博士論文

The effects of cognitive load on human postural control

（ヒト立位姿勢制御における認知負荷の影響）

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CONTENTS OF THE THESIS

ABBREVIATIONS

The main abbreviations used in this thesis are as follows.

A/D: analog to digital

ANOVA: analysis of variance

AP: anterior-posterior

CNS: central nervous system

COM: center of mass

COP: center of pressure

DC: direct current

DOF: degree of freedom

EMG: electromyogram

GRFs: ground reaction forces

MAXD: maximum distance

MD: mean distance

MG: medial head of the gastrocnemius

ML: mediolateral

PSD: auto-power spectral density function

SIP: single inverted pendulum

Sol: soleus

SEE: series elastic elements

TA: tibialis anterior

UCM: uncontrolled manifold

PREFACE

Maintaining upright standing is performed seemingly automatically as a precondition for performing the activities of daily living. Therefore, some people may assume upright standing to be a simple and effortless task for the central nervous system (CNS). On the contrary, upright standing is maintained by a complex process that requires the interaction of dynamic sensorimotor processes, including cognitive processing (Horak, 2006; Balasubramaniam and Wing, 2002).

It has been shown that the performance of concurrent secondary cognitive tasks (such as a visual search task, reaction task or mental arithmetic task) decreases as the attentional demands of postural control increase (Teasdale et al., 1993; Lajoie et al., 1996; see review by Woollacott and Shumway-Cook, 2002). The effects of performing an attention-demanding cognitive task on the control of posture have also been examined (see review by Fraizer and Mitra, 2008). It is suggested that the performance of tasks that require cognitive processing impedes stable standing postural control (i.e. the increase in postural sway) in elderly people and balance-impaired people (see review by Woollacott and Shumway-Cook, 2002). On the other hand, recently, it has been reported that performing a cognitive task while standing has been found to decrease postural sway in young healthy adults (Winter et al., 1990; Maylor et al., 2001; Dault et al., 2001; McNevin et al., 2002; Huxhold et al., 2006; Ueta et al., 2014).

Although many researchers have proposed a mechanism of decreasing postural sway during standing while performing a mental cognitive task, these have not been fully understood. The general purpose of this thesis is to examine effects of cognitive intervention on human postural control. To this end, three studies were conducted.

CHAPTER 1

BRIEF HISTORY OF THE STUDY

1-1. UPRIGHT STANDING BALANCE

Standing balance control is a complex process controlled by the CNS, due to the fact that the system is intrinsically unstable; a large body mass is located at a high position above the relatively small base of support (in particular the heavy head is at the top). Since spontaneous sway in quiet standing is considered to be a consequence or side effect of the motion-control process, many researchers have analyzed the spontaneous body sway to reveal the nature of the human postural control system.

1-1-1. Biomechanics of human quiet standing

Many studies (Gage et al., 2004; Masani et al., 2003; Peterka, 2002; Winter et al., 1998) on human postural control in quiet standing have assumed the single-joint model "a single inverted pendulum (SIP)" assuming that the ankle joint controls the movement of the center of mass (COM) (Fig.1-1). This is because, with a small perturbation or no perturbation, an upright stance can be maintained during quiet standing by the motion of the ankle, and movement around the hip joint is negligible (Horak and Nashner, 1986). The dynamic equation of the SIP model is as follows:

$$I\ddot{\theta} = mgh\ \sin\theta\ - T_a \qquad (1\text{-}1)$$

where I is the moment of inertia of the body, θ is the COM angle, m is the body mass, g is the acceleration of gravity, h is the distance of the COM from the ankle joints, T_a is the ankle torque. Equation (1-1) can be simplified as Eq. (1-2) by using the small-angle approximation; i.e., $sin\theta \approx \theta$.

$$I\ddot{\theta} = mgh\theta - T_a \qquad (1\text{-}2)$$

Since the COM is usually maintained a few centimeters in front of the ankle joints during quiet standing (Smith, 1957; Gatev et al., 1999), gravity continuously acts on the pendulum to produce a toppling torque. The ankle extensor muscles coupled to the pendulum by the series elastic elements (SEE) pull the pendulum backward to prevent falling. Equation (1-2) indicates that the difference between the forward torque due to gravity and the backward torque due to the ankle joint is proportional to the angular acceleration of the pendulum.

1-1-2. Posturography - assessment of postural sway

In studies using the SIP model, the center of pressure (COP) as measured by a force plate is often used as an evaluation index (Fig.1-1). The following equation expresses the equilibrium between torques around the single (ankle) joint on the sagittal plane.

$$T_a + f_v u \approx 0 \qquad (1\text{-}3)$$

where f_v is the vertical component of the reaction force and u is the COP position in the anteror-posterior (AP) direction from the ankle joint. The vertical component of the reaction force can be approximated to the body weight at quiet standing ($f_v \approx mg$). Since the body weight (mg) is constant, it is clear that the COP position is proportional to the magnitude of the ankle joint torque. Furthermore, Eq. (1-2) can be converted into the following equation.

$$y - u \approx \frac{I}{mg}\ddot{\theta} \qquad (1\text{-}4)$$

where y is the position of the COM in the AP direction as seen from the ankle joint, and

$\ddot{\theta}$ is the angular acceleration of the COM. This expression indicates that the error between the COM and COP positions determines the angular acceleration $\ddot{\theta}$ around the ankle joint. In this way, the COP is a variable reflecting the COM position and the angular acceleration $\ddot{\theta}$ generated by the error of torque.

In many of the studies that measure the COP to assess upright standing posture, indexes of COP velocity (e.g. mean velocity and total path length) and indexes of COP amplitude (e.g. standard deviation, SD; root mean square, RMS; maximum distance, MAXD; and mean distance, MD) have been used. The former are regarded as indicators of the amount of control activity necessary to maintain upright standing (Maki et al., 1990). In the SIP model, for adjusting posture while standing without disturbance, it is important to demonstrate torque by appropriate muscle activity around the ankle joint. This muscle activity includes a motor error that prevents generating a perfect torque response (Jeka et al. 2004). As mentioned above, this error of torque generated the angular acceleration and, as a result, it causes the body to sway (see Eq. (1-2)). The COM acceleration is highly correlated with the mean COP velocity (Masani et al., 2014). Therefore, the mean COP velocity reflects the fluctuation of ankle joint torque in the standing posture, indicating that it may be related to the postural control strategy (Kirshenbaum et al., 2001). Indexes of COP amplitude are regarded as indicators to evaluate the magnitude of postural sway in a trial. Since this becomes larger with aging, deficit of sensory information, and some disorders, a smaller COP amplitude is interpreted as better stability in most studies. The previous simulation study of quiet standing showed that COP measures related to amplitude (e.g. MD, RMS and MAXD) were positively correlated with noises level (Maurer and Peterka, 2005). Although the noises have not been clearly defined, these have seemed to contain neural or physical noise that is assumed in motor control and endogenous perturbation that is produced by the biogenic activity (e.g. heartbeat, breathing and error in the motor command) (Masani et al., 2007). Besides, depending on the cause of the postural instability, velocity-related sway measures are more sensitive to Parkinson's disease (Burleigh et al. 1995; Maurer et al. 2003; Rocchi et al. 2002), peripheral neuropathy (Dickstein et al. 2001; Horak et al. 2002), and postural instability in elderly adults (Maki et al. 1990; Prieto et al. 1996) than amplitude-related ones. Based on these factors, careful interpretation of the COP indicator and use of multiple measures is necessary.

1-1-3. The history of human bipedal posture study

Mechano-reflex hypothesis

Ankle torque can consist of passive and active components (Loram and Lakie, 2002a). The passive components of ankle joint torque consist of intrinsic mechanical properties (i.e., stiffness and/or viscosity) of the joint structure such as the ankle muscle, tendon, ligament, fascia, and soft tissue. In contrast, the active torque component is generated by contractile muscle elements that are regulated by neural commands from the CNS to counter torque due to gravity.

It had been generally assumed that the control of human upright stance solely relies solely on low-level mechano-reflex mechanisms that operate through the spinal cord and brainstem (Gurfinkel et al. 1995; Fitzpatrick et al. 1994, 1996; Horak 2006). This is based on the idea that complicated control is built by combining multiple reflexes. That is, when the body swayed forward, stretched calf muscles passively generated the restoring force in a spring-like manner. If passive stiffness could not provide adequate force to counteract gravity on its own, additional force to restore the pendulum provided by reflexes was needed. Such regulation mechanisms operate as a negative feedback position controller, typically used in the engineering design of servomechanisms.

Stiffness control hypothesis

In the mechano-reflex hypothesis, the control via reflex only through the spinal cord induced by sensory input can cause instability of the system because physiological time delay is inevitable. Winter et al. (1998) argued against the hypothesis and proposed a relatively simple control scheme to maintain upright posture the "stiffness control hypothesis". Their argument was based on the fact that (1) proprioceptive feedback signals during quiet standing are below physiological thresholds and (2) if mechano-reflex mechanisms were present, inevitable neuromuscular (150–250 ms) delay would create feedback instability. The stiffness control hypothesis insists that passive mechanical stiffness of the active calf muscle itself can provide sufficient torque to restore the pendulum. In their theory, CNS intervention is limited to setting the appropriate tension of the calf muscle. Then Winter et al. (2001) estimated the ankle stiffness from the regression line of the COM sway angle and ankle torque in order to show that the stiffness of the ankle exceeds the load stiffness. In their result, the ankle stiffness was 8.8% above the load stiffness (mgh) due to gravity, and they concluded that the quiet standing was maintained with passive stiffness.

Active, non-springlike control hypothesis

Immediately after that proposal, the researchers encountered problems with the stiffness control hypothesis. Their method of estimating ankle joint stiffness used by Winter et al. (2001) cannot distinguish passive stiffness or stiffness due to muscle activity via the CNS. Furthermore, the physiological value of intrinsic ankle stiffness (the word "intrinsic" here means without neural modulation) were overestimated (Morasso and Sanguineti, 2002). Subsequently, some researchers (Loram and Lakie, 2002b; Casadio et al., 2005; Vette et al., 2010) measured the passive "intrinsic" ankle stiffness of the ankle joint by giving the body a small ankle rotational disturbance like a sway while quiet standing. As a result of multiple experiments, it has been shown that the passive intrinsic ankle stiffness was 91 ± 23% (Loram and Lakie, 2002b), 64 ± 8% (Casadio et al., 2005), and 83 ± 7% (Vette et al., 2010) of the stiffness required to maintain upright standing (load stiffness). Furthermore, this measured stiffness showed a constant value irrespective of the ankle joint torque, although the triceps surae muscle was activated (Loram and Lakie, 2002b). When measuring the stiffness of a series-arranged spring with different constants, the combined stiffness is less than the stiffness of the most compliant spring. In the situation of human quiet standing, constant contraction of the muscles cannot provide sufficient rigidity, since conformable series elastic elements (SEE) limit the combined stiffness of the series arrangement:

$$\frac{1}{K_{total}} = \frac{1}{K_{muscle}} + \frac{1}{K_{SEE}} \qquad (1\text{-}5)$$

In order to raise the combined stiffness (K_{total}) above the load stiffness (mgh), the contractile muscle element is predicted to behave like a "negative spring". That is, when the body sways forward (and the muscle-tendon complex of the calf muscles, as a whole, lengthens), the contractile muscle element shortens (and vice versa when the body sways backward). Loram et al. (2005) proved the validity of this prediction via automated analysis of ultrasound images to track tiny muscular movements (10 μm) occurring during unperturbed, quiet standing. They found that the muscle length changes in the opposite direction to COM fluctuation (i.e. paradoxical Muscle Movement) and has a negative correlation to the cross-correlation function (Loram et al., 2005). They also showed that the calf muscles are actively adjusted 2.6 times per second (1.3 Hz, which corresponds to a mean time period of ~800 ms) and 2.8 times per unidirectional sway of the COM; these on average, and that these small movements provide impulsive, ballistic regulation of COM movement. This duration is longer than the reflex-initiated response time (about 65 ms; Evans et al. 1983) and is comparable to a reaction time requiring prediction and planning using internal models. Therefore, they suggested that standing is controlled not solely by local reflex but also by the CNS including the cerebellum (Loram et al., 2005). However, the CNS control mechanism has not yet been fully elucidated.

Multi-joint coordination process

In the SIP model, it is assumed that only the movement of the ankle controls the movement of the COM during standing. However, the results of recent research emphasize the importance of considering the control of multiple lower limb joints even in quiet standing (Aramaki et al. 2001; Creath et al. 2005; Hsu et al., 2007; Pinter et al., 2008). For example, Aramaki et al. (2001) revealed that the angular displacement, velocity, and acceleration of the hip were significantly greater than those of the ankle. Moreover, they reported a consistent reciprocal relationship between the angular accelerations of the ankle and hip joints (anti-phase pattern) during quiet standing. In addition, Creath et al. (2005) investigated the angular relationship between the leg and the trunk using frequency domain analysis. They demonstrated that in-phase and anti-phase relationships coexist between the legs and the trunk. That is, the angular motion of both segments was in-phase less than 1 Hz and anti-phase for frequencies above 1 Hz.

In recent years, an uncontrolled manifold (UCM) analysis of balance recovery from perturbation has shown that most joint variability is not related to whole-body postural variability but represents inter-joint coordination that leaves the COM or head positions constant (Scholz et al. 2007; Hsu et al., 2007). Krishnamoorthy et al. (2005) studied joint coordination in subjects who stood as quietly as possible on a narrow beam both with and without vision. They suggested that this multi-joint coordination effect was enhanced when vision was eliminated and that most of the increased joint variability due to the restriction of visual information is consistent with a control strategy that uses redundant joint combinations to control the COM position. Taken together, it is reasonable to conclude that postural control of stance is a multi-joint coordination process. They asserted that the results of this analysis confirm that the CNS makes special efforts to limit body sway by coordinating variance at all joints such that most joint motion is decoupled from the motion of the body in space (Hsu et al., 2007), whereas the ankle joint contributes greatly.

1-1-4. Spinal reflex modulation during upright standing

H-reflex and M-wave

The H-reflex is a monosynaptic reflex induced by an electrical stimulation and its test is used as a neurophysiological measurement to examine the influence of group Ia monosynaptic projection on spinal α-motoneuron activation. The H-reflex is generally studied in the Sol muscle that responds to the electrical unipolar stimulation delivered to peripheral nerves that consist of mixed sensory and motor-nerves fibers. When starting to increase stimulation intensity, the Ia afferent fiber that is thicker than others reacts first. This fiber passes from the muscle spindle through the dorsal root of the spinal cord and has a monosynaptic connection with α-motoneurons that dominate muscle spindle muscles. Thus, the ignition of Ia afferent fibers by electrical stimulation induces monosynaptic reflex (H-reflex) of the dominant muscle to which these are connected. The H-reflex of the triceps surae appears 30 to 50 ms after stimulation. As the stimulation intensity increases, the amplitude of the H-reflex also increases. When stimulus intensity exceeds a certain level, the first EMG response, observed with surface or indwelling electrodes placed over the muscles of interest, arises from the direct activation of motor axons of the α-motoneurons and its termed the M-wave. Since the position of the stimulus is close to the muscle, the action potential directly caused by the stimulus reaches the muscle earlier than the H-reflex that needs to pass through the spinal. The M-wave appears approximately 8 ms after stimulation. The amplitudes of these responses increase until a certain stimulus intensity is reached. As the stimulation intensity further increases, the amplitude of the H-reflex decreases. This is because the action potential of the stimulus-induced alpha motor fiber is conducted in both directions, namely, the direction of the muscle and the direction of the spinal cord. That is, the retrograde action potential of the spinal cord counteracts the monosynaptic action potential from the Ia afferent fiber.

H-reflex modulation during postural tasks

It has been suggested that the Sol H-reflex modulation is related to postural stability (Earles et al., 2000; Koceja et al., 1995). For example, Hayashi et al. (1992) observed that the soleus (Sol) H-reflex was suppressed while maintaining upright standing, as compared with sitting, even at the same level of background muscle activity. Koceja et al. (1995) showed that young subjects reduced the amplitude of the H-reflex from the prone position (Hmax/Mmax = 73.6%) where is with lower muscle activity to the standing (Hmax/Mmax = 59.9%) condition with higher muscle activity. It is considered that this suppression of the H-reflex of soleus muscle in standing is related to loading on body segments. It has been reported that as the ankle- or knee-joint load was reduced, the H-reflex was significantly enhanced (Nakazawa et al., 2004). On the other hand, it has been reported that mechanical ipsilateral loading of the sole of the foot reduced the amplitude of the Sol H-reflex (Knikou and Conway, 2001). A possible explanation of the suppression of the H-reflex of Sol muscle during standing is the presynaptic inhibition. Some researchers suggest the involvement of descending commands as the neural origin of presynaptic inhibition (Chen and Zhou, 2011). It is suggested that suppression of the H-reflex amplitude during standing prevents excessive autogenic excitation of the Sol motoneuron and helps the muscle to receive reliably central descending commands. It is also suggested that the CNS controls voluntary movements and may effectively inhibits sensory inputs of low importance using "presynaptic inhibition" as well as activating muscles (Seki et al., 2003).

It also has been shown that peripheral information during standing is involved in modulation of the H-reflex. In some literature (Taube et al., 2008; Earles et al., 2000; Huang et al., 2009), it has been reported that the amplitude of the H-reflex was modulated by the magnitude of the COP displacement. Tokuno et al. (2008) examined the position- and direction-dependent modulation of the Sol and MG H-reflex in relation to body sway in the anterior–posterior direction. The size of the triceps surae H-reflex was greater when swaying in the forward direction or in the forward position than when swaying in the backward direction or in the forward position. Load-related sensory information from the lower limb muscles and joints and the sole of the foot likely contributes to the modulation of the Sol H-reflex during standing (see review by Nakazawa et al., 2012).

1-2. COGNITIVE PROCESSING FOR POSTURAL CONTROL

Postural control is a complex process that requires the interaction of dynamic sensorimotor processes (Horak, 2006; Balasubramaniam and Wing, 2002). Horak (2006) described the resources required for postural control stability and orientation, which are divided into six categories (Fig.1-2). A disorder in any of these categories can lead to postural instability (Horak, 2006). In his description, cognitive processing is one of the components for postural stability and includes attention and learning.

1-2-1. Cognitive function

Cognition has a broad meaning. It is commonly understood to be the prerequisite to any purposeful and goal-directed action and was originally thought to be regulated exclusively by the CNS. Cognitive function also includes many elements, such as attention, orientation, memory, gnosis, executive functions, praxis, language, social reasoning and visuospatial skills.

Cognitive function including attention will be defined here as the information-processing capacity of the individual. An assumption regarding this information-processing capacity is that it is limited to the individual and that performing any task requires a given portion of capacity. Physiological investigations show that performing mentally effortful tasks places a high demand on cognitive processes (Logan and Gordon, 2001; Watter and Logan, 2006). Thus, when one or more tasks are performed at a time (dual tasking), and they require more than the total available processing capacity, execution of one or some of them may be slower and inaccurate. The more capacity is given to the task, the faster and more accurate the performance will be. While performing cognitive tasks, the subject makes an effort and bears a cognitive load. In this study, we use cognitive intervention to manipulate the allocation of processing capacity (i.e. manipulating cognitive load) by instruction to direct one's attention to own posture and impose two concurrent tasks.

1-2-2. Dual-task paradigm

Two broad frameworks (Logan and Gordon, 2001; Watter and Logan, 2006; Pashler, 1994; Fraizer and Mitra, 2008) have been proposed to impose a theoretical structure on posture-cognition outcomes. Capacity theory, on the one hand, views dual-task interference as arising from the parallel sharing of a limited set of general-purpose resources or multiple resources each of which is specialized. On this account, when combined task demands exceed the (centralized or particular) resource supply, degraded performance is observed on one task or both. In other words, we can do two tasks in parallel if enough processing capacity is available. Bottleneck theory, in contrast, is based on the idea that parallel processing may be impossible for certain mental operations. This model emphasizes the serial (i.e., sequential) nature of the dual-task process in terms of the single-channel filtering or scheduling of information at the stimulus encoding, identification, or decision-response stage (Pashler, 1994; Fraizer and Mitra, 2008). When two tasks need a channel at the same time, one or both tasks will be delayed or otherwise impaired, resulting in reduced performance of the non-priority task. In either theoretical model, the amount of processing resources required for task execution is not uniquely determined by the type or difficulty of the task. A task repeated enough times requires fewer processing resources (e.g., walking or driving), and it is generally called "automatic." Most psychologists define an automatic task as a mental operation that has two properties: proceeding without voluntary control (being obligatory) and not requiring capacity or processing resources (Pashler, 1994). Automaticity will emerge one becomes accustomed to a task or attains proficiency in a task.

Although a theoretical model has been proposed for dual-task interference, there are still many unknown points about the neural basis of the interference. In recent years, Watanabe and Funahashi (2014) showed that interference in cognitive–cognitive dual tasks and capacity limitation occurs at the nerve level. They recorded single-neuron activities in the lateral prefrontal cortex while monkeys performed dual tasks and showed that prefrontal neuron activities showed a decreased ability to represent task-relevant information to a degree proportional to the increased demand of the concurrent counterpart task. This indicates that the interference between two cognitive tasks that need to be handled in the same area competes for neural activity resources. Does this competition in cognitive–cognitive dual tasking also occur in cognitive–motor dual tasking? It is unknown whether similar neural resource integration occurs in the interference of human cognitive–motor dual tasks, which are thought to be mainly responsible for processing. It is expected that more research will be conducted.

1-2-3. Effects of postural task on cognitive task

Some researchers have shown that postural control requires significant allocation of attentional resources (Kerr et al., 1985; Lajoie et al., 1993; see review Woollacott and Shumway-Cook, 2002). Kerr et al. (1985) published the first article to demonstrate the attentional demands of stance postural control in young adults by using dual-task paradigm. The dual-task paradigm, in which subjects attempt to execute an additional non-postural task during a postural task, has been used to investigate the role of cognitive demands in postural control. This method requires the performance of a primary task, such as maintaining an upright stance, concurrent with a secondary task, such as an attention-demanding cognitive or motor focal task. Kerr and his colleagues (1985) adapted a spatial processing task and a verbal task as a cognitive task. In the spatial task, the subjects were asked to place numbers on the imagined 4 × 4 squares according to the instructions to be read and to memorize their positions. In non-spatial verbal task, they were asked to memorize adjectives to be read in turn. The performance of the spatial task decreased more while maintaining a standing posture than while sitting. They concluded that cognitive spatial processing relies on neural mechanisms that are also necessary for the regulation of standing posture, whereas the verbal control task did not interfere and thus does not require spatial processing. Subsequently, several researchers (see review Woollacott and Shumway-Cook, 2002) have revealed that the difficulty of the postural task itself also influences the attentional demands of posture. It was found that the performance of cognitive tasks decreases as the attentional demands of postural control increase (Teasdale et al., 1993; Lajoie et al., 1996). Therefore, it has been suggested that more cognitive involvement or more attention is required with increasing postural task difficulty (Lajoie et al., 1996; Vuillerme and Nougier, 2004).

1-2-4. Effects of cognitive task on postural task

Other authors have examined the effects of performing an attention-demanding cognitive task on postural control (see review Fraizer and Mitra, 2008). These studies on the effects of a cognitive task on a postural task have mainly chosen a task with minimal postural demand (e.g., unperturbed stance, large base of support). It was obvious that in elderly people postural sway increased by the execution of simultaneous cognitive task during standing (Shumway-Cook et al., 1997; Jamet et al., 2007; Huxhold et al., 2006). It is suggested that significant attentional demands associated with postural control in older adults. It has also been reported that the impact of dual tasks on postural control is significantly higher in geriatric patients with cognitive impairment and a history of falls resulting in injury than same-age subjects without these impairments (Shumway-Cook et al., 1997; Hauer et al., 2003).

In young adults, however, several unexplained discrepancies in the results weaken this literature. I showed the summary of results of postural-cognitive dual-task studies in Table 1-1. Pellecchia (2003) tried to quantify the difficulty of the cognitive task using a method of information reduction based on the idea of Posner (1966) and reported that the amplitude of postural sway increases with the difficulty level of the cognitive task. He suggested that the increase in postural sway observed in his study might reflect a greater challenge encountered in coordinating standing balance and cognitive tasks as the cognitive activity is rendered more difficult, though he did not provide a test of competing hypotheses. However, it has pointed out that some of tasks set as cognitive tasks in previous studies require a physical response to secondary tasks and may not correctly evaluate the interference of cognitive processing in standing control (see review by Fraizer and Mitra, 2008). For example, in the reaction time task, the action of pressing a button may generate mechanical demands that produce changes in sway that have nothing to do with cognitive load. Similarly, in tasks that require a verbal answer to evaluate the performance of a cognitive task during posture measurement, the utterance generates a larger sway.

On the other hand, it has been reported that performing a mental cognitive task while standing has been found to decrease postural sway in young healthy adults (Winter et al., 1990; Maylor et al., 2001; Dault et al., 2001; McNevin et al., 2002; O. Huxhold et al., 2006; Ueta et al., 2014). For example, performing visuospatial working memory task (Dault et al., 2001) and visual task (Bonnnet and Baudry, 2016A) while standing has led to reduced sway amplitude. In these tasks, individuals need to succeed in precise visual tasks upright and should not be more unstable and inefficient. It is suggested that synergy between visual and postural processes may be required to succeed in precise visual tasks (see review Bonnnet and Baudry, 2016B). The central nervous system should work adaptively in a way that enables success in these tasks.

The reduced sway amplitude is also confirmed in silently counting backward (mental arithmetic) tasks (Stins et al., 2011; Andersson et al., 2002) and short-term memory tasks (Riley et al., 2003; Ueta et al., 2012) that needs orally answered during measurement. This effect has been interpreted as resulting from improved postural control and has been suggested to be attributable to the utilization of more automatic control processes of the standing posture (Ueta et al., 2014) or increased arousal (Riley et al., 2003). However, it is important to note that these are conceptual explanations.

Another attempt has been made to explain the decrease in postural sway observed under the dual-task condition as being due to an individual's focus of attention (Wulf et al., 2004). These researchers pointed out that manipulation of the focus of attention, that is, the focus on body movement (internal focus), causes an increase in the mean velocity of postural sway, and focusing on the concurrent task (external focus) causes a decrease in postural sway amplitude. In their view, in a fully automated movement such as standing, the internal focus constrains the motor system and disrupts automatic motor control processes through the conscious control of movements. On the other hand, directing attention to the suprapostural task (external focus) implies that attention is drawn away from postural control. Therefore, they argue that the execution of a cognitive task causes excessive automation and improvement of postural stability. In perturbed standing, however, it has been shown that there is a negative influence on the response to perturbations during the cognitive task as compared with no additional task. The actual mechanisms of how cognitive processing reduces postural sway are poorly understood.

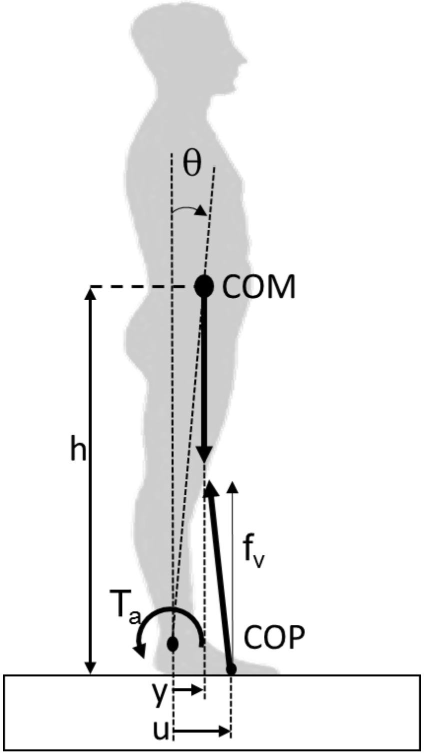

Fig.1-1. An inverted pendulum model of quiet standing in the sagittal plane. COM is the center of mass, COP is the center of pressure, h is the COM height from ankle joint, θ is the sway angle, T_a is the ankle torque, f_v is the vertical component of the reaction force, *y* is the horizontal distance of the COM in the AP direction from the ankle joint, and *u* is the horizontal distance of the COP in the AP direction from the ankle joint

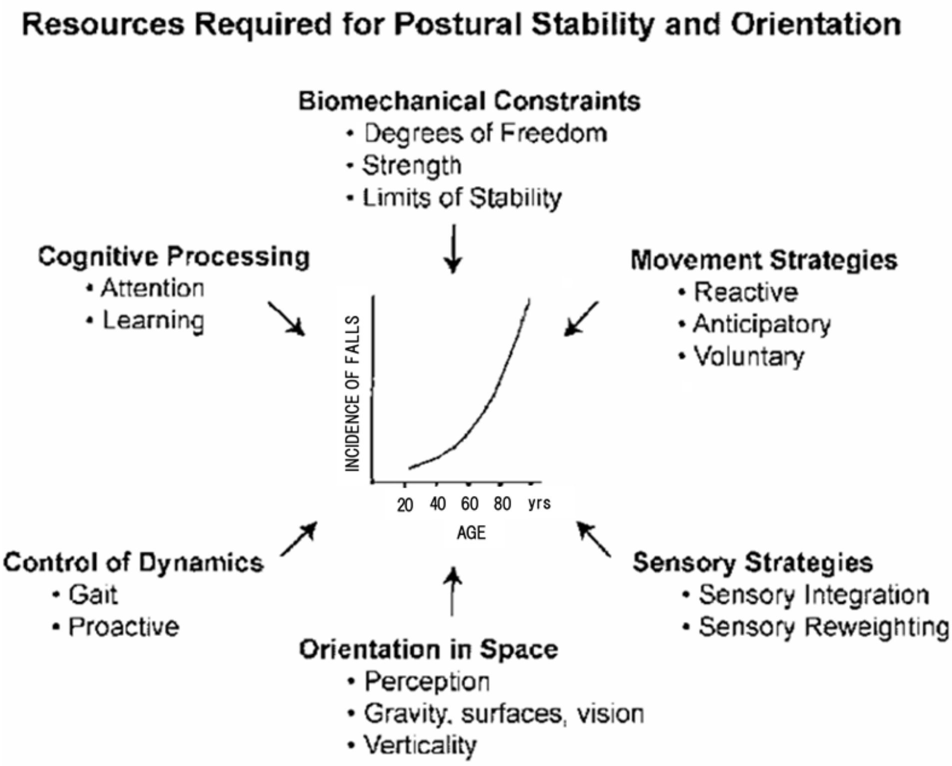

Fig.1-2. Six components of resources required for postural stability and orientation (Horak, 2006). *Age and Ageing*, 35-S2, ii7-ii11

Table 1-1. Summary of results of postural-cognitive dual-task studies for young adults

Reference	Postural Task	Cognitive Task	measurements	postural sway
visual search or spatial task				
Kerr et al. (1985)	Tandem Romberg stance	Brooks spatial memory task	COP path length, SD	no change
		Nonspatial verbal memory task	COP path length, SD	no change
Yardley et al. (2001)	standing with eyes closed	Spatial reaction time task	Velocity, RMS of COP	no change
			Velocity, RMS of COP	no change
		Non-spatial reaction time task	Velocity, RMS of COP	no change
			Velocity, RMS of COP	no change
Storegen et al. (2000)	quiet stance	Visual search task	SD of head position	decreased
Maylor et al. (2001)	quiet stance	Brook's spatial memory task	COP velocity	decreased
			SD of COP	decreased
		Brook's nonspatial memory task	COP velocity	increased
			SD of COP	increased
Prado et al. (2007)	quiet stance	Visual search task	RMS of AP sways	decreased
			COP velocity (AP)	decreased
			RMS of ML sways	no change
			COP velocity (ML)	no change
Richer et al. (2017)	quiet stance	digit search task	SD of COP	decreased
			COP velocity	increased
tasks that need to physically respond				
Teasdale et al. (1993)	standing on form surface	Auditory reaction time task	mean speed of COP	no change
Dault et al. (2001)	quiet stance	Stroop task	RMS of COP amplitude	no change
			Mean frequency	increased
			RMS of COP velocity	no change
Pellecchia (2003)	quiet stance	Count backwards by 3s (oral reply)	COP path length	increased
			Variability	increased
Kuczynski et al. (2011)	quiet stance	Stroop task	Variability	no hange(AP) decreased(ML)
			Mean speed	increased(AP) increased(ML)
			Frequency	no hange(AP) increased(ML)
			Sample entropy	no hange(AP) increased(ML)
Remaud et al. (2013)	quiet stance	Reaction time task (verbal response)	area of the 95 % confidence ellipse	no change
			COP mean velocity	decreased
Shumway-Cook. (1997)		Judgment of Line Orientation (JOLO)task	COP path length	no change
		Sentence Completion task	COP path length	increased

tasks without verbal answers				
Andersson et al. (2002)	quiet stance	Backward silent counting task	Mean AP torque variance	no change
			Mean ML torque variance	no change
	standing with vibratory calf stimulation	Backward silent counting task	Mean AP torque variance	decreased
			Mean ML torque variance	decreased
Riley et al. (2003)	quiet stance	Digit memory task (easy)	SD of COP (AP)	no change
		Digit memory task (midium difficulty)	SD of COP (AP)	decreased
		Digit memory task (difficult)	SD of COP (AP)	decreased
Stins et al. (2011)	quiet stance	Backward silent counting task	SD of COP (AP)	decreased
			MPF of COP (AP)	increased
			Sample entropy	increased
Ueta et al. (2012)	quiet stance	Digit memory task	RMS of COP velocity	decreased
			MPF of COP	increased
			the power density on the low frequency of COP	decreased

CHAPTER 2

PURPOSE OF THE THESIS

As mentioned in the PREFACE, although numerous researchers have investigated the effects of cognitive intervention on human postural control and its mechanisms, these have not been fully understood. Especially the mechanism of change in postural sway by cognitive intervention is not clarified, although conceptual theory is advocated in previous researches. The general purpose of this thesis is to examine the effects of cognitive intervention on human postural control. To this end, three studies were conducted.

As described in CHAPTER 1, while the COP represents the net neuromuscular response of the CNS to control displacements of the COM (Winter et al., 1996), it is difficult to estimate whole-body postural control using only COP parameters. Therefore, in order to examine the influence of cognitive interventions on the kinematic and kinetic variables of the standing posture, three experiments were conducted in Study 1, as will be reported in CHAPTER 3. Specifically, I conducted experiments focusing on the aspects of kinematics (Experiment 1), respiration movements (Experiment 2), and muscle activity (Experiment 3).

It is suggested that the amplitude of postural sway decreases during the execution of the concurrent cognitive task and that the decrease in sway is not due to the stabilization of posture (in Study 1). Modulation of postural control (stabilization and destabilization) is emphasized in dynamic postural tasks. If posture gets instability during dual tasking, there will be a change in postural adjustment after perturbation. Therefore, I aimed to clarify the influence of the cognitive task on posture response induced by the anterior-posterior direction (backward and forward) perturbation during standing (Study 2). In order to test its effect, subjects were required to maintain an upright standing posture against perturbation sliding horizontally as well as forward and backward.

In Study 2, there was no change in muscle activity after the disturbance, but it is necessary to verify the possibility that the sub-threshold neural adjustment occurs. Then, in Study 3, I aimed to investigate the effect of the cognitive load during upright standing on the efficacy of the Ia afferent pathway to discharge motor neurons. To this end, I evaluated the Sol H-reflexes elicited by stimulating the right tibial nerve in two cognitive interventions in comparison with simple quiet standing.

CHAPTER 3

EFFECTS OF COGNITIVE INTERVENTIONS ON MODIFICATIONS OF KINEMATICS AND KINETICS DURING UPRIGHT STANDING.

3-1. Introduction

Previously, many studies revealed that postural control and cognitive processing influence each other, depending on the complexity of both tasks (Fraizer and Mitra, 2008; Lacour et al., 2008). In most of these studies, researchers have attempted to reveal attentional demand for postural control through interactions between cognitive tasks and postural steadiness as assessed by COP-based measurements. Although these studies have been on for about three decades, the consensus has not been obtained for changes in postural sway due to cognitive intervention (see CHAPTER 1). For this reason, we first evaluate the COP of large number of subjects during cognitive intervention.

The attention to one’s own movement has been shown to increase the mean velocity and mean power frequency of COP (Ueta et al., 2015). Some researchers suggest that this attention leads the control of voluntary processes, and that enhanced motor outputs, such as increased joint stiffness, damping, and co-contraction of the ankle muscles, account for increased COP velocity (Ueta et al., 2015; Reynolds, 2010). On the other hand, the addition of a cognitive task during upright standing has been shown to decrease COP amplitude (Andersson et al., 2002, Riley et al., 2003; Stins et al., 2011 Ueta et al., 2015; Richer et al., 2017). Some theories exist as to how cognitive processing reduces postural sway (see CHAPTER 1). However, while the COP represents the net neuromuscular response of the central nervous system (CNS) to control displacements of the COM, it is difficult to estimate whole-body postural control by only COP parameters. To approach the mechanisms of postural control during cognitive intervention, the analyses of kinematics are beneficial. Especially, the analyses of the effect of joint configuration variance, including those other than the ankle, on the stability of the COM is needed to estimate whole-body postural control. Therefore, in Experiment-1 of Study 1, I aimed to identify how instructions to limit sway and a mental cognitive task modify kinematics of standing posture. The knowledge gained from this study can provide important insight into precise mechanisms of interactions between cognitive demand and postural control during standing.

Then, to approach the cause of reduced postural sway during cognitive tasking I focused on respiration in the present study. Respiratory movement is known to affect postural sway during standing under a condition wherein respiration is manipulated experimentally (Hodges et al., 2002; Caron et al., 2004; Kuznetsov et al., 2012), although the effect is insignificant in quiet breathing due to compensation by the body segments (Hodges et al., 2002). Respiration is also known to be affected by psychological stressors, including a cognitive task (Porges and Byrne, 1992). The execution of a cognitive task causes sympathetic nerve activation (e.g., Callister et al., 1992), which is well recognized to increase with the respiration rate. Taken together, it is possible that postural sway and cognitive processing are related through respiration. Therefore, in the present study, I hypothesized that cognitive processing reduces postural sway due to changes in the respiratory pattern. To confirm this hypothesis, in Experiment-2 of Study 1, I investigated the relationship between the respiratory movement and standing postural sway (COP displacement, joint fluctuation) which was influenced by a cognitive task.

Previous research indicates that the mean velocity of postural sway increase when subjects pay attention to own posture and intention to minimize sway (Ueta et al., 2014; Reynolds, 2010). It is suggested that this is due to an increase in postural control activity as a result of enhanced voluntary processes (see CHAPTER 1). There is, indeed, a possibility that excessive postural correction for small postural sway may be carried out due to attention to his or her own posture. Therefore, in the conscious standing that was observed to increase mean COP velocity, an increase in muscle activity of the lower leg should be observed. On the other hand, recent studies have confirmed a decrease in the COP amplitude observed in young adults' posture during dual tasking (see CHAPTER 1). In these studies, the decrease of COP amplitude has been interpreted as resulting from improved postural control. As mentioned above, there is a possibility that reduction of respiratory movement may be involved in reduction of postural sway amplitude. If so, it is considered that in also the unperturbed standing the reduction of posture adjustment exists, and it may be mistaken that the decrease in sway amplitude is the result of stabilization of control. Certainly, small adjustments may be sufficient for the small sway. In order to show existence of the reduction of posture adjustment, it is necessary to evaluate posture adjustment independent of sway size, that is, to compare in the same postural state. In Experiment-3 of Study 1, therefore, I aimed to investigate the influence of cognitive intervention on the lower leg muscle activity during standing.

3-2. Methods

Subjects

Fifty-one healthy adult males (age 22.7 ± 5.9 years; height 172.0 ± 5.4 cm; weight 68.6

± 15.8 kg) participated in the measurement of the ground reaction forces (GRFs) by the force platform (Experiment-1). In the residual part of Experiment-1 and Experiment-2 and 3, subjects were fifteen healthy young males (age 23.9 ± 3.0 years; height 173.8 ± 6.5 cm; weight 66.5 ± 8.7 kg) of them. They had no history of neurological disorders. The experimental procedures used in this experiment were in accordance with the declaration of Helsinki and were approved by the ethical standards of the committee on Human Experimentation at the Graduate School of Arts and Sciences of the University of Tokyo. All subjects gave their written informed consent to participate in this study after receiving a detailed explanation of the purpose, potential benefits, and risks involved.

Procedure

The barefoot subjects were required to keep an upright stance for 30 s on a force platform (Type 9281B, Kistler, Winterthur, Switzerland) with their eyes open and with feet parallel 15 cm apart from the centers of both heels. The subjects held their arms by their sides looking at a target placed at eye level 1.5 m in front of the subjects.

Three conditions were set as follows. In the quiet standing (QSt) condition, subjects were asked to relax and maintain an upright stance. In the conscious standing (CSt) condition, they were asked to concentrate on their postural sway and reduce it as much as possible. In the standing with mental arithmetic (MASt) condition, they were asked to mentally count backward from randomly selected 3-digit numbers in steps of a single digit number (i.e. 6, 7, 8, 9) as fast and as accurately as possible and report the final number verbally at the end of each trial. Before the experiment, subjects practiced this task several times while sitting; its performance for 30 s was recorded once. Three trials were conducted under each condition in a randomized order. Short rests of 1 to 3 min were provided between each trial to exclude the influence of fatigue.

Measurement

In Experiment-1, the ground reaction forces (GRFs) were obtained by the force platform from all subjects. The GRF data were collected at a sampling rate of 1 kHz. The COP displacement in the anterior-posterior (AP) and the medio-lateral (ML) directions were calculated from the measured GRFs and low-pass filtered with a cutoff frequency of 10 Hz using a second-order low-pass Butterworth filter.

In 15 of the subjects, kinematics data were obtained by an optical motion capture system (OptiTrack: V100R2, NaturalPoint, Corvallis, OR, USA) composed of six infrared cameras in a semicircular arrangement. Six reflective markers (5 mm diameter) were placed over surface landmarks to monitor the motion of the trunk, thigh, lower leg, and foot (Fig.3-1A). The ankle, knee, and hip angles were calculated from the marker coordinates (Fig. 3-1B). These data were sampled at 100 Hz and low-pass filtered with a cutoff frequency of 2 Hz using a second-order low-pass Butterworth filter.

In Experiment-3, The EMG signals were recorded by surface EMG sensors with an inter-electrode distance of 10 mm (DE-2.1, DELSYS, Boston, MA, USA); the sensors were connected to a differential amplifier with a filter bandwidth of 20–450 Hz (Bagnoli-8, DELSYS, Boston, MA, USA). EMGs were recorded from the right medial gastrocnemius (MG), soleus (Sol) and tibialis anterior (TA). The EMG signals were stored with a sampling frequency of 1 kHz. For collecting and synchronizing force, kinematic data and EMG signals, the software of LabVIEW2012 on a single computer with DAQ system (USB-6221, National Instrument, TX, USA) was used. The EMG data acquired from a subject was excluded because of obvious measurement fault (MG and TA were attached in reverse).

Data analysis

Experiment-1

The root mean square (RMS) of COP displacement and the mean velocity (MV) of COP trajectory were calculated in the AP and ML directions to compare the COP variables. The MV was calculated by dividing the total length of the COP path by the sampling time.

In order to confirm that the tendency of posture has not changed, the average joint angle in three joints was calculated and compared between the conditions. The SDs of the joint angles were calculated to assess the amplitude of the angular displacement of each joint. To estimate the change of postural control in terms of joint mobility, joint stiffness was calculated. First, the joint torque at each joint was calculated from kinematic data and standard anthropometric parameters (Winter, 2009). Then, the intrinsic joint stiffness was estimated from the linear regression of the joint torque and joint angle over the measuring period for each joint according to the methods of Winter et al (2001).

Uncontrolled manifold (UCM) analysis was performed to quantify the structure of the coordination between the major postural chain joints (ankle, knee, and hip). The analysis relies on a geometric link segment model of the body that relates the changes in these individual joints to the changes in the COM position in the AP direction. The outputs of this analysis are an index of the amount of joint variability that leaves the COM constant (V_{UCM}) and an index of the amount of joint variability that leads to changes in the COM position (V_{ORT}). The ratio of the V_{UCM} and V_{ORT} (V_{UCM}/V_{ORT}, UCM ratio) was also calculated. The UCM ratio reflects how variables that are involved in the motor tasks are coordinated to control variables that are regulated by neural control mechanisms (Park et al., 2016). In the present study, the UCM ratio was the index of the strength of multi-joint coordination that preserves the COM position consistent with its mean position across time. A lower ratio (≤ 1) suggests coordination in which the preserved COM position is weak or not present, while a higher ratio (> 1) reflects better multi-joint coordination (Latash et al., 2002; Scholz et al., 2003).

Experiment-2

As with Experiment-1, two COP variables, RMS and MV of COP were calculated in the AP and ML directions. The SD of the joint angles were calculated to assess the amplitude of the angular displacement of each joint.

Respiratory movements were estimated from the distance between the abdominal markers (L4 and the navel) because most of the subjects were using abdominal respiration. The displacement of respiratory movements was low-pass filtered with a cutoff frequency of 0.5 Hz, using a second-order low-pass Butterworth filter to clarify the trunk motion with breathing. The average respiratory rate was calculated from the displacement of the abdominal respiratory movement. Time points of maximal inspiration were estimated from the displacement, and the number of points was expressed in breaths per minute. The average breathing rate was calculated from the peak-to-peak times and expressed in breaths per minute. To estimate the amplitude of respiratory movement, the SD of the displacement of respiratory movement was calculated.

To normalize calculated variables among the subjects, the change in each amplitude variable from QSt to MASt were obtained as a rate of change by the following equation:

$$\text{Rate of change (\%)} = \frac{M_{MASt} - M_{QSt}}{M_{QSt}} \times 100$$

Accordingly, M_{MASt} and M_{QSt} are the mean values of variables (COP measures, joint variability, and amplitude of respiratory movement) during the MASt task (M_{MASt}) and

QSt (M_{QSt}) conditions. In this analysis, RMS of the COP in the ML direction was excluded. This is because the markers that identify respiration are L4 and the navel; thus, the spread of the thorax in the lateral direction is not taken into consideration.

Experiment-3

All EMG signals were first numerically rectified and processed by the second-order Butterworth low-pass filter with a cut-off frequency of 12 Hz (pEMG: processed EMG). The digitalized EMG signals were full-wave rectified after subtraction of the DC bias. To determine average EMG activity, RMS of EMG signals were calculated for each muscle. The data for a single trial was divided into 29 segments (half-overlapping), each with 2^{13} data points. A Fast-Fourier Transform algorithm was applied to generate a Fourier spectrum for each subset after being passed through a Hamming-window. Then, auto-power spectral density function (PSD) for the rectified EMGs were calculated. I defined EMG activities occurring at 0.1−5.1 Hz as the low-frequency component (LF). Then, I defined that occurring at 7.0−12.0 Hz as the high-frequency component (HF). This was because I wanted to have the same 5 Hz band as the LF around the peak power frequency (the peak power frequency in the QSt condition was observed at 8.68 ± 0.64 Hz for Sol, 9.18 ± 0.87 Hz for MG, and 9.25 ± 0.78 Hz for TA, see Fig. 3-9). Finally, I calculated the energy of each component by integrating the PSD and peak frequency at HF.

The muscle activation phase (on-period: activating; off-period; inactivating) from EMG data was determined from two low-pass filtered EMG signals according to the methods of Nomura et al. (2007) and Tanabe et al. (2017). All pEMG signals were low-pass filtered with a cutoff frequency of 0.01 Hz using a second-order low-pass Butterworth filter to obtain trend curves, which represent tonic muscle activity components. The pEMG signals were low-pass filtered with a cutoff frequency of 2 Hz to obtained the smoothed pEMG signals (sEMG: smoothed EMG), assuming that the trend curve subtracted from sEMG represents intermittent muscle activation due to postural control via the CNS. If the sEMG was above the trend curve for a certain period, I considered that the muscle was active in that period (EMG-on phase). Some EMG-on phase was excluded based on the following two criteria; (1) the sEMG does not exceed half of the maximum value of the sEMG (2) the period from turning on to next turning off is less than 500 milliseconds. An example of on/off periods determined by a single pEMG signal is shown in Figure 3-2. To match the number of data points to the COM time series, each filtered EMGs time series were first downsampled by a factor 10.

The COM position was estimated by three segments model (Winter, 2009) in A-P direction. The COM angle was calculated as an angle formed by the extension line from COM point to the ankle joint and the horizontal line. The COM angular velocity time series were obtained by differentiating the COM angle. The time series of COM angle and angular velocity was transformed into z-scores by mean SD of COM angle and angular velocity in MASt condition to standardize the amplitude of each subject. Mean SD of COM angle and angular velocity in MASt condition was used because their SD value was the smallest among the three conditions. From this, the COM angle and angular velocity parameters for the seven experimental conditions of experiment were determined (A: ~ 1.5 SD; B: -1.5 ~ -1.0 SD; C: -1.0 ~ -0.5 SD; D: -0.5 ~ 0.5 SD; E: 0.5 ~ 1.0 SD; F: 1.0 ~ 1.5 SD; G: 1.5 SD ~) for each. We calculated the probability (the rate of MG or TA-on) that muscle activity is on when it is in such specific COM states. The positive SD value means that COM was at the anterior position or was moving in the positive (i.e., forward) direction. On the other hand, the negative SD value means the state of COM was in the opposite way. The soleus shows tonic activity during quiet standing, so it has been impossible to detect on/off switching for such muscles.

Statistical analysis

Experiment-1

To compare the mental arithmetic performances of sitting and dual conditions, a paired t-test was performed. To compare each variable of COP displacement, amplitudes of joint motions, estimated joint stiffness, and components of UCM analysis among the three task conditions (QSt, CSt, MASt), a one-way analysis of variance (ANOVA) with repeated measures was performed to examine the effects of the task condition on the dependent measures. The sequentially rejective Bonferroni correction (the Holm method) was used to adjust the p values obtained in the post hoc analyses. The significance level was set at $p < 0.05$.

Experiment-2

To approach changes in respiration and postural sway during the cognitive task, I was interested only in comparing QSt and MASt. Paired t-tests were performed to examine the effects of two task conditions on each variable. Spearman correlations were used to explore relationships between the change rates of the postural sway amplitude (i.e., the RMS of the COP displacement and the SD of each joint) and the respiratory amplitude. The significance level was set at $p < 0.05$.

Experiment-3

To compare the variable of RMS of EMGs, integrated PSD at each frequency components (LF, HF) and peak frequency at HF among the three task conditions (QSt, CSt, MASt), a one-way analysis of variance (ANOVA) with repeated measures was performed to examine the effects of the task condition on the dependent measures. The sequentially rejective Bonferroni correction (the Holm method) was used to adjust the p values obtained in the post hoc analyses. A series of 3 (condition: QSt, CSt, MASt) × 7 (the state of COM angle and angular velocity) repeated-measures variance analyses were performed to examine changes in the rate of EMGs-on of each muscle (MG and TA). Simple effect analyses with Bonferroni corrections ($p = 0.05$/number of comparisons) were performed. All significance level was set at $p < 0.05$.

3-3. Results

Experiment-1

Performance of an arithmetic task

Performance of a mental arithmetic task while sitting or standing was not significantly different ($p = 0.13$; mean repetition times, sitting = 11.4 ± 4.7 times, standing = 15.6 ± 8. 8 times), indicating that the subjects' attention to the cognitive demand did not decrease when standing as compared to when sitting.

COP variables

The mean values of the COP variables in the AP and ML directions for three task conditions are shown in Fig.3-3. The amplitude of the COP displacement was smaller in the MASt condition than in the other two tasks in the AP and ML directions. ANOVA showed a main effect of task condition (AP, $F_{(2,28)} = 9.98$, $p < 0.01$; ML, $F_{(2,28)} = 9.40$, $p < 0.01$). Post hoc comparisons showed significantly smaller RMSs in the MASt condition than in the other two conditions in both directions (Fig.3-3A).

The MV was larger for the CSt condition than for the other conditions. ANOVA showed a main effect of task condition in both directions (AP, $F_{(2,28)} = 11.95$, $p < 0.01$; ML, $F_{(2,28)} = 17.75$, $p < 0.01$). Post hoc comparisons showed a significantly greater MV in the CSt condition than in the other two conditions (Fig.3-3B).

Joint movements

Figure 3-4 shows the averages of angle in each joint. Analysis of the averages of angle did not show a significant main effect of task condition (ankle, $F_{(2,28)} = 0.06$, $p = 0.95$; knee, $F_{(2,28)} = 0.01$, $p = 0.99$; hip, $F_{(2,28)} = 0.05$, $p = 0.95$). Figure 3-5 shows the SDs of angular displacement in each joint for three task conditions. The amplitudes of joint movements in the ankle and hip were decreased in the MASt condition. There was a significant main effect of task condition on the SDs of angular displacements in the ankle and hip joints (ankle, $F_{(2,28)} = 7.44$, $p < 0.01$; hip, $F_{(2,28)} = 9.02$, $p < 0.01$), whereas a difference was not found in the knee joint ($F_{(2,28)} = 2.59$, $p = 0.08$). Post hoc comparisons showed that the SD of the ankle was significantly smaller in the MASt condition as compared with the QSt and CSt conditions. The SD of the hip was also significantly smaller in the MASt condition as compared with the other two conditions.

Joint stiffness

Figure 3-6 shows the estimated joint stiffness in the ankle, knee, and hip joints. The tasks used in this study did not affect the stiffness in each joint. There was no main effect of task condition in all three joints (ankle, $F_{(2,28)} = 0.37$, $p = 0.67$; knee, $F_{(2,28)} = 0.21$, $p = 0.79$; hip, $F_{(2, 28)} = 0.76$, $p = 0.45$).

UCM analysis

UCM analysis was performed to quantify the structure of the coordination between the major postural chain joints. Figure 3-7 shows two averaged indices (V_{UCM} and V_{ORT}) and its ratio obtained from UCM analysis. The results demonstrated that variations at all joints were affected by the task conditions, whereas the strength of multi-joint coordination was not different among all three conditions.

The cognitive task led to a smaller overall variance than in the other conditions. There was a main effect of task condition in V_{UCM} ($F_{(2,28)} = 5.46$, $p < 0.01$) and V_{ORT} ($F_{(2,28)} = 5.67$, $p < 0.01$). Post hoc comparison showed a significantly smaller V_{UCM} and V_{ORT} in the MASt condition as compared to the other two conditions (Fig.3-7A). Analysis of the UCM ratio (V_{UCM}/V_{ORT}) did not show a significant main effect of task condition ($F_{(2,28)} = 1.36$, $p = 0.25$).

Experiment-2

COP measures

The COP measures of 15 subjects who acquired kinematics data were summarized in order to confirm whether the trend of change in COP was consistent with all subjects (51 subjects). The mean values of COP variables are shown in Table 3-1. A mental arithmetic task in the present study reduced the sway variability of the COP. Paired t-tests revealed that the MASt condition produced a significantly smaller RMS of COP displacement as compared to the QSt condition in both the AP and ML directions (AP: $p < 0.01$; ML: $p < 0.01$). The MASt condition also produced a significantly smaller MV of COP displacement as compared to the QSt condition in the ML direction ($p < 0.01$), whereas no significant difference of the MV was observed in the AP direction ($p = 0.44$).

Respiratory measures

The mean values of the respiratory variables are shown in Table 3-2. The respiration of the subjects was greatly affected by a mental arithmetic task in the present study. A paired t-test revealed that the respiratory rate in the MASt condition was faster than that in the QSt condition ($p < 0.01$). The amplitude of the respiration movement in the MASt condition was smaller than that in the QSt condition ($p < 0.01$).

Correlation between change rates

The regression line, the correlation coefficient value, and its significance are presented in Fig. 3-8. A significant positive correlation was found between the rate of change in respiration and the COP amplitude (COP-AP, $r = 0.78$, $p < 0.01$; COP-ML, $r = 0.67$, $p < 0.05$) and between the rate of change in the respiratory amplitude and the SD of hip movement ($r = 0.56$, $p < 0.05$). No significant correlation was found between the change rate of the respiratory amplitude and the SD of ankle or knee joint movements (Ankle, $r = 0.41$, $p = 0.14$; Knee, $r = 0.11$, $p = 0.72$).

Experiment-3

RMS of EMG

The mean values of RMS of EMG are shown in Figure 3-9. There was no main effect of task condition in Sol and TA muscles (Sol, $F_{(2,26)} = 1.23$, $p > 0.05$; TA, $F_{(2, 26)} = 2.49$, $p > 0.05$). There was a significant main effect of task condition on the RMSs in the MG ($F_{(2,26)} = 3.28$, $p = 0.04$). However, post hoc comparisons showed no significance in any comparisons.

Frequency domain analysis

Figure 3-10 shows the results of frequency domain analysis. There was no significant difference in the LF in the Sol ($F_{(2,26)} = 0.49$, $p = 0.59$). There was a significant main effect of task condition on the LF in the MG ($F_{(2,26)} = 6.56$, $p < 0.01$). Post hoc comparisons showed that the LF in the MG was significantly smaller in the MASt condition as compared with the QSt and CSt conditions. There was a significant main effect of task condition on the LF in the TA ($F_{(2,26)} = 3.30$, $p = 0.04$). However, post hoc comparisons showed no significance in any comparisons ($p > 0.05$). There was no significant difference in the HF among the conditions for either muscle (Sol: $F_{(2,26)} = 1.14$, $p = 0.30$; MG: $F_{(2,26)} = 0.76$, $p = 0.45$; TA: $F_{(2,26)} = 2.65$, $p = 0.07$).

Activation probability (EMG on-rate) relate to COM state

Figure 3-11 shows the MG activation on-rate relate to COM state. The COM angle significantly affected the MG activation on-rate ($F_{(6,78)} = 3.94$, $p < 0.01$), whereby larger the MG activation on-rate occurred in a positive COM angle (G: 1.5 SD~) compared with a negative COM angle (A: ~ -1.5 SD; B: -1.5 ~ -1.0 SD; $p < 0.001$). There was a significant main effect of task condition on the MG activation on-rate relate to COM angle ($F_{(2,26)} = 3.23$, $p = 0.03$). Post hoc comparisons showed that the MG activation on-rate relate to COM angle was significantly smaller in the MASt condition as compared with the CSt condition ($p < 0.001$). There was a significant main effect of task condition on the MG activation on-rate relate to COM angular velocity ($F_{(2,26)} = 4.76$, $p < 0.01$). Post hoc comparisons showed that the MG activation on-rate relate to COM angle was significantly smaller in the MASt condition as compared with the other two conditions ($p < 0.001$).

Figure 3-12 shows the TA activation on-rate relate to COM state. There was a significant main effect of task condition on the TA activation relate to COM angle ($F_{(2,26)} = 8.17$, $p < 0.01$). Post hoc comparisons showed that the TA activation on-rate relate to COM angle was significantly larger in the CSt condition as compared with the other two conditions ($p < 0.001$). There was a significant main effect of task condition on the TA activation relate to COM angular velocity ($F_{(2,26)} = 10.80$, $p < 0.01$). Post hoc comparisons showed that the TA activation on-rate relate to COM angular velocity was significantly larger in the CSt condition as compared with the other two conditions ($p < 0.001$).

3-4. Discussion

Effects of directing attention to postural sway

The results showed that directing attention to postural sway (i.e., the CSt condition) increased the COP velocity as compared to normal quiet standing (Experiment-1**)**. Some researchers suggest that this condition is more under the control of voluntary processes and that enhanced motor outputs, such as increased joint stiffness, damping, and co-contraction of the ankle muscles, account for increased COP velocity (Ueta et al., 2014; Reynolds, 2010). This implication is based on suggestion that the COP velocity is related to providing information on postural corrections required to maintain postural stability (Bonnet and Baudry, 2016; Maki et al., 1990). I did not observe improvement of postural steadiness (i.e., COP amplitude) when the subjects were instructed to minimize their postural sway (Experiment-1**)**. This result is not surprising, since many previous studies have shown that directing attention to postural sway has little or no effect on postural sway itself as compared to directing participants' attention to the effects of postural sway (Park et al., 2015).

In the present study, the strategy of stiffening joints, which is a candidate for the cause of the increase in mean COP velocity, was estimated. The results demonstrated that there was no significant difference in ankle stiffness between normal standing and this condition, meaning that the strategy of stiffening joints seems unlikely to explain increased COP velocity (Experiment-1**)**. There was no sustained increase in tonic activity in the evaluation of muscle activity. In addition, the UCM ratios, suggesting the strength of multi-joint coordination that preserves the mean COM position in the present study, were also not significantly different (Experiment-1). Our results demonstrated that directing the subjects' attention to their postural sway did not modify kinematic parameters reflecting these postural control strategies. A study of postural threat has suggested that a more conscious control of posture accompanies changes in balance perceptions and increases the frequency of postural adjustment showed as mean power frequency of sway (Huffman et al., 2009). The instruction that directs participants' attention to postural sway would also lead to conscious control of postural sway.

There were no significant differences in the mean values of RMS of EMGs between the simply quiet standing condition and consciously standing condition (Experiment-3). As mentioned above, there was no change in joints stiffness (Experiment-1), so it seems that the strategy of stiffen joints by muscle co-contraction of the plantar flexion and the dorsiflexion is not cause the change in COP velocity. On the other hand, the TA activation on-rate relate to COM angle was significantly larger in the CSt condition as compared with the other two conditions. There is a possibility that corticospinal excitability is related to this change of activation. Previously, to evaluate the excitability of the corticospinal tract, using the motor-evoked potentials (MEPs) by motor cortex transcranial magnetic stimulation (TMS), it is showed that MEPs of Sol and TA increase with increasing task difficulty considered to require more attention to own posture (Tokuno et al., 2018). Studies on the stretch reflex of the muscles around the ankle of during standing showed that the medium- (MLR) and long-latency stretch reflex (LLR) of TA increased compared to the sitting position (Obata et al., 2012). This long-latency reflex in TA muscle is at least partly transcortical (Petersen et al., 1998; Doornik et al., 2004). Additionally, it was shown that the TA responses were strongly influenced by changes in the preparatory state (Fujio et al., 2016). The corticomuscular connection seems to be stronger in the TA than other calf muscles. It is suggested that adjustment of posture by tibialis anterior muscle is a mediated through the transcortical pathway. On that account, conscious postural control in CSt condition may lead the accentuation of TA activity.

Effects of executing a cognitive task

The cognitive task that draws attention away from postural control (i.e., the MASt condition) decreased the COP amplitude as compared to normal quiet standing (Experiment-1). It is in line with previous studies (Andersson et al., 2002, Riley et al., 2003; Stins et al., 2011 Ueta et al., 2014; Richer et al., 2017). This effect has been interpreted as resulting from improved postural control and has been suggested to be attributable to the utilization of more automatic control processes of the standing posture (Ueta et al., 2014) or increased arousal (Riley et al., 2003). The present result demonstrated that drawing attention away from postural control reduced movements of the ankle and hip joints (Experiment-1). To our knowledge, this is the first investigation to report kinematic changes during cognitive interventions. A possible interpretation of the decrease in ankle and hip joint movements is increased joint stiffness accompanied. Dault et al. (2001) suggest that the central nervous system chooses a co-contraction control strategy during working memory tasks, since the reciprocal control of postural muscles requires more attention. However, present results revealed no change in joint stiffness across conditions (Experiment-1). These results are supported by previous studies demonstrating no effect of cognitive task in muscle activity around the ankle joint (Richer et al., 2017). The results of the strength of joint coordination for the COM were also negative for this consideration.

As a mechanism of decreasing of postural sway, I focused on respiratory movement and hypothesized that the execution of cognitive processing reduces postural sway due to changes in the respiratory pattern (Experiment-2). The results showed that the respiratory rate increased and the respiratory amplitude decreased during the mental arithmetic task. In addition, there was a significant positive correlation between the change in respiration and the COP amplitude. To our knowledge, this is the first study showing that reduced standing postural sway during cognitive tasks is related to changes in respiratory amplitude. Increase in the respiratory rate during a mental arithmetic task is consistent with results obtained in previous studies during sitting (Mulder and Mulder, 1981; Allen and Crowell, 1989). The enhancement of sympathetic nerve activity is the most probable explanation, since increased heart rates were reported along with increased respiratory rates in a previous study (Allen and Crowell, 1989). While many researches have been conducted on the influence of the task at the sitting position, no study had reported a change in respiratory during a mental arithmetic task while standing. Some researchers have suggested the interdependence of respiratory movement and postural sway during the experimental manipulation of breathing (Hodges et al., 2002; Caron et al., 2004). These studies suggested at respiratory movement is likely to disturb standing posture. Therefore, in the result of Experiment-2, decreased postural disturbance through reduced respiratory amplitude was possible to reduce the amplitude of COP displacement. The reduced respiratory amplitude also possibly reduced the amplitude of hip joint displacement, since the hip joint is closer to the rib cage.

In Experiment-3, there were no significant differences in the mean values of RMS of EMGs compared with the simply quiet standing condition, whereas it has been clarified that the body sway amplitude decreases in the cognitive task condition (Experiment-1). It was also revealed that the LF in the MG was smaller during standing with the mental arithmetic task as compared with during simple quiet standing. Loram et al. (2005b) showed that the interaction between the COM angle and the ankle extensors occurs in the lower frequency band (0-3 Hz). They suggested that the frequency of ~3-4 Hz is almost comparable to the time scale of the active neural modulation of the ankle plantar flexor activities during quiet standing. Therefore, the result of frequency domain analysis suggested that the active neural modulation of the ankle plantar flexor was decreased. The present result also revealed that the rate of MG-on with respect to the same magnitude of body sway is small during a dual-task condition simultaneously performing a cognitive task with quiet standing compared to quiet standing task condition. Even at the same state point in COM angular velocity, the rate of MG-on was lower during performing the cognitive task than quiet standing (Experiment-3). In comparison in COM angle, this tendency was shown, whereas statistically significant difference was shown only between CSt and MASt conditions (Experiment-3).

From these results, it is suggested that reduction of the COP amplitude does not indicate posture stabilization and the postural adjustments were rather reduced during dual tasking condition. It supports the suggestion that postural sway reduction is decreased disturbance due to thoracic movement (Experiment-2). It has been revealed that the performance or the response of muscle activity for postural recovery from surface perturbation was deteriorated when a cognitive task requiring working memory was added during standing (Rankin et al., 2000; Teasdale and Simoneau, 2001; Norrie et al., 2002). The results of this study can provide a unified explanation for the interference with postural control performance of the dual-task. On the other hand, TA activation on-rate relate to COM angle did not change MASt condition. TA activation on-rate in QSt as the baseline condition was very low and was substantially zero. As a result, depression of TA activity may be not observed.

In summary, in Study 1, it was shown that the characteristics of standing posture during cognitive intervention differ according to the task. Focusing own posture during standing increases the mean COP velocity, whereas the joint stiffness and the strength of multi-joint coordination were constant. I also found the higher TA activity rate in conscious standing condition. On the other hand, I also found that the performing simultaneous cognitive tasks decreased COP amplitude and the variability of the lower leg joints. Furthermore, it was suggested that the decrease of the COP amplitude strongly correlated with the decrease of respiratory amplitude. In addition, it was shown that the activity rate of ankle flexor decreased under this condition.

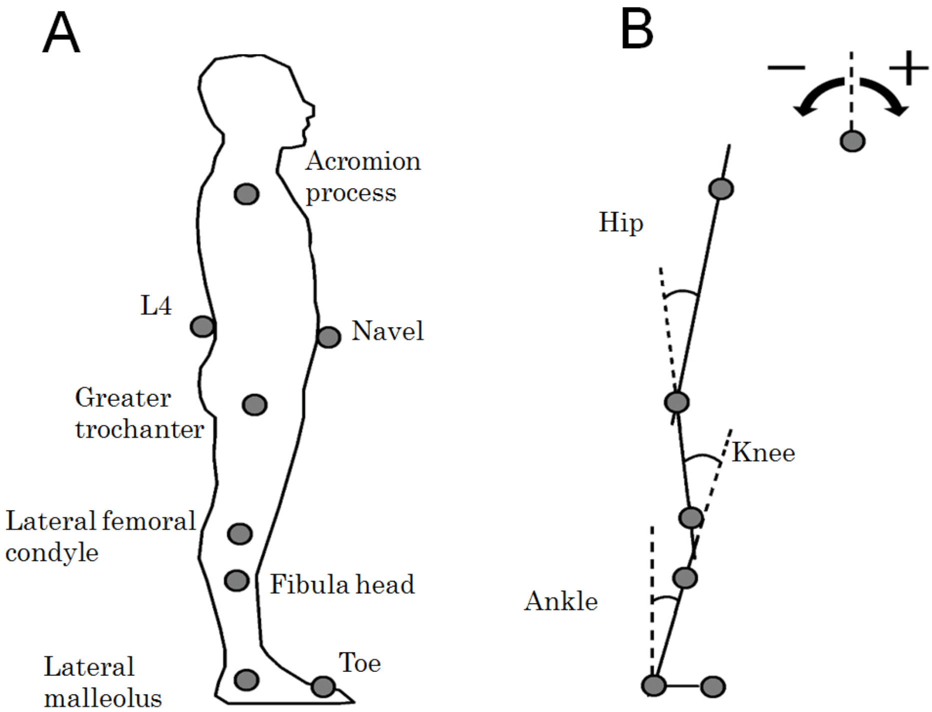

Fig. 3-1 Experimental Setup. Marker Placements (A) and the method for measuring angular motion (B)

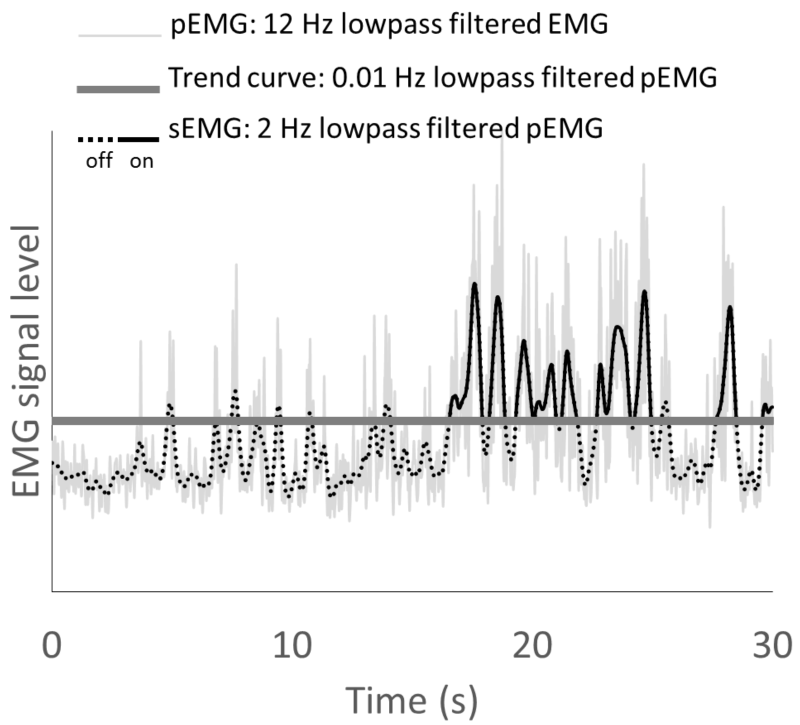

Fig. 3-2. An example of on/off detection for 30 s of EMG data. A thin gray plot represents 12 Hz low-pass-filtered EMG (pEMG). A thick gray gradual curve represents the trend curve (0.01-Hz low-pass-filtered pEMG) and dotted-bold black line represents sEMG (2 Hz lowpass filtered pEMG), respectively.

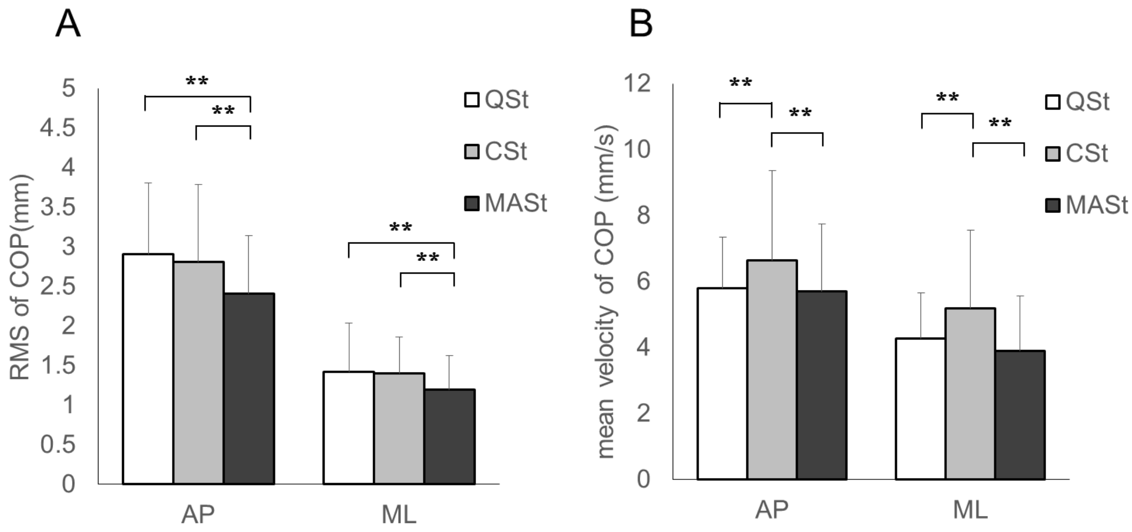

Fig.3-3. Mean and standard deviation of the root mean square (RMS) and the mean velocity (MV) of the center of pressure (COP) displacement in each condition
AP: anterior-posterior direction; ML: medio-lateral direction. (*p < 0.05, **p < 0.01)

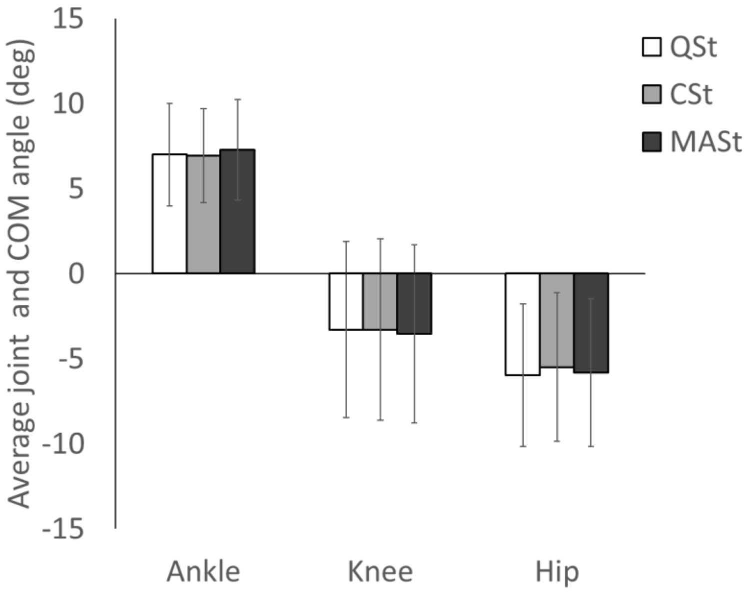

Fig.3-4. The average of the joint angles in each condition

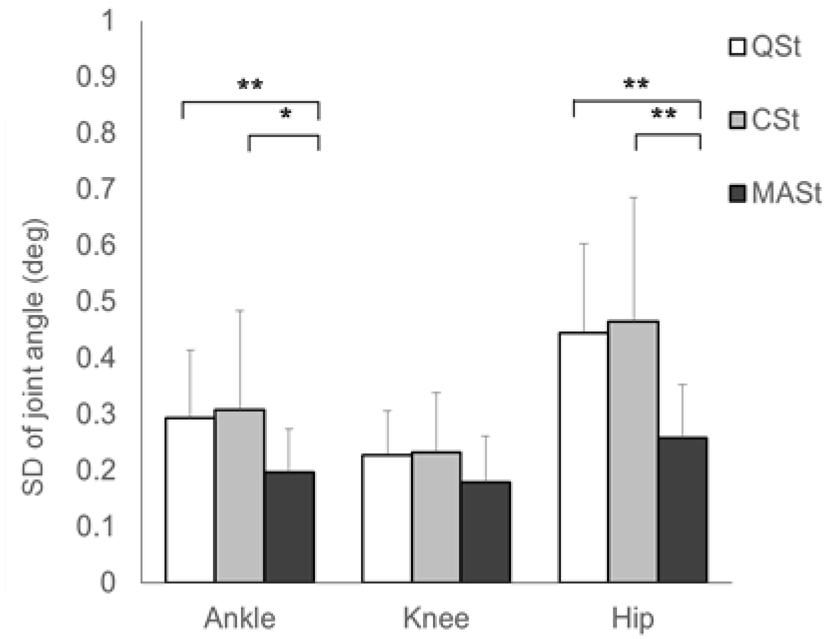

Fig.3-5. Mean and standard deviation of the SD of the center of the joint angle in each condition (*p < 0.05, **p < 0.01)

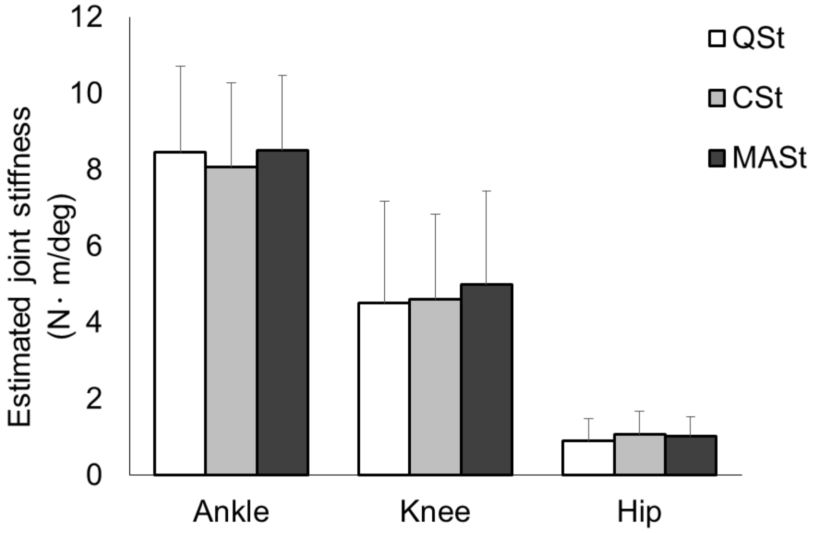

Fig.3-6. Mean and standard deviation of the estimated joint stiffness in the ankle, knee, and hip joints (*p < 0.05, **p < 0.01)

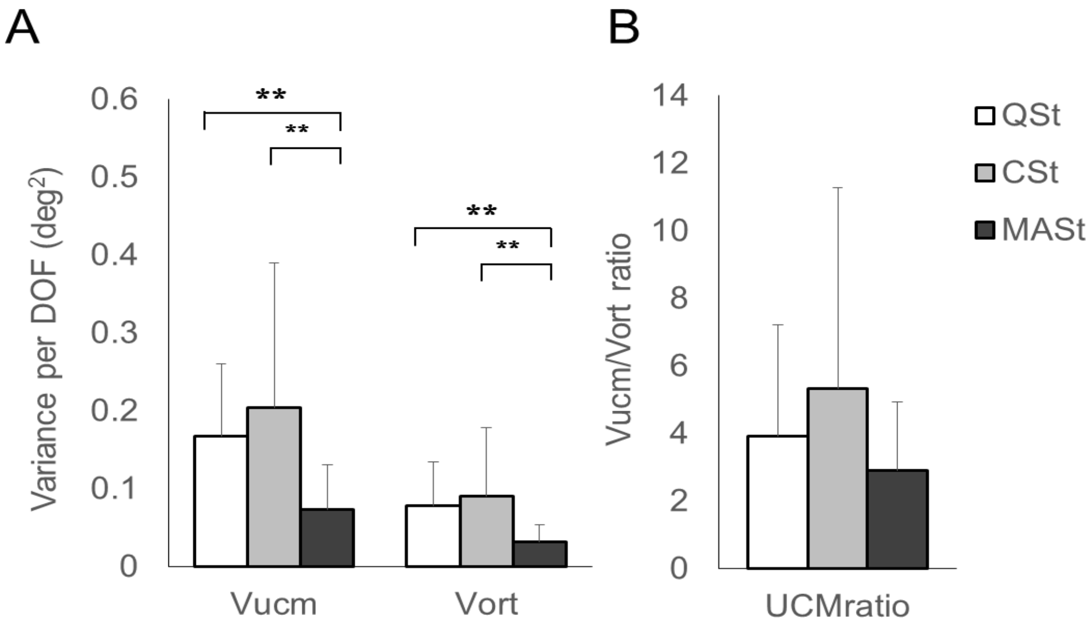

Fig.3-7. Mean and standard deviation variance parallel to the uncontrolled manifold (V_{UCM}) and variance orthogonal to the UCM (V_{ORT}) and its ratio (V_{UCM}/V_{ORT}) (*p < 0.05, **p < 0.01) DOF = degrees of freedom.

Table 3-1. COP measurements during quiet standing (QSt) and standing with mental arithmetic (MASt) conditions

	QSt		MASt		
COP measures	Mean	SE	Mean	SE	p
RMS_AP (mm)	3.47	0.66	2.49	0.43	$p < 0.01$
RMS_ML(mm)	1.33	0.23	1.05	0.19	$p < 0.01$
MV_AP (mm/s)	6.10	0.86	5.87	1.05	0.44 (n.s.)
MV_ML (mm/s)	4.11	0.66	3.39	0.57	$p < 0.01$

Table 3-2. Respiratory measurements during quiet standing (QSt) and standing with mental arithmetic (MASt) conditions

Breathing measurements	QSt		MASt		p
	Mean	SE	Mean	SE	
Breathing rate (times/min)	16.04	1.77	19.76	1.93	< 0.01
Abdominal movement SD (cm)	2.39	0.75	1.58	0.4	< 0.05

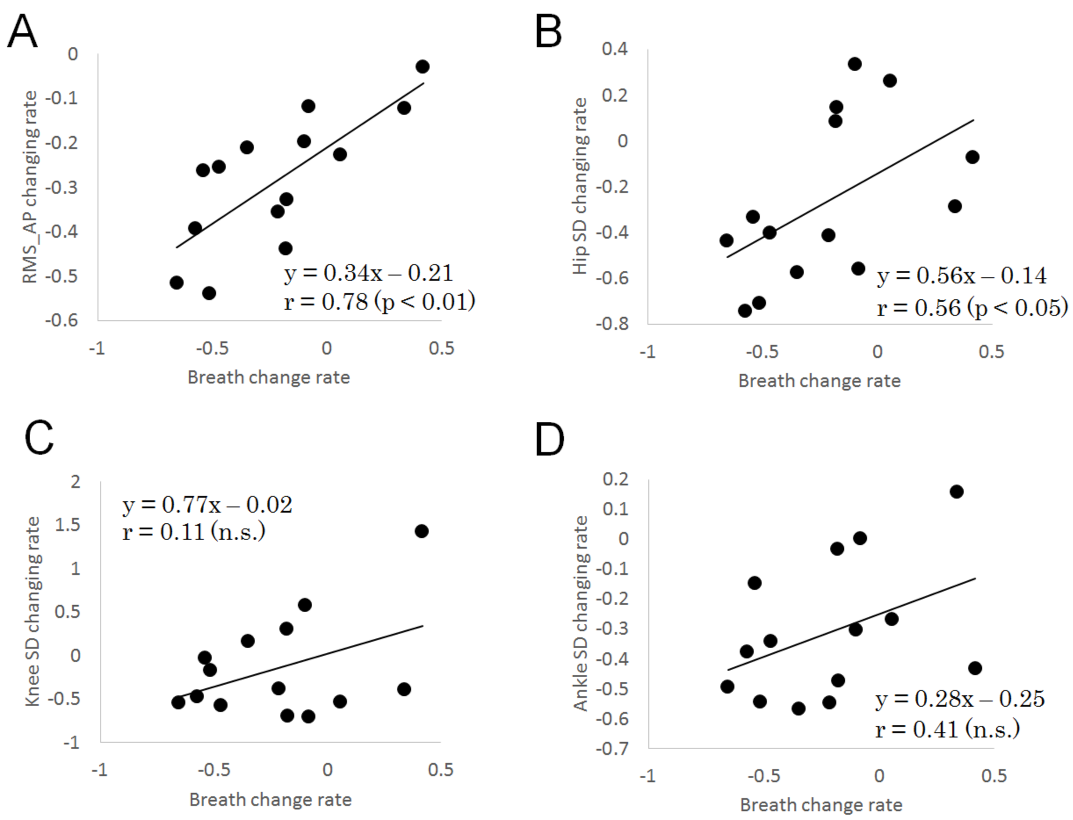

Fig. 3-8. Correlation between the change rate of the respiratory amplitude and each amplitude variable

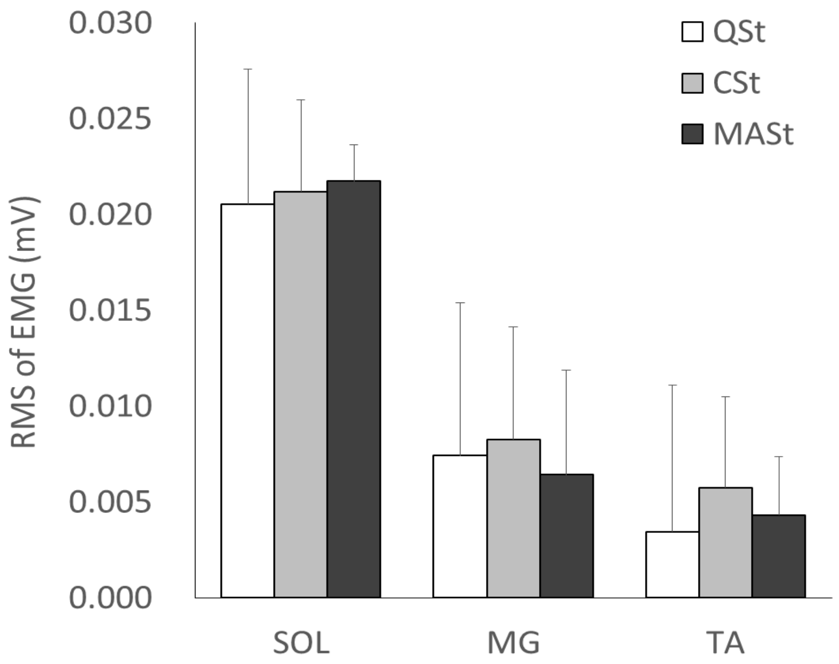

Fig. 3-9. Mean and standard deviation of RMS of EMGs (*p < 0.05, **p < 0.01)

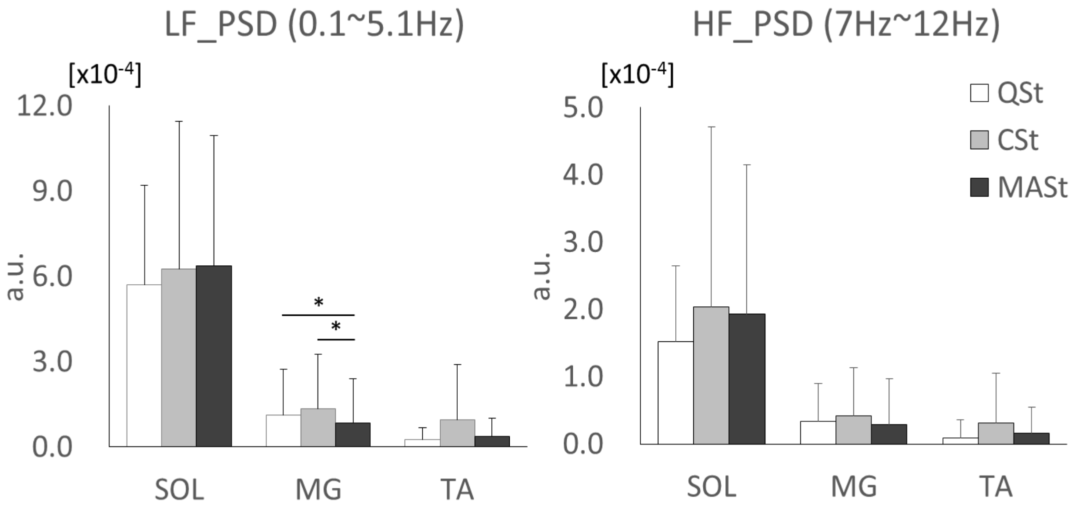

Fig. 3-10. The relative energy of the LF (left) and HF (right). (*$p < 0.05$, **$p < 0.01$)

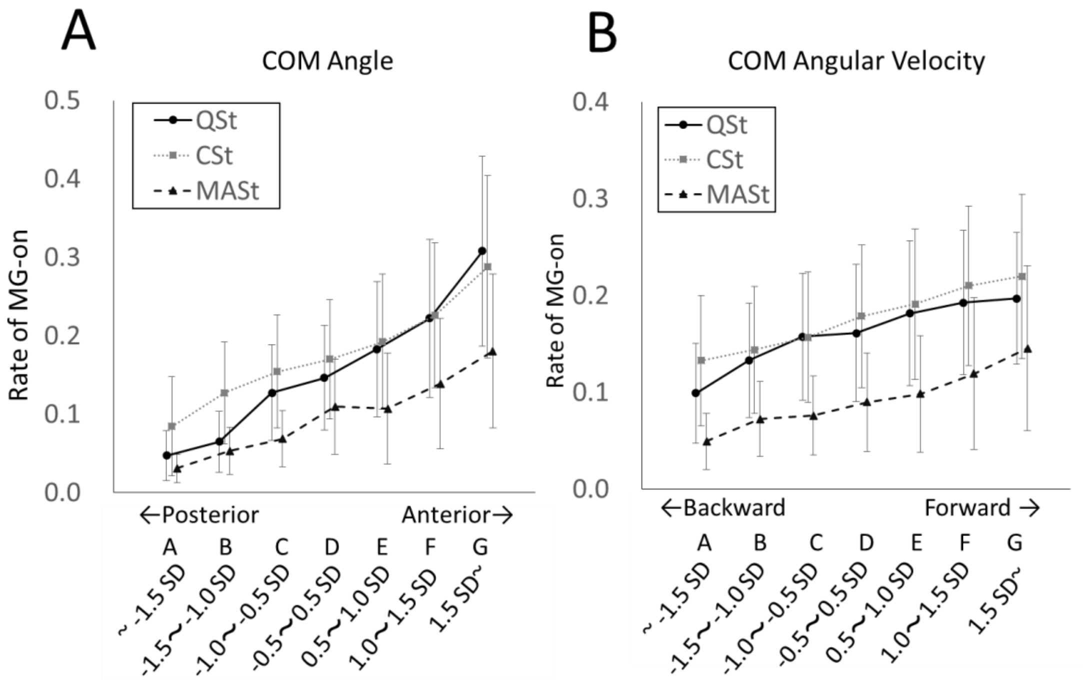

Fig. 3-11. The on-rate of the MG activation relate to COM angle (A) and COM angular velocity.

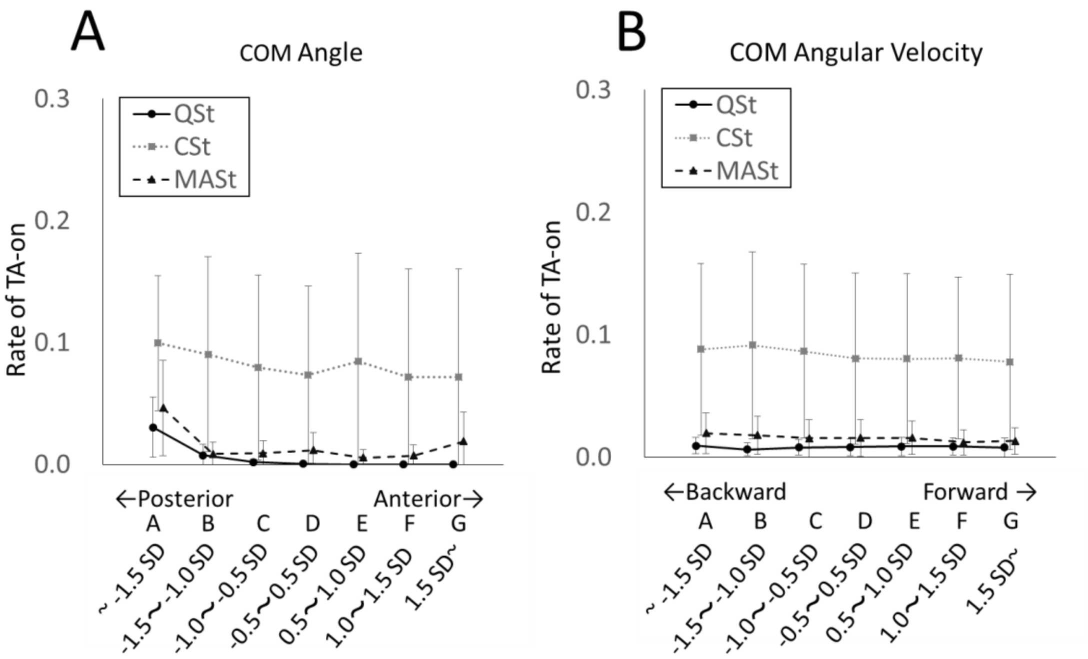

Fig. 3-12. The on-rate of the TA activation relate to COM angle (A) and COM angular velocity.

CHAPTER 4

EFFECT OF COGNITIVE LOAD ON BALANCE RECOVERY FOLLOWING PERTURBATION OF UPRIGHT STANCE.

4-1. Introduction

As a result of examining the effect of cognitive intervention on the posture of quiet standing, it became clear that the execution of simultaneous cognitive task reduces the amplitude of postural sway (Experiment 1 of Study 1). The reduction of the postural sway amplitude may be regarded as getting more stabilization. On the other hand, there was no change in the estimated joints stiffness and the strength of coordination between lower leg joints, which were considered to reflect the postural control strategy (Experiment 2 of Study 1). Further examination to evaluate the change in muscle activity revealed that the activation rate of the gastrocnemius is decreasing in dual-task condition (Experiment 3 of Study 1). This result suggests that the decrease in the amplitude of postural sway is not a postural stabilization. In addition, we show the possibility that a deterioration of postural control may occur with the decrease in the amplitude of postural sway. In this case, simultaneous cognitive task execution should degrade the compensatory postural response after a perturbation (i.e. grater postural sway).

A compensatory balancing reaction to sudden perturbation is necessary for maintenance of stable standing posture. Generally, when the balance is disturbed by mechanical perturbation during standing, stretch reflex and subsequent postural response initiated 40 to 120 ms after the start of perturbation. These are classified by the latency to the appearance of the response. The response that appears in the short latency is a monosynaptic Ia-mediated spinal reflex. It is known that its amplitude modulated by the magnitude of stimuli, prediction and background EMG activity. The later components, including medium- (MLR) and long-latency responses (LLR), are modified more flexibly by the subject's prior experience (Horak et al., 1989), threat of falling (Carpenter et al., 2004) or predictions (Fujio et al., 2016).

Brown et al. (1999) demonstrated delays in performing a counting backward task during the execution of a compensatory balancing reaction in young and older adults. Then, their group showed that the decreased in lower limb muscles activity response to a sudden sliding perturbation during performing the concurrent cognitive task compare to simply standing especially in older adults (Rankin et al., 2000). The authors highlighted that the cognitive task interventional effect was seen in muscle response amplitude for the agonist (MG) after backward perturbation and antagonist (TA) muscle after forward perturbation at later phase of response (between 350 to 500 ms). Besides, in previous studies that measured posturography during the postural response to both backward and forward perturbations, it is showed that the influence of performing of the concurrent task to postural response appears in the forward perturbation in young adults (Norrie et al., 2002; Quant et al., 2002). Taken together, it may suggest that the modulation of muscle activity, which is the agonist for response to disturbance (especially forward perturbation), may be more susceptible to the intervention of simultaneous cognitive tasks. In unperturbed quiet standing, decreased MG activation on-rate was shown in MASt condition (Experiment 3 of Study 1).

Modulation of postural control is emphasized in dynamic postural tasks. If the postural response after perturbation decreases during dual tasking, then postural control is deteriorated. If so, the response to perturbation during standing should also be modulated. Moreover, the differences in pathways that can be identified by postural response provide more detailed information on the impact of performing cognitive task.

Therefore, I aimed to clarify the influence of the cognitive task on posture response induced by anterior-posterior direction (backward and forward) disturbance during standing. In order to test its effect, subjects were required to maintain an upright standing against perturbation sliding horizontally to forward and backward.

4-2. Methods

Subjects

Ten healthy subjects (age 29.6 ± 5.6 years; height 173.1 ± 5.6 cm; weight 71.7 ± 9.3 kg) with no known neuromuscular or orthopedic disease participated in this study. All participants gave written informed consent in accordance with the Declaration of Helsinki. The Ethics Review Committee approved the protocol for Experimental Research with Human Subjects of the Graduate School of Arts and Sciences, The University of Tokyo.

Procedure

This experiment made use of a hydraulically activated movable platform system (Kyowa, Tokyo, Japan) that moved horizontally either forward or backward. The amplitude of movement was 7 cm, and the movement velocity was 30 cm/s. Subjects were instructed to stand barefoot on a movable platform with feet parallel 15 cm apart from the centers of both heels to try to maintain their balance throughout the experiment without taking a step. The subjects held their arms by their sides looking at a target placed at eye level 1.5m in front of the subjects. Subjects were exposed to 10 stimulations of perturbation during quiet standing (QSt) and with mental arithmetic (MASt). The direction of perturbation was constant in a trial. The minimum interval of stimulations was 10 s. The sufficient rests (at least 3min) were provided between each trial giving a series of stimuli to exclude the influence of fatigue.

Measurement

The ground reaction forces (GRFs) were measured by the platform. The GRF data were collected at a sampling rate of 1 kHz. The COP displacement in the anterior-posterior (AP) and the medio-lateral (ML) directions were calculated from the measured GRFs and low-pass filtered with a cutoff frequency of 10 Hz using a second-order low-pass Butterworth filter.

The disposable bipolar Ag/AgCl surface electrodes (diameter: 7 mm) were placed in a bipolar configuration over the subject's right Sol, MG and tibialis anterior (TA) to measure electromyogram (EMG) with an inter-electrode distance of 10 mm. Standard skin preparations using alcohol and tape for abrasion were applied before the attachment of electrodes. The EMG signals were stored with a sampling frequency of 10 kHz. Thin elastic bandages were wrapped to hold electrodes stably on the muscles and lead lines as well. EMG signals were amplified ×1000 using a bioelectric amplifier (MEG-6108, Nihon Kohden, Japan) with a bandpass filter (15–1000 Hz) and digitized at a sampling rate of 10 kHz.

Data analysis

Balance reactions to evoked perturbations were compared between QSt and MASt, using the AP COP excursion. The amplitude of the peak COP excursion and its timing. All the digitalized EMG signals were full-wave rectified after subtraction of the DC bias. EMG onsets were determined as the point when the signal rose greater than three standard deviations (SD) from the baseline level of activity that was calculated for 50 ms prior to the stimulation of the perturbation. For each subject, integrated EMG (iEMG) in agonist (Sol, MG for the backward perturbation, TA for the forward perturbation) were obtained by integration over a window from 36 to 500 ms following perturbation onset. The integrated output was then divided by the time at designated integration bins of 36–70, 70–150, 150–350, and 350–500 ms (Fig.4-1) according to the methods of previous study (Rankin et al., 2000).

Statistical analysis

Paired t-tests were performed to examine the effects of two task conditions (i.e., QSt and MASt) on the peak time of COP, the amplitude of the peak COP, EMGs onset. Two-way ANOVAs for repeated measures were used to assess the effects of the task (QSt and MASt) and time (36–70, 70–150, 150–350, and 350–500 ms) on EMG response (iEMG). When the two-way factorial ANOVA test showed significant interaction effects, simple main effect tests were conducted to examine the source of the significant interactions. Then, each significant simple main effect of task was followed by the Bonferroni post hoc test. The significance level was set at $p < 0.05$.

4-3. Results

Peak timing and displacement of COP

Figure 4-3 shows representative ensemble-averaged EMG and kinematic recordings among different conditions from a single subject. The mean values of the peak timing and peak displacement of COP are shown in Figure 4-3A and B, respectively. There was no significant difference in the peak timing of COP among the conditions in both directions of perturbation (backward perturbation: $p = 0.93$, forward perturbation: $p = 0.36$). In the forward perturbation, the peak displacement of COP in the MASt condition was greater (i.e. the COP traveled farther backward following the forward perturbation) than that in the QSt condition ($p < 0.01$). There was no significant difference in the peak displacement of COP following the backward perturbation ($p = 0.24$).

EMG response

The mean values of the onset of EMGs from agonist are shown in Figure 4-4. There was no significant difference in the onset of EMG from agonist for each direction of perturbation (Sol: $p = 0.50$, MG: $p = 0.41$, and TA: $p = 0.83$). Figure 4-5 represents the mean value of iEMGs from agonist divided by the time at 4 bins. In Sol (Figure 4-5A), only earliest bin (36-70 ms) showed the significant difference between two conditions, and it was slightly greater in the MASt condition than in the QSt condition ($p < 0.05$). There were no main effects of task in all bin of the iEMG (Sol: $F_{[1,72]} = 0.02$, $p = 0.88$; MG: $F_{[1,72]} = 0.01$, $p = 0.92$; TA: $F_{[1,72]} = 0.01$, $p = 0.92$). There was also no age × task interaction for the iEMG (Sol: $F_{[3,72]} = 0.06$, $p = 0.98$; MG: $F_{[3,72]} = 0.02$, $p = 0.99$; TA: $F_{[3,72]} = 0.09$, $p = 0.96$).

4-4. Discussion

In order to clarify the influence of the cognitive task on posture response induced by anterior-posterior direction (backward and forward) disturbance during standing, I evaluated the spatial and the temporal characteristics of postural sway (COP trajectory) and muscle activity. This was achieved by examination of the neuromuscular responses elicited by a balance disturbance when a mental arithmetic task was performed versus a simply quiet standing condition.

The peak value of COP displacement after the forward perturbation was slightly larger than QSt (no cognitive task) condition. This means that the cognitive task led to a modest change in the balance response to the forward sliding perturbation, whereas this value in the backward perturbation did not change. This confirmed the instability of the posture under the MASt condition suggested in Study 1. In temporal characteristics, there is no effect of cognitive task on peak COP timing regardless of directions of perturbation. These results are consistent with those of Norrie et al. (2002), who found that concurrent motor tracking task led to lager the COP excursion and greater oscillation of the COP in forward-translation. They suggested that the early phase of the postural reaction was not affected by performing the additional task, whereas later components of the reaction (they assumed the response during 250 ms or more after perturbation onset as the later components) showed evidence of heightened instability (increased COP excursion) in comparison to single tasking trials. In the previous study, it is predicted that COP response theoretically has a considerable lag (150–250 ms) as a neuromuscular transmission delay (Winter et al., 1998). Considering the results that peak COP timing is around 250 ms and the peak value of COP displacement was larger in dual tasking condition, there is no denying the possibility that cognitive processing could have been affected modulating or programming features of the response of earlier phase or around 250 ms.

In response of muscle activity following perturbation, there is no effect of cognitive task on the timing of EMG onset and iEMG for three muscles regardless of directions of perturbation. Previous studies (Rankin et al., 2000) showed a decrease in iEMG magnitude of MG and TA due to the performance of the mental calculation task in the later phase (350-500 ms) although it should be noted that the subjects included the elderly group. Considering the fact that a greater sway to the backward after forward perturbation was observed also in the result of this study, the activity of muscles as the agonist was expected to reduce. It was proposed that the perturbation-evoked balance reaction comprises at least two phases (Nashner and Cordo, 1981). An initial one is an automatic phase that does not require attentional resources and a later one is an attention-demanding phase likely associated with efforts to regain a state of equilibrium. Especially, it is well known that the activity of the TA is more under the control of the supraspinal mechanism (Armstrong, 1988). It is shown that the long-latency reflex in TA muscle is at least partly transcortical in studies on the stretch reflex (Petersen et al., 1998; Doornik et al., 2004). Therefore, there is a possibility that intervention of a cognitive task requiring central processing may be greatly affected. In this result, however, a decrease in muscle activity that explains the change in COP response was not confirmed.

The absence of change in EMG activity may be due to the fact that the magnitude and speed of the disturbance were smaller and slower than that study (the amplitude of movement was 15 cm and the movement velocities ranged from 20 to 60 cm/s). The decreasing trend (-15% compare to QSt) of later phase (350-500 ms) of TA in dual tasking condition may support this. The latencies of EMGs (average 96.5 ± 11.2 ms) were comparable to previous studies using similar sliding disturbances (Rankin et al., 2000; Norrie et al., 2002), but slower than the latency of stretch reflexes of ankle muscles induced by rotational perturbation (sudden drops of support surface) while standing (Horak and Nashner, 1986). It is known that stretch reflexes and subsequent correcting responses are altered by the magnitude of stimuli. The horizontal sliding disturbances used this study may have been insufficient to detect the difference of the muscle activity response between the conditions. The possibility of neural accommodation in sub-threshold level cannot be denied.

It is known that the response to a disturbance that threatens the stability of posture changes its strategy of compensation depending on its impact and the condition of the subjects (i.e. age, predictions) (Nashner and Cordo, 1981; Horak and Nashner, 1986). The ankle joint takes part in main role for posture control in the upright standing (see Chapter 1), but there is a high possibility that the hip strategy is taken when the subject is given greater impact of perturbation (Horak and Nashner, 1986). It may changes to the hip strategy may be occurring when the cognitive task is carried out, although the evaluation of the muscles around ankle joint was conducted in this study.

In summary, I found that the peak value of the COP trajectory after the forward perturbation increased in the mental arithmetic condition. This result indicates that performing the simultaneous cognitive task attenuates the postural response to postural disturbance. However, no change in muscle activity supporting this decline was observed.

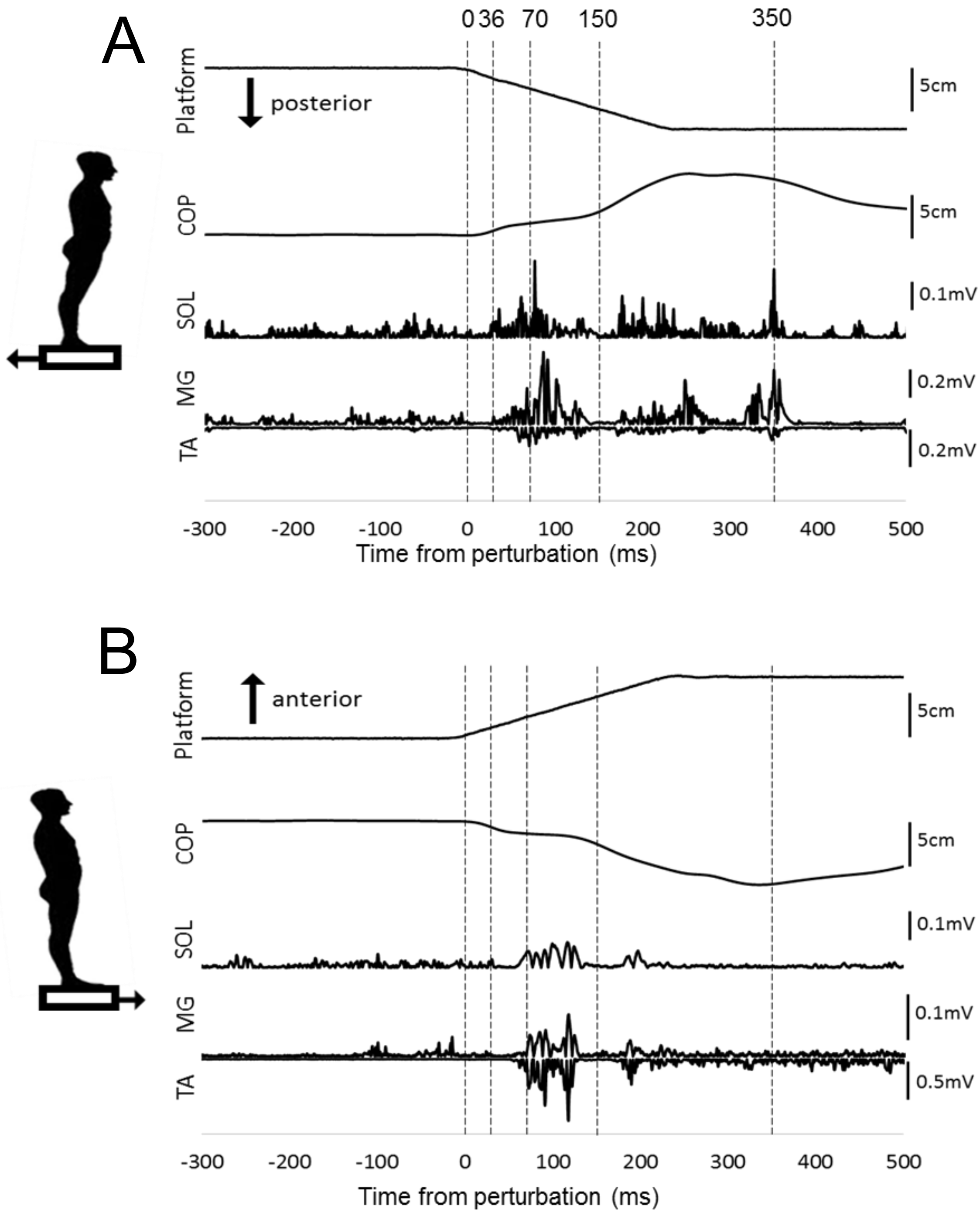

Fig.4-1. Example of response includes anterior (A)- posterior (B) center of pressure (COP) excursion and EMG activity in quiet standing condition.

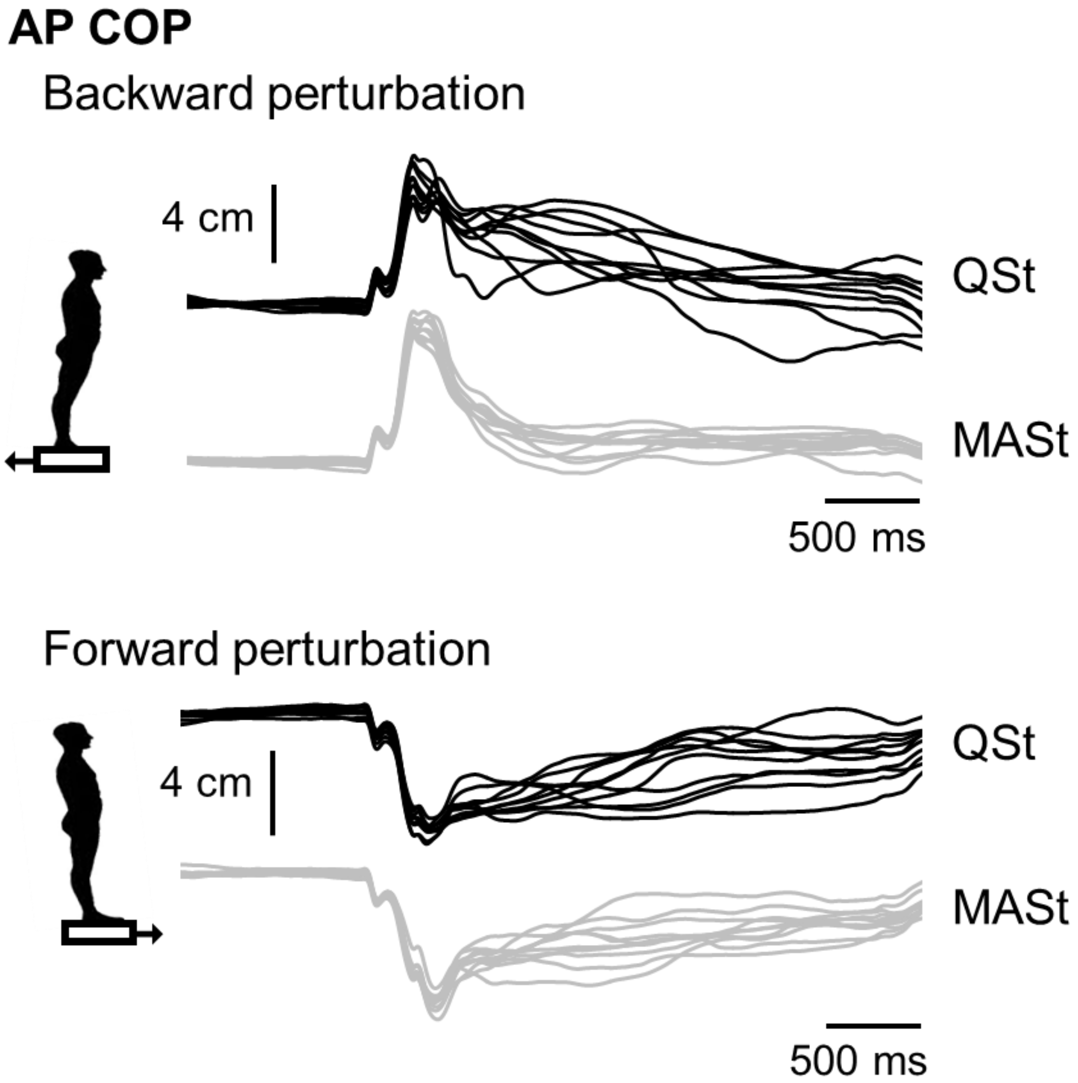

Fig.4-2. Anterior–posterior (AP) center of pressure (COP) displacements of all trials in one subject. The black lines represent data during quiet standing (QSt) condition and the gray lines represent data during mental arithmetic standing (MASt) conditions. Upper part represents the response to the backward perturbation, and lower panel represents the response to the forward perturbation.

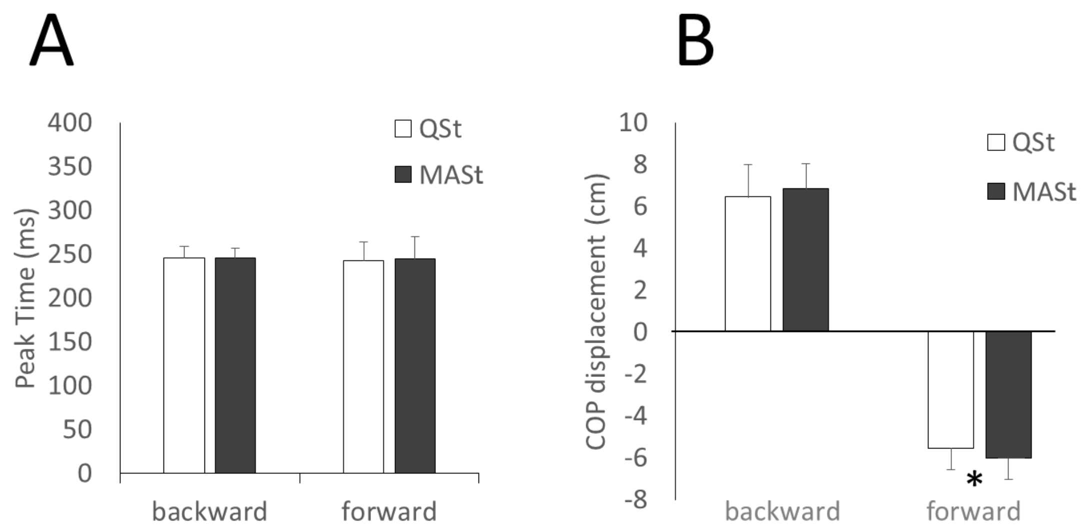

Fig.4-3. **A.** Average peak COP time (measured time between the perturbation onsets to the first peak in COP displacement), **B.** amplitude of the first peak in COP displacement after each direction perturbations. (*$p < 0.05$)

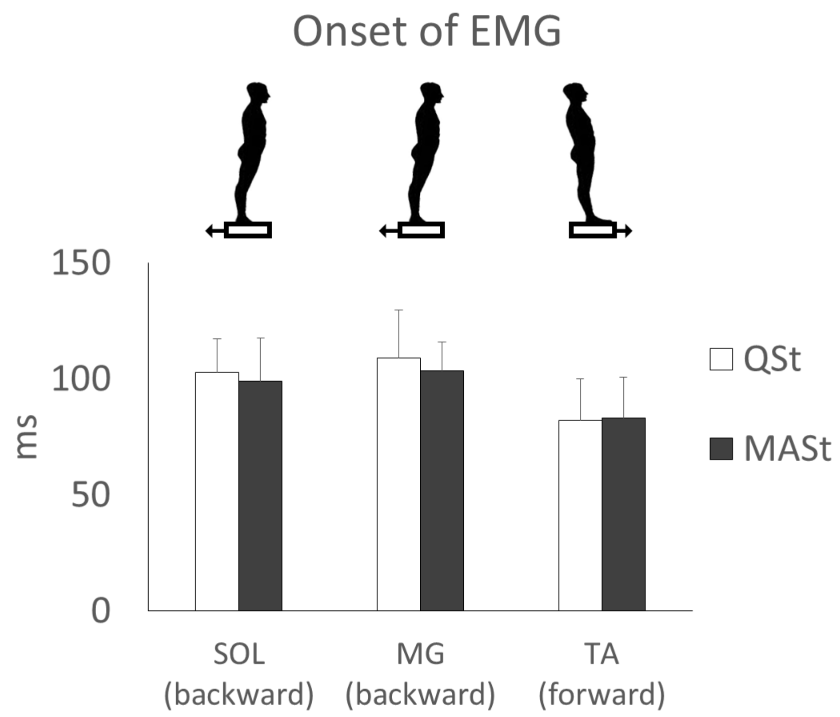

Fig.4-4. Average onset of evoked EMG reactions from ankle muscles, relative to onset of perturbation. The data of SOL, MG represents data in the backward perturbation, and the data of TA represents data in the forward perturbation.

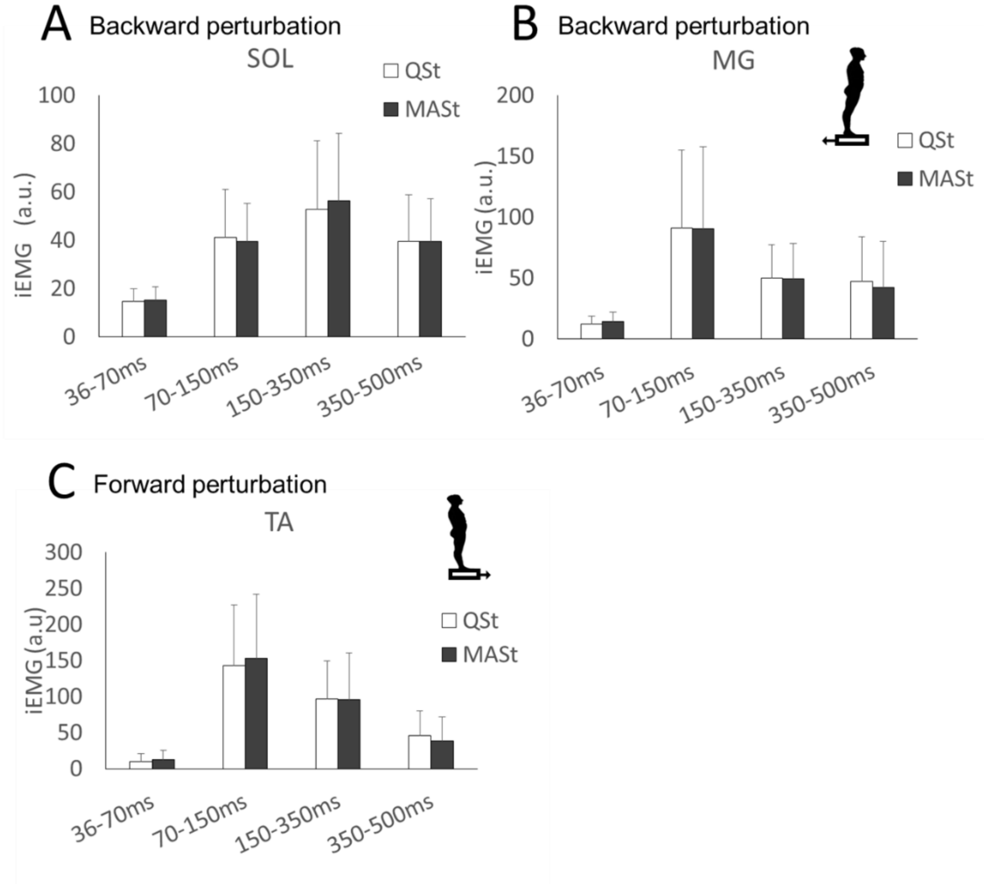

Fig.4-4. The integrated EMG in each integration bins (36–70, 70–150, 150–350, and 350–500 ms). **A** The data of SOL represents data in the backward perturbation. **B** The data of MG represents data in the backward perturbation. **C** The data of TA represents data in the forward perturbation.

CHAPTER 5

EFFECT OF COGNITIVE LOAD ON SPINAL EXCITABILITY DURING UPRIGHT STANDING.

5-1. Introduction

I investigated the influence of cognitive intervention on postural control by evaluation of postural sway and muscle activity so far during quiet standing and maintaining upright posture after perturbation. As a result, it was suggested that the mental arithmetic as a cognitive task reduces the stability of standing posture. Furthermore, it is suggested that the correction action to maintaining posture increases as shown in sway velocity and on-rate of muscle activation increase when paying attention to own posture. In this study, I focused on H-reflex by which to evaluate the excitability of the spinal reflex pathway, in order to obtain further insight on the mechanism of change in the postural control.

The modulation of the excitability of the spinal reflex pathway in postural control have been studied since long before (see CHAPTER 1). The H-reflex is a monosynaptic reflex that reflects the influence of group Ia monosynaptic projection on spinal α-motoneuron activation. Previous studies have found the Sol H-reflex amplitude to be modulated according to the postural situation. For example, the unstable postures, such as standing or walking, are associated with smaller H-reflex amplitudes compared with stable postures, such as lying or sitting (Koceja et al., 1993; Tokuno et al., 2007; Zehr, 2002). Furthermore, it is suggested that modulation of H-reflex is dependent on the phase and direction of postural sway (Tokuno et al., 2008). It is also known that H-reflex decreases in challenging postural tasks that require dynamic and conscious control (Koceja et al. 1993; Hoffman and Koceja 1995; Earles et al. 2000). Therefore, paying more attention to the standing posture than usual may lead greater suppression of the H-reflex.

In the study exploring about the influence of cognitive intervention on the excitability of the spinal reflex pathway in postural control, opinions about the change in amplitude of H-reflex have been divided. Weaver et al. (2012) investigated the excitability of the spinal reflex pathway during dual tasking. They used an auditory reaction time task as a concurrent cognitive task. Results indicated that the H-reflex (evaluated by H-max/M-max) was 6.4 ± 2.3% smaller when dual tasking compared to single tasking. They suggested that a reduction in spinal excitability would minimize unintended, reflex-mediated contractions of the plantar flexors and consequently, postural stability would be ensured when less cognitive resources are allocated to postural control due to the performance of a concurrent cognitive task. Later, Baudry and Gaillard (2014) investigated effects of dual tasking on the excitability of the spinal reflex by using Stroop task in young and elderly adults. In the result, the H-reflex amplitude (% Mmax) did not change with the cognitive load during dual task. Therefore, they concluded that increasing the difficulty of a cognitive task does not influence the efficacy of Ia afferent pathway to discharge Sol motoneuron during postural dual task in young and elderly adults.

Modulation of the amplitude of the H-reflex is mainly caused by presynaptic suppression (Hultborn et al., 1987). In addition to this, a direct pathway to the muscle spindle of sympathetic nerves has been confirmed (Barker and Saito, 1981), and the enhancement of the sympathetic nervous system also modifies the amplitude of the H-reflex. Changes in respiration during simultaneous mental arithmetic tasks observed in

Experiment 2 of Study 1 suggested changes in the autonomic nervous system (in particular, relative enhancement of the sympathetic nervous system). During the cognitive task at the sitting position, it has been confirmed that stretch reflex of the relaxed soleus muscle enhances (Hjortskov et al., 2005). It was also shown that this enhancement of the stretch reflex is not accompanied by an enhancement of H-reflex (Kamibayashi et al., 2009) during performing cognitive task only. During the cognitive tasking the enhancement of the sympathetic nervous system may occur. It was also suggested that H-reflex is more sensitive to presynaptic inhibition than stretch reflex (Morita et al., 1998; Andersen and Sinkjær, 1999). Taken together, it is suggested that modulation of presynaptic inhibition may occur when the modulation of the H-reflex amplitude is observed during standing.

Accordingly, I aimed to investigate the effect of the cognitive intervention during upright standing on the efficacy of the Ia afferent pathway to discharge motoneurons. To this end, I evaluated Sol H-reflexes elicited by stimulating the right tibial nerve in two cognitive interventions in contrast with simply quiet standing.

5-2. Methods

Eight healthy young males (age: 26.0 ± 3.2 years, height: 176.5 ± 6.9 cm, and weight: 71.9 ± 11.0 kg) participated in this experiments. Subjects had no history of neurological or cardiopulmonary disorders. The experimental procedures used in the study were in accordance with the declaration of Helsinki and were approved by the ethical standards of the committee on Human Experimentation at the Graduate School of Arts and Sciences, the University of Tokyo. All subjects gave their informed written consent after receiving a detailed explanation of the purpose, potential benefits, and risks involved in the execution of the study.

Procedure

Barefoot quietly on a form that was flush mounted with the ground with their eyes open and with feet parallel 15 cm apart between the centers of their heels. Subjects stood with their arms by their sides, facing forward with their eyes open, and fixated on a target 3 m in front at eye level.

Pairs of surface bipolar Ag/AgCl surface electrodes (Vitrode F-150S, 18 × 36 mm, 2 cm interelectrode distance; Nihon Kohden Corporation, Tokyo, Japan) were placed in a bipolar configuration over the subject's right Sol, MG and tibialis anterior (TA) to measure electromyogram (EMG). The M-wave and H-reflex amplitudes were obtained from electrodes placed in the right Sol within the windows from 5 ms to 25 ms and from 30 ms to 50 ms, respectively. All EMG recordings were sampled at 4 kHz and were amplified (×1,000), bandpass filtered (15–3k Hz) with an amplifier (MEG-610B, Nihon Kohden Corporation). The Sol H-reflexes were elicited by stimulating the right tibial nerve (rectangular pulse, 1-ms duration) with an electrical stimulator (DS7A, Digitimer, UK), with a cathode (18 × 36 mm) on the popliteal fossa and an anode (50 × 50 mm) placed over the patella. The electrodes were secured with adhesive tape to prevent their movement during the experiment. The minimum interval of stimulations was 5 s. Before a set of tasks, a maximal M response (Mmax) was evoked and measured as the peak-to-peak amplitude. The size of the M wave was kept around 5% (4-6%) of the Mmax and it was continually monitored to assess the stability of stimulation.

Conditions

Three standing conditions were executed within 3 minutes' trials. In the quiet standing (QSt) condition, subjects were asked to relax and maintain an upright stance. In the conscious standing (CSt) condition, they were asked to concentrate on their postural sway and reduce it as much as possible. In the standing with mental arithmetic (MASt) condition, they were asked to mentally count backward from randomly selected 3-digit numbers in steps of a single digit number (i.e. 6, 7, 8, 9) as fast and as accurately as possible and report the final number verbally at the end of each trial. Before the experiment, subjects practiced this task several times. Three trials were conducted under each condition in a randomized order. Short rests of 1 to 3 min were provided between each trial to exclude the influence of fatigue. Sixty successful stimuli in each condition were analyzed.

Data analysis

The peak-to-peak amplitudes of the H-reflex, M-wave, and Mmax evoked at each condition were measured offline. To minimize individual variability, the H-reflex and M-wave amplitude were each expressed as a percentage of Mmax (H-amplitude/Mmax and M-amplitude/Mmax, respectively). The background electromyographic activity (BGA) levels in the right Sol, MG and TA muscles were determined as the root mean square (RMS) values of the EMG signals for 50 ms before stimulation.

Statistical analysis

To compare each variable among the three task conditions (QSt, CSt, MASt), a one-way analysis of variance (ANOVA) with repeated measures was performed to examine the effects of the task condition on the dependent measures. The sequentially rejective

Bonferroni correction (the Holm method) was used to adjust the p values obtained in the post hoc analyses. All significance level was set at $p < 0.05$.

5-3. Results

Background EMGs

The BGA in each conditions are shown in Table 6-1. There was no significant main effect of postural condition for three muscles (Sol: $F_{(2,7)} = 1.04$, $p = 0.33$; MG: $F_{(2,7)} = 1.79$, $p = 0.16$; TA: $F_{(2,7)} = 1.71$, $p = 0.82$).

H-reflex amplitude

The mean M-waves amplitude (Table 6-1) were not significantly different among three conditions (QSt: 4.86 ± 0.29 %Mmax; CSt: 4.92 ± 0.35 %Mmax; MASt: 4.83 ± 0.26 %Mmax: $F_{(2,7)} = 0.29$, $p = 0.73$), a parameter that indicates stable stimulation and recording procedures. The normalized H-reflex amplitude (%M-max) during each condition are shown in Fig. 6-1. There was a significant main effect of task condition on the normalized H-reflex amplitude ($F_{(2,7)} = 6.31$., $p < 0.01$). Post hoc comparisons showed that the normalized H-reflex amplitude in the MASt condition was significantly greater as compared with QSt condition (mean difference of 7.8 ± 6.6, $p < 0.05$).

5-4. Discussion

In this study, I aimed to investigate the effect of the cognitive load during upright standing on the efficacy of the Ia afferent pathway to discharge motoneurons. The result showed that H-reflex was slightly but significantly greater in MASt condition compared with QSt condition. The result of this study means that the mental arithmetic task enhances the spinal excitability in soleus muscle, which showed tonic activities during standing.

The results were in line with studies that observed facilitations in the H-reflex during brief mental arithmetic in the relaxed soleus muscle (Bussel et al., 1978; Zehr and Stein, 1999; Gregory et al., 2001; Rossi-Durand, 2002). A candidate for this mechanism is presynaptic inhibition. Modulations in the central command can increase H-reflex amplitudes (Hultborn et al., 1987; Nielsen and Kagamihara, 1993) via reduction of presynaptic inhibition (i.e., presynaptic disinhibition). Other potential pathways to greater H-reflex amplitude involve sympathetic outflow. It is known that the mental tasks heighten sympathetic outflow during sitting (Callister et al., 1992; Mark et al., 1985; Ng et al., 1994) and that muscle sympathetic nerve activity shows sustained increase during the mental task (Anderson et al., 1991; Callister et al., 1992; Matsukawa et al., 1995). It is also showed that muscle sympathetic nerve innervates muscle spindles in animal study (Barker and Saito, 1981; Grassi et al., 1993). Contrary, it was suggested that the H-reflex is more sensitive to presynaptic inhibition compared with the stretch reflex (Morita et al., 1998; Andersen and Sinkjær, 1999). Moreover, it was reported a sustained facilitation not in the H-reflex but in the stretch reflex of the relaxed soleus muscle during the mental arithmetic task when subjects were sitting (Kamibayashi et al., 2009). Taken together, the increase in H-reflex specifically observed in MASt condition is more likely related to merely an increase in pre-presynaptic disinhibition, or the modulation combined with an increase in pre-presynaptic disinhibition and enhancement of reflexes by the sympathetic nervous system.

On the other hand, attention to its standing posture was not significant, but it tended to decrease the H-reflex amplitude. In the previous research, it is known that H-reflex decreases in challenging postural tasks that require dynamic control (Koceja et al. 1993;

Hoffman and Koceja 1995; Earles et al. 2000). Some researchers (Sibley et al., 2007; Horslen et al., 2013) showed that H-reflex amplitude was attenuated in the high (i.e. the edge of elevated platform). It may be because suppression of the H-reflex amplitude during standing avoids excessive autogenic excitation of the Sol motoneuron and helps to receive reliably central descending commands (Solopova et al., 2003). In a study of monkeys (Seki et al., 2003), it was suggested that this presynaptic inhibition has an appropriate functional consequence for motion generation and it effectively forms a basis for an increase in perceptual threshold during active exercise by effectively suppressing less important sensory input.

In this study, I aimed to investigate the effect of the cognitive intervention during upright standing on the efficacy of the Ia afferent pathway to discharge motoneurons. The result showed that H-reflex was slightly but significantly greater in MASt condition compared with QSt condition, whereas background EMGs activities of three muscles were not different. This means that the mental arithmetic task enhances the efficacy of the Ia afferent pathway. On the other hand, attention to its standing posture was not significant, but it tended to decrease the H-reflex amplitude.

Table.5-1. Background EMG and M-wave amplitudes in each condition

	QSt	CSt	MASt
SOL BGA (μV)	26.23 ± 8.56	26.50 ± 9.25	29.49 ± 10.52
MG BGA (μV)	39.00 ± 23.05	40.27 ± 23.02	37.63 ± 21.54
TA BGA (μV)	82.03 ± 42.05	82.98 ± 39.94	83.19 ± 39.95
M-wave (%Mmax)	4.86 ± 0.29	4.92 ± 0.35	4.83 ± 0.26

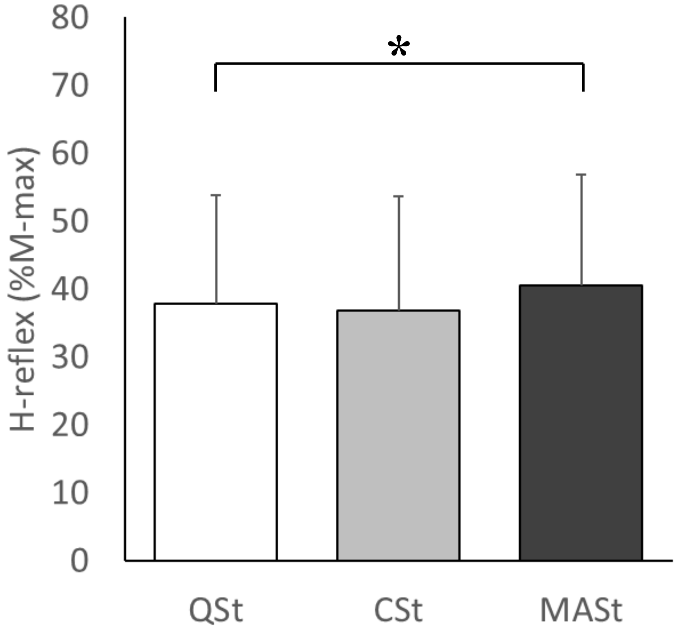

Fig.5-1. H-reflex amplitudes in each condition (with SD bars). (*p < 0.05)

CHAPTER 6

GENERAL DISCUSSION

In this thesis, I aimed to examine the effects of cognitive intervention on human postural control. To this end, I set two cognitive conditions, in which the subjects were instructed (1) to focus more attention on their postural sway (i.e., the CSt condition) and (2) to perform a secondary cognitive task (mental arithmetic task; i.e., the MASt condition), and compared the results to those under the normal quiet standing (QSt) condition.

6-1. Effects of directing attention to postural sway

From the results of Experiment 1, it is revealed that the COP velocity is greater under the conscious standing condition than under the quiet standing condition. This result is in line with previous studies (Ueta et al., 2014; Reynolds, 2010). It has been also suggested that the COP velocity is related to providing information on postural corrections required to maintain postural stability (Bonnet & Baudry, 2016; Maki et al., 1990). Some researchers have suggested that directing attention to one's own postural sway leads the control of voluntary processes and that enhanced motor outputs, such as increased joint stiffness, damping, and co-contraction of the ankle muscles, account for the increased COP velocity (Ueta et al., 2014; Reynolds, 2010).

In the results, however, there were no significant differences in joint movement variability, stiffness, the strength of multi-joint coordination (the UCM ratios), or overall muscle activity during trials under the QSt and CSt conditions. This means that the strategy of stiffening joints that has been considered a candidate for the mechanism of increase in sway velocity in conscious standing in the previous study seems unlikely to explain the increased COP velocity. Instead, the rate of TA-on was significantly larger in the CSt condition as compared with the other two conditions. Considering that there is no increase in sustained tonic muscle contraction as represented by the RMS of EMGs, it is suggested that the increased activity of TA is an increase in phasic activity. The absence of an increase in the estimated joint stiffness also supports this. Therefore, it is suggested that increased activity of TA might increase the frequency of voluntary postural correction. On the other hand, the rate of MG-on did not change under the CSt condition. This may reflect differences in the characteristics of muscles and dominant nerves.

In Study 3, there was no statistically significant difference in the H-reflex amplitude of the Sol, but directing attention to postural sway tended to decrease the H-reflex. One of the causes of the change in the H-reflex was the adjustment of excitability of the spinal reflex pathway by presynaptic inhibition. It is known that Ia afferents receive input from the CNS by presynaptic inhibition through interneurons. It may be because suppression of the H-reflex amplitude during standing avoids excessive autogenic excitation of the Sol motoneuron and helps to receive central descending commands reliably (Solopova et al., 2003). Previously, it was shown that the excitability of the corticospinal tract (evaluated by MEPs) of the Sol and TA increases with the increasing postural task difficulty considered to require more attention to one's own posture (Tokuno et al., 2018). Taken together, the results indicate that greater attention to own posture cause increases in input from central command.

The results of this thesis show that the influence of directing attention to one's own postural sway differs depending on the muscle and may be related to the characteristics of the role that muscle plays during standing. As a result, it is suggested that CNS increases conscious and frequent postural correction to achieve the objective of reducing the postural sway when subjects were asked to concentrate on their postural sway and reduce it as much as possible.

6-2. Effects of executing a cognitive task

It was shown that the COP amplitude decreased under the MASt condition as compared with the QSt condition. This result is in line with previous studies (Andersson et al., 2002; Riley et al., 2003; Stins et al., 2011; Ueta et al., 2014; Richer et al., 2017). This effect has been suggested to be attributable to the utilization of the more automatic control processes of the standing posture (Ueta et al., 2014) or increased arousal (Riley et al., 2003). In this thesis, there are two main findings of the reduction of COP amplitude during the mental arithmetic task.

First, the results demonstrated that performing the mental arithmetic task during standing reduced movements of the ankle and hip joints (Experiment 1 of Study 1) without the joint stiffness and strength of joint coordination for the COM. Second, the results of measuring respiratory movement (Experiment 2 of Study 1) showed that the respiratory rate increased and the respiratory amplitude decreased during the mental arithmetic task. Further, there was a significant positive correlation between the change in respiration and the COP amplitude. Therefore, decreased postural disturbance through reduced respiratory amplitude, but not improved postural control, reduced the amplitude of COP displacement.

As mentioned above, the MG's muscle activity and low frequency power of the MG are decreased during unperturbed standing (Experiment 3 of Study 1), and postural sway after forward perturbation increases (Study 2) when a concurrent cognitive task is performed. The results of analysis of the EMG activation rate and frequency domain analysis of the EMG suggested that the activation of the ankle plantar flexor was modulated by some neural mechanisms. On the other hand, there was no modulation in the ankle dorsi flexor during the quiet standing task. This seems to be due to the fact that its activity frequency is already low during the posture task alone. Therefore, no significant decrease was observed. In the response after the perturbation, which has a higher impact than quiet standing, increased oscillation after forward perturbation may support that it cannot be said that the performance of additional cognitive tasks improves postural control.

It was found that the H-reflex increased markedly in the assessment of excitability of the spinal reflex pathway during the dual task (Study 3). Increased presynaptic disinhibition and increased sympathetic outflow are assumed as possible pathways for increased H-reflex. Changes in respiration during simultaneous mental arithmetic tasks observed in Experiment 2 of Study 1 suggested changes in the autonomic nervous system. Increased sympathetic outflow may increase muscle tone. However, a sustained increase in muscle activity has not been confirmed (Experiment 3 of Study 1). The possibility that the modulation in a subthreshold has occurred cannot be denied. On the other hand, it was suggested that the modulation of presynaptic inhibition also occurs because the H-reflex was increased under the dual task condition. If presynaptic inhibition is affected by commands from the CNS, active postural control (by phasic muscle contraction) by the CNS may have decreased. In other words, it was suggested that during the simultaneous cognitive tasks, postural adjustment at the central origin decreased and changed to postural control that relied more on spinal reflexes.

The results for the H-reflex under the dual tasking condition clearly contrast with the results seen when directing attention to one’s own postural sway. Consequently, the results of this thesis showed a neural mechanism behind the utilization of the automatic control process when performing a concurrent cognitive task. The relationship between this modulation of the process of postural control and the COP amplitude is unclear, and it is difficult to examine whether this modulation reduces the COP amplitude.

6-3. Conclusion of the thesis

In order to examine the effects of the cognitive load on human postural control during standing, I have conducted three studies. As a result, it was revealed that the features of postural control vary under each task. It was shown for the first time that the decrease in respiration amplitude is related to the decrease in the amplitude of postural sway during posture–cognitive dual tasking in young adults. Furthermore, the results suggested that postural control became more dependent on the automatic process of the reflex pathway during the performance of cognitive tasks. On the other hand, it was suggested that paying more attention to one's own standing posture leads to the greater contribution of supraspinal control of the standing posture and a subsequent increase in activity to maintain an upright stance.

REFERENCES

1. Akram, S. B., Frank, J. S., Patla, A. E. and Allum, J. H. J. Balance control during continuous rotational perturbations of the support surface. Gait Posture 27, 393–398 (2008).

2. Allen, M. T. and Crowell, M. D. Patterns of autonomic response during laboratory stressors. Psychophysiology 26, 603–614 (1989).

3. Andersen, J. B. and Sinkjzr, T. The stretch reflex and H-reflex of the human soleus muscle during walking. Motor Control 3, 151–157 (1999).

4. Andersson, G., Hagman, J., Talianzadeh, R., Svedberg, A. and Larsen, H. C. Effect of cognitive load on postural control. Brain Res. Bull. 58, 135–139 (2002).

5. Aramaki, Y. et al. Reciprocal angular acceleration of the ankle and hip joints during quiet standing in humans. Exp. Brain Res. 136, 463–473 (2001).

6. Armstrong, B. Y. D. M. The supraspinal control of mammalian locomotion. J. Physiol. 405, 1–37 (1988).

7. Aspects, P. Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? Age Ageing 35, 7–11 (2006).

8. Balasubramaniam, R. and Wing, A. M. The dynamics of standing balance. Trends Cognitive Sciences 6, 531-536 (2002).

9. Barker, D. and Saito, M. Autonomic innervation of receptors and muscle fibres in cat skeletal muscle. Proceeding R. Soc. B 212, 317–332 (1981).

10. Baudry, S. and Gaillard, V. Cognitive demand does not influence the responsiveness of homonymous Ia afferents pathway during postural dual task in young and elderly adults. Eur. J. Appl. Physiol. 114, 295–303 (2014).

11. Bonnet, C. T. and Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture 48, 57–63 (2016A).

12. Bonnet, C. T. and Baudry S. A functional synergistic model to explain postural control during precise visual tasks. Gait and Posture 50, 120–125 (2016B).

13. Brown, L. A., Shumway-cook, A. and Woollacott, M. H. Attentional demands and postural recovery: The effects of aging. J. Gerontol. 54A, 165–171 (1999).

14. Burleigh, A., Horak, F., Nutt, J. and Frank, J. Levodopa reduces muscle Tone and lower extremity tremor in Parkinson's disease. Can. J. Neurol. Sci. 22, 280–285 (1995).

15. Callister, B. Y. R., Suwarno, N. O. and Seals, D. R. Sympathetic activity is influenced by task difficulty and stress perception during mental challenge in humans. J. Physiol. 454, 373–387 (1992).

16. Caron, O., Fontanari, P., Cremieux, J. and Joulia, F. Effects of ventilation on body sway during human standing. Neurosci. Lett. 366, 6–9 (2004).

17. Carpenter, M. G., Frank, J. S., Silcher, C. P. and Peysar, G. W. The influence of postural threat on the control of upright stance. Exp. Brain Res. 210–218 (2001).

18. Casadio, M., Morasso, P. G. and Sanguineti, V. Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application. Gait Posture 21, 410–424 (2005).

