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Large postural fluctuations but unchanged postural sway dynamics during tiptoe standing compared to quiet standing

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ABSTRACT

The purpose of this study was to detect the characteristics of center of pressure (COP) movement during tiptoe standing (TS) compared to quiet standing (QS). Eight healthy subjects were asked to perform QS and TS on a force platform. During standing, surface electromyograms (EMGs) were recorded from the soleus (SOL), flexor hallucis brevis (FHB), medial gastrocnemius (MG), lateral gastrocnemius (LG), and tibialis anterior (TA) muscles. The path length and rectangular area of the COP trajectory were significantly larger during TS than during QS. In contrast, irrespective of standing condition, the scaling coefficients in the short and long regions were above and below 0.5, respectively. The coherence spectrum between the COP and EMG from the SOL and FHB muscles was statistically significant during TS at frequencies up to 17 Hz, while that for the QS was only significant below 1 Hz. In conclusion, the control of COP movement during TS was similar to that during QS despite large COP fluctuations during TS. Our results suggest that unstable posture during TS is compensated for by the activities of the SOL and FHB muscles, which enhance postural control.

1. Introduction

Tiptoe standing (TS) refers to an upright stance with the heels raised; humans use this stance frequently. For example, we must maintain our posture on tiptoe while walking during the single-support and push-off phases. Moreover, when we want to reach something at a high place in daily living, we must keep stable TS for long durations. Ballet dancers provide another specific example; they often maintain themselves on their tiptoes and stably keep that posture. Although postural maintenance during TS represents a significant component of human movement, surprisingly few studies have examined postural control during TS. Maintaining TS seems to require more complex control dynamics than quiet standing (QS) for three reasons: (1) TS is mechanically more unstable than QS; (2) TS has anatomical disadvantages compared to QS, such as an insufficient contribution of muscles and tendons to maintain postural stability; and (3) the input of somatosensory information is reduced during TS compared with QS.

Posture during TS is mechanically more complex than during QS because the degrees of freedom of the joints are larger and the center of mass (COM) is located higher above a small base of support. In the bipedal posture of humans, the COM is located in front of the ankle joint, so backward torque must be applied around the ankle joint to prevent the body from falling forward. TS makes the inherent unstableness of the human bipedal stance more mechanically unstable because of the increase in the joints' degrees of freedom and the higher COM with a smaller surface of the base of support.

In general, the human bipedal stance is approximated as a single-segment single-joint inverted pendulum that rotates about the ankle joint (Morasso and Schieppati, 1999). Based on the dynamic properties of QS, the plantar flexor muscles play a significant role in stabilizing the body (Masani et al., 2003). During TS, however, because the coordinate center of the ankle joint fluctuates due to heel-raising, the body cannot be approximated as an inverted pendulum around the ankle joint. From the mechanical point of view, it is reasonable to assume that the muscle group around the metatarsophalangeal (MP) joint and its tendon have an important role in controlling posture during TS. During QS, ankle stiffness, which is dominated by the soft, linear Achilles' tendon (Loram et al., 2007), has been suggested to play an important role in generating the required backward torque. During TS, however, the stiffness may be less sufficient to generate the backward torque compared to QS because the muscles (e.g., the flexor hallucis brevis and the flexor hallucis longus), their tendons, and the aponeurosis around the MP joints are smaller than those of the main working muscle groups that function during QS. Thus, there seem to be some differences in the postural dynamics that control QS and TS, and these dynamics must be affected by the coordination of sensory inputs, muscle activity, and the central nervous system (CNS).

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Somatosensory information has been proposed to affect postural control (Peterka, 2002). Because the heels do not touch the ground during TS, the lack of somatosensory information from the sole should cause large amplitude of body fluctuations; that is, because somatosensory information has been considered to affect the higher frequency COP fluctuations, less somatosensory information during TS may change the power spectral density of higher frequencies. Hence, to prevent falling, this deficit due to TS must be compensated for by some muscle activity. Therefore, we investigated body fluctuations and electromyogram (EMG) activity during TS.

The majority of COP fluctuations have been reported to be in the frequency range below 1 Hz (Kouzaki et al., 2007; Kouzaki and Shinohara, 2010). The displacement of the COP results from the coordinated interaction of various sensory components such as the visual, vestibular, proprioceptive and cutaneous systems. To achieve postural stability, sensory systems responding to afferent signals of body fluctuations transmit sensory signals to the postural controller, which exists in the CNS and sends out efferent signals to muscles and tendons as motor commands. In this study, we assessed the relationship between the instability of TS and body fluctuations through calculation of the parameters of body sway derived from the COP time series. We first calculated the total path length of the COP trajectory and the rectangular area that covers the whole COP trajectory. Alternately, Collins and De Luca (1993) proposed the method of stabilogram-diffusion analysis. They regarded COP displacement as analogous to the Brownian movements of a molecule in a fluid and resolved the computed stabilogram-diffusion plot into open-loop and closed-loop mechanisms. Although the interpretation of open and closed mechanisms in this physiological phenomenon is still controversial, this method has been used to detect unstable postural control in the elderly (Collins and De Luca, 1995), Parkinson's disease patients (Mitchell et al., 1995), and during muscle fatigue (Gimmon et al., 2011). In the current study, we applied stabilogram-diffusion analysis to COP sway to infer the postural control mechanism that operates during TS.

Postural feedback control involves integrated information from the vestibular, visual, and somatosensory systems (Dietz et al., 1992), and each sensory input is reflected in specific frequency bands of COP fluctuations; i.e., low frequency (under 1–2 Hz) COP fluctuations are the result of postural control based on visual and vestