic muscles [39,111,112]. Although the role of these beta-band oscillations is not understood fully, they are reported to reflect the corticospinal activity and contribute to the steady and precise force output [49,50,60,113]. This doctoral thesis revealed that the beta-band coherence was evident between the unilateral plantar flexor muscles during the unstable balance-challenging postural tasks. Interestingly, aging was found to increase the beta-band coherence during the single-leg stance. Furthermore, the beta-band coherence positively correlated with the variability of postural displacements particularly in the elderly adults. These results may be ascribed to (unsuccessful [74]) compensation for the age-related general decline in sensorimotor function [114-116] to cope with the difficult postural task, or an impairment in the cortical inhibitory system interfering with accurate force production [75]. On the other hand, the elderly adults exhibited the reduced ability to increase the beta-band coherence during the forward postural lean. It appears that the effect of aging on the beta-band coherence depends on type of the postural task, unilateral or bilateral, and that the relation between the beta-band coherence and postural performance is not straightforward. It is possible that aging impairs the ability to effectively and functionally modulate the beta-band oscillations. Additional works are, however, needed to further explore the possibility and the underlying mechanisms.

Although more detailed investigations on the neural oscillations during the postural tasks are required, here we propose the potential interventions that might reduce the fall risks. Combining the results from previous studies on force control and this doctoral thesis, there are two possible ways to prevent or reduce the age-related decline in the ability to voluntarily control posture under unstable conditions: 1) decrease the delta-band oscillations in the muscle activity; 2) increase the ability to functionally modulate the beta-band oscillations. In a recent study, effect of visual guidelines on the variability of force output was investigated during a seated isometric contraction task in healthy young adults [117]. These visual guidelines were placed ±1 SDs of force output around the target line, and the low-frequency force and EMG burst oscillations were revealed to be smaller with than without the visual guidelines [117]. Furthermore, it has been reported that the cortical beta-band oscillations can be modulated by self-regulation of brain oscillation. For example, providing real-time feedback of the cortical beta-band oscillations could facilitate the ability to modulate the beta-band oscillations [118,119]. Although further studies are needed, such as to examine whether these improvements will persist over the long term and whether the physiological changes in the neural oscillations are accompanied by improvement in postural control, the techniques and tools mentioned here can be applied to the development of a new fall-prevention intervention based on the neural oscillations.

If this new intervention is developed, it may be integrated in a fall-prevention program in community and clinical settings. For example, frail elderly adults who are physically too weak to perform balance-challenging exercise may start with a simple brain and muscle activity modulation exercise. Like a video game, one may sit on a chair with simple portable EMG and electroencephalogram devices and try to control the brain and muscle activities shown on a monitor. Virtual reality technology may further provide an option for performing this type of exercise in simulated three-dimensional environments. As supervision of therapists or caregivers would not be needed, this intervention may also be used as home exercise and for home-based care. Moreover, tailor-made interventions may be designed based on the neural oscillations of each individual. Recent and future advances in technology will help to facilitate the implementation of this new intervention.

In conclusion, we demonstrated that aging influences the synchronous oscillations between the plantar flexor muscles during single-leg stance and forward postural lean. This doctoral thesis enhances our understanding of the effect of aging on voluntary neuromuscular control of posture under unstable conditions. By providing a link between the voluntary controls of steady-state force and posture, this work has the potential to advance the prevention approaches for falls.

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