19. Chen, Y. and Zhou, S. Soleus H-reflex and its relation to static postural control. Gait Posture 33, 169–178 (2011).

20. Creath, R., Kiemel, T., Horak, F., Peterka, R. and Jeka, J. A unified view of quiet and perturbed stance: simultaneous co-existing excitable modes. Neurosci. Lett. 377, 75–80 (2005).

21. Dault, M. C., Frank, J. S. and Allard, F. Influence of a visuo-spatial, verbal and central executive working memory task on postural control. Gait Posture 14, 110–116 (2001).

22. Dickstein, R., Shupert, C. L. and Horak, F. B. Fingertip touch improves postural stability in patients with peripheral neuropathy. Gait Posture 14, 238–247 (2001).

23. Dietz, V. Human neuronal control of automatic functional movements: interaction between central programs and afferent input. Physiol. Rev. 72, 33–58 (1992).

24. Doornik, J. Van, Masakado, Y., Sinkjaer, T. and Bo, J. The suppression of the long-latency stretch reflex in the human tibialis anterior muscle by transcranial magnetic stimulation. Exp. Brain Res. 403–406 (2004).

25. Earles, D. R., Koceja, D. M. and Shively, C. W. Environmental changes in

soleus H-reflex excitability in young and elderly subjects. Int. J. Neurosci. 105, 1–13 (2000).

26. Evans, C. M., Fellows, S. J., Rack, P. M. H., Ross, H. F. and Walters, D. K. W. Response of the normal human ankle joint to imposed sinusoidal movements. J. Physiol. 344, 483–502 (1983).

27. Fitzpatrick, R., Rogers, D. K. and Mccloskey, D. I. Stable human standing with lower-limb muscle afferents providing the only sensory input. J. Physiol. 480, 395–403 (1994).

28. Fraizer E. V., Mitra S. Methodological and interpretive issues in posture-cognition dual-tasking in upright stance. Gait and Posture 27, 271–279 (2008).

29. Fujio, K., Obata, H., Kawashima, N. and Nakazawa, K. The effects of temporal and spatial predictions on stretch reflexes of ankle flexor and extensor muscles while standing. PLoS One 11(7), e0158721 (2016).

30. Gage, W. H., Winter, D. A., Frank, J. S. and Adkin, A. L. Kinematic and kinetic validity of the inverted pendulum model in quiet standing. Gait Posture 19, 124–132 (2004).

31. Gatev, P., Thomas, S., Kepple, T. and Hallett, M. Feedforward ankle strategy of balance during quiet stance in adults. J. Physiol. (1999).

32. Gauchard, C., Vanc, G. and Perrin, P. P. Age-related part taken by attentional cognitive processes in standing postural control in a dual-task context. Gait Posture 25, 179–184 (2007).

33. Grassi, C., Deriu, F. and Passatore, M. Sympathetic nervous. J. Physiol. 469, 601–613 (1993).

34. Gurfinkel, V. S., Ivanenko, Y. P., Levik, Y. S. and Babakova, I. A. Kinesthetic reference for human posture. Neuroscience 68, 229–243 (1995).

35. Hauer, K., Pfisterer, Ã. M., Weber, Ã. C. and Wezler, Ã. N. Cognitive impairment decreases postural control during dual tasks in geriatric patients with a history of severe falls. J. Am. Geriatr. Soc. 51, 1638–1644 (2003).

36. Hayashi, R., Tako, K., Takahiko, T. and Yanagisawa, N. Comparison of amplitude of human soleus H-reflex during sitting and standing. Neurosci. Res. 13, 227–233 (1992).

37. Héroux, M. E., Dakin, C. J., Luu, B. L., Inglis, J. T. and Blouin, J.-S. Absence of

lateral gastrocnemius activity and differential motor unit behavior in soleus and medial gastrocnemius during standing balance. J. Appl. Physiol. 116, 140–148 (2014).

38. Hjortskov, N. et al. Sympathetic outflow enhances the stretch reflex response in the relaxed soleus muscle in humans. J. Appl. Physiol. 98, 1366–1370 (2005).

39. Hodges, P., Gurfinkel, V., Brumagne, S., Smith, T. and Cordo, P. Coexistence of stability and mobility in postural control: Evidence from postural compensation for respiration. Exp. Brain Res. 144, 293–302 (2002).

40. Hoffman, M. A. and Koceja, D. M. The effects of vision and task complexity on Hoffmann reflex gain. Brain Res. 700, 303–307 (1995).

41. Horak, F. B. et al. Diabetic neuropathy and surface sway-referencing disrupt somatosensory information for postural stability in stance. Somatosens. Mot. Res. 19, 316–326 (2002).

42. Horak, F. B. and Nashner, L. M. Central programming of postural movements: adaptation to altered support-surface configurations. J. Neurophysiol. 55, 1369–1381 (1986).

43. Horslen, B. C., Murnaghan, C. D., Inglis, J. T., Chua, R. and Carpenter, M. G. Effects of postural threat on spinal stretch reflexes: evidence for increased muscle spindle sensitivity? J. Neurophysiol. 110, 899–906 (2013).

44. Hsu, W. L., Scholz, J. P., Schöner, G., Jeka, J. J. and Kiemel, T. Control and estimation of posture during quiet stance depends on multijoint coordination. J. Neurophysiol. 97, 3024–3035 (2007).

45. Huang, C., Cherng, R. and Yang, Z. Modulation of soleus H reflex due to stance pattern and haptic stabilization of posture. J. Electromyogr. Kinesiol. 19, 492–499 (2009).

46. Huffman, J. L., Horslen, B. C., Carpenter, M. G. and Adkin, A. L. Does increased postural threat lead to more conscious control of posture? Gait Posture 30, 528–532 (2009).

47. Hultborn, H., Meunier, S., Pierrot-Deseilligny, E. and M. Shindo. Changes in presynaptic inhibition of Ia fibres at the onset of voluntary contraction in man. J. Physiol. 389, 757–772 (1987).

48. Huxhold, O., Li, S., Schmiedek, F. and Lindenberger, U. Dual-tasking postural

control: Aging and the effects of cognitive demand in conjunction with focus of attention. Brain Res. Bull. 69, 294–305 (2006).

49. Jeka, J., Oie, K., Scho, G., Oie, K. and Scho, G. Position and velocity coupling of postural sway to somatosensory drive. J. Neurophysiol. 79, 1661–1674 (1998).

50. Johannsson, J., Duchateau, J. and Baudry, S. Spinal and corticospinal pathways are differently modulated when standing at the bottom and the top of a three-step staircase in young and older adults. Eur. J. Appl. Physiol. 117, 1165–1174 (2017).

51. Kamibayashi, K. et al. Invariable H-reflex and sustained facilitation of stretch reflex with heightened sympathetic outflow. J. Electromyogr. Kinesiol. 19, 1053–1060 (2009).

52. Kerr, B. B., Condon, S. M. and Mcdonald, L. A. Cognitive spatial processing and the regulation of posture. J. Exp. Psychol. Hum. Percept. Perform. 11, (1985).

53. Kirshenbaum, N., Riach, C. L. and J, L. S. Non-linear development of postural

control and strategy use in young children: a longitudinal study. Exp. Brain Res. 140, 420–431 (2001).

54. Knikou, M. and Conway, B. A. Modulation of soleus H-re ¯ ex following ipsilateral mechanical loading of the sole of the foot in normal and complete spinal cord injured humans. Neurosci. Lett. 303, 107–110 (2001).

55. Koceja, D. M. Influence of quadriceps conditioning on soleus motoneuron excitability in young and old adults. Med. Sci. Sports Exerc. 25, 245–250 (1992).

56. Koceja, D. M., Trimble, M. H. and Earles, D. R. Inhibition of the soleus H-reflex in standing man. Brain Res. 629, 155–158 (1993).

57. Koceja, D. M., Markus, C. A. and Trimble, M. H. Postural modulation of the soleus H reflex in young and old subjects. Electroencephalogr. Clin. Neurophysiol. 97, 387–393 (1995).

58. Krishnamoorthy, V., Yang, J. and Scholz, J. P. Joint coordination during quiet stance: effects of vision. Exp. Brain Res. 164, 1–17 (2005).

59. Kuznetsov, N. A. and Riley, M. A. Effects of breathing on multijoint control of

center of mass position during upright stance. J. Mot. Behavio 44, 241–253 (2012).

60. Lacour, M., Bernard-Demanze, L., and Dumitrescu, M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clinical Neurophysiology38, 411—421 (2008).

61. Lajoie, Y. et al. Attentional demands for static and dynamic equilibrium. Exp. Brain Res. 97, 139–144 (1993).

62. Latash, M. L., Scholz, J. P. and Schöner, G. Motor control strategies revealed in the structure of motor variability. Exerc. Sport Sci. Rev. 30, 26–31 (2002).

63. Logan, G. D. and Gordon, R. D. Executive control of visual attention in dual-task situations. Physiol. Rev. 108, 393–434 (2001).

64. Loram, I. D. and Lakie, M. Human balancing of an inverted pendulum: position control by small, ballistic-like, throw and catch movements. J. Physiol. 540, 1111–1124 (2002).

65. Loram, I. D. and Lakie, M. Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for

stability. 1041–1053 (2002).

66. Loram, I. D., Maganaris, C. N. and Lakie, M. Human postural sway results from frequent, ballistic bias impulses by soleus and gastrocnemius. J. Physiol. 564, 295–311 (2005).

67. Maki, B. E., Holliday, P. J. and Fernie, G. R. Aging and postural control. A comparison of spontaneous- and induced-sway balance Tests. J. Am. Geriatr. Soc. 38, 1–9 (1990).

68. Martin, J. P. S., Schöner, G., Hsu, W. L., Jeka, J. J. and Horak, F. V. Motor equivalent control of the center of mass in response to support surface perturbations. Exp. Brain Res. 180, 163–179 (2007).

69. Masani, K., Vette, A. H., Abe, M. O. and Nakazawa, K. Center of pressure velocity reflects body acceleration rather than body velocity during quiet standing. Gait Posture 39, 946–952 (2014).

70. Masani K, Vette A. H., Kouzaki M., Kanehisa H, Fukunaga T, Popovic M. R. Larger center of pressure minus center of gravity in the elderly induces larger body acceleration during quiet standing. Neurosci. Lett. 422, 202–206 (2007).

71. Matsukawa, T., Sugiyama, Y. and Mano, T. Increased muscle sympathetic nerve activity during delayed auditory feedback in human. Jpn. J. Physiol. 45, 905–911 (1995).

72. Maurer, C. et al. Effect of chronic bilateral subthalamic nucleus (STN) stimulation on postural control in Parkinson's disease. Brain 126, 1146–1163 (2003).

73. Maurer, C. and Peterka, R. J. A new interpretation of spontaneous sway measures based on a simple model of Human postural control. J. Neurophysiol. 93, 189–200 (2005).

74. Maylor, E. A., Allison, S. and Wing, A. M. Effects of spatial and nonspatial cognitive activity on postural stability. Br. J. Psychol. 92, 319–338 (2001).

75. Morasso, P. G. and Sanguineti, V. Ankle muscle stiffness alone cannot stabilize balance during quiet standing. J. Neurophysiol. 88, 2157–2162 (2002).

76. Morita, H., Petersen, N., Christensen, L. O. D., Sinkjær, T. and Nielsen, J. Sensitivity of H-reflexes and stretch reflexes to presynaptic inhibition in humans. J. Neurophysiol. 80, 610–620 (1998).

77. Mulder, G., Mulder, L. J. M. Information processing and cardiovascular control. Psychophysiology 18, 392–402 (1981).

78. Nashner L. M, Cordo P. J. Relation of automatic postural responses and reaction-time voluntary movements of human leg muscles. Exp Brain Res 43,395–405 (1981).

79. Nakazawa, K. et al. Effects of loading and unloading of lower limb joints on the soleus H-reflex in standing humans. Clin. Neurophysiol. 115, 1296–1304 (2004).

80. Ng, V. et al. Sympathetic neural reactivity to stress does not increase with age in healthy humans. Am. J. Physiol. 267, H344-353 (1994).

81. Nielsen, J. and Kagamihara, Y. The regulation of presynaptic inhibition during co-contraction of antagonistic muscles in man. J. Physiol. 464, 575–593 (1993).

82. Norrie, R. G., Maki, B. E., Staines, W. R. and Mcilroy, W. E. The time course of attention shifts following perturbation of upright stance. Exp Brain Res 146, 315–321 (2002).

83. Obata, H., Kawashima, N., Ohtsuki, T. and Nakazawa, K. Aging effects on

posture-related modulation of stretch reflex excitability in the ankle muscles in humans. J. Electromyogr. Kinesiol. 22, 31–36 (2012).

84. Olivier, I., Cuisinier, R., Vaugoyeau, M., Nougier, V. and Assaiante, C. Age-related differences in cognitive and postural dual-task performance. Gait Posture 32, 494–499 (2010).

85. Park, E., Reimann, H. and Schöner, G. Coordination of muscle torques stabilizes upright standing posture: an UCM analysis. Exp. Brain Res. 234, 1757–1767 (2016).

86. Pashler, H. Dual-task interference in simple tasks: data and theory. 116, 220–244 (1994).

87. Pashler, H. Dual-Task Interference in Simple Tasks: Data and Theory. Psychol. Bull. 116, 220–244 (1994).

88. Pellecchia, G. L. Postural sway increases with attentional demands of concurrent cogniti v e task. Gait Posture 18, 29–34 (2003).

89. Peterka, R. J. Sensorimotor integration in human postural control. J. Neurophysiol. 88, 1097–1118 (2002).

90. Petersen, N. et al. Evidence that a transcortical pathway contributes to stretch reflexes in the tibialis anterior muscle in man. J. Physiol. 512, 267–276 (1998).

91. Pinter, I. J., Swigchem, R. Van, Soest, A. J. K. Van and Rozendaal, L. A. The dynamics of postural sway cannot be captured using a one-segmentinverted pendulum model: a PCA on segment rotations during unperturbed stance. J. Neurophysiol. 100, 3197–3208 (2008).

92. Porges, W. and Byrne, E. A. Research methods for measurement of heart rate and respiration. Biol. Psychol. 34, 93–130 (1992).

93. Prieto, T. E. et al. Measures of postural steadiness: differences between healthy young and elderly adults. IEEE 43, 956–966 (1996).

94. Quant, S., Adkin, A. L., Staines, W. R., Maki, B. E. and Mcilroy, W. E. The effect of a concurrent cognitive task on cortical potentials evoked by unpredictable balance perturbations. BMC Neurosci. 5, 1–12 (2004).

95. Rankin, J. K., Woollacott, M. H., Shumway-cook, A. and Brown, L. A. Cognitive influence on postural stability: a neuromuscular analysis in young and older adults. J. Gerontol. 55, 112–119 (2000).

96. Reynolds, R. F. The ability to voluntarily control sway reflects the difficulty of the standing task. Gait Posture 31, 78–81 (2010).

97. Richer, N., Saunders, D., Polskaia, N. and Lajoie, Y. The effects of attentional focus and cognitive tasks on postural sway may be the result of automaticity. Gait Posture 54, 45–49 (2017).

98. Riley, M. A., Baker, A. A. and Schmit, J. M. Inverse relation between postural variability and difficulty of a concurrent short-term memory task. Brain Res. Bull. 62, 191–195 (2003).

99. Riley, M. A. et al. Effects of visual and auditory short-term memory tasks on the spatiotemporal dynamics and variability of postural sway. J. Mot. Behav. 37, 311–324 (2010).

100. Rossi-durand, C. The influence of increased muscle spindle sensitivity on Achilles tendon jerk and H-reflex in relaxed human subjects. Somatosens. Mot. Res. 19, 286–295 (2002).

101. Scholz, J. P., Kang, N., Patterson, D. and Latash, M. L. Uncontrolled manifold analysis of single trials during multi-finger force production by

persons with and without Down syndrome. Exp. Brain Res. 153, 45–58 (2003).

102. Shapiro, J. et al. Importance of body sway velocity information in controlling ankle extensor activities during quiet stance. J. Neurophysiol. 90, 3774–3782 (2013).

103. Shumway-cook, A., Woollacott, M., Kerns, K. A. and Baldwin, M. The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J. Gerontol. 52, 232–240 (1997).

104. Sibley, K. M., Carpenter, M. G., Perry, J. C. and Frank, J. S. Effects of postural anxiety on the soleus H-reflex. Hum. Mov. Sci. 26, 103–112 (2007).

105. Siu, K. and Woollacott, M. H. Attentional demands of postural control: The ability to selectively allocate information-processing resources. Gait Posture 25, 121–126 (2007).

106. Smith, J. M. The forces operating at the human ankle joint during standing. J. Anat. 91, 545–564 (1957).

107. Solopova, I. A., Kazennikov, O. V, Deniskina, N. B., Levik, Y. S. and Ivanenko, Y. P. Postural instability enhances motor responses to transcranial magnetic

stimulation in humans. Neurosci. Lett. 337, 25–28 (2003).

108. Stins, J. F., Roerdink, M. and Beek, P. J. To freeze or not to freeze? Affective and cognitive perturbations have markedly different effects on postural control. Hum. Mov. Sci. 30, 190–202 (2011).

109. Stoffregen, T. A., Pagulayan, R. J., Bardy, B. G. and Hettinger, L. J. Modulating postural control to facilitate visual performance. Hum. Mov. Sci. 19, 203–220 (2000).

110. Tanabe, H., Fujii, K. and Kouzaki, M. Intermittent muscle activity in the feedback loop of postural control system during natural quiet standing. Sci. Rep. 1–21 (2017).

111. Taube, W., Gruber, M. and Gollhofer, A. Spinal and supraspinal adaptations associated with balance training and their functional relevance. Acta Physiol. 193, 101–116 (2008).

112. Teasdale, N. and Simoneau, M. Attentional demands for postural control: the effects of aging and sensory reintegration. Gait Posture 14, 203–210 (2001).

113. Tokuno, C. D., Carpenter, M. G., Thorstensson, A., Garland, S. J. and

Cresswell, A. G. Control of the triceps surae during the postural sway of quiet standing. Acta Physiol. 191, 229–236 (2007).

114. Tokuno, C. D., Garland, S. J., Carpenter, M. G., Thorstensson, A. and Cresswell, A. G. Sway-dependent modulation of the triceps surae H-reflex during standing. J. Appl. Physiol. 104, 1359–1365 (2008).

115. Tokuno C. D., Martin Keller, M., Carpenter, M. G. Márquez, G. and Taube, W. Alterations in the cortical control of standing posture during varying levels of postural threat and task difficulty. Journal of Neurophysiology. 120(3), 1010-1016 (2018).

116. Ueta, K., Okada, Y., Nakano, H., Osumi, M. and Morioka, S. Effects of voluntary and automatic control of center of pressure sway during quiet standing. J. Mot. Behav. 47, 37–41 (2014).

117. Vette, A. H., Masani, K., Nakazawa, K. and Popovic, M. R. Neural-mechanical feedback control scheme fluctuation during quiet stance. IEEE 18, 86–95 (2010).

118. Vuillerme, N., Nougier, V. and Teasdale, N. Effects of a reaction time task on

postural control in humans. Neurosci. Lett. 291, 77–80 (2000).

119. Watanabe, K. and Funahashi, S. Neural mechanisms of dual-task interference and cognitive capacity limitation in the prefrontal cortex. Nat. Neurosci. 17, 601–611 (2014).

120. Watter, S. and Logan, G. D. Parallel response selection in dual-task situations. J. Percept. Psychophys. 68, 254–277 (2006).

121. Weaver, T. B., Janzen, M. R., Adkin, A. L. and Tokuno, C. D. Performance changes in spinal excitability during dual task performance. J. Mot. Behav. ISSN 44, 289–294 (2012).

122. Winter, D. A. Biomechanics and motor control of human movement. (John Wiley and Sons, 2009).

123. Winter, D. A., Patla, A. E. and Rietdyk, S. Ankle muscle stiffness in the control of balance during quiet standing. J. Neurophysiol. 85, 2630–2633 (2001).

124. Winter, D. A., Patla, A. E., Prince, F., Ishac, M. and Gielo-Perczak, K. Stiffness control of balance in quiet standing stiffness control of balance in

quiet standing. J. Neurophysiol. 80, 1211–1221 (1998).

125. Woollacott, M. H. and Shumway-Cook, A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture 16, 1–14 (2002).

126. Zehr, E. P. Considerations for use of the Hoffmann reflex in exercise studies. Eur. J. Appl. Physiol. 86, 455–468 (2002).

127. Zhang, W., Scholz, J. P., Zatsiorsky, V. M. and Latash, M. L. What do synergies do? Effects of secondary constraints on multidigit synergies in accurate force-production tasks. J. Neurophysiol. 99, 500–513 (2010).

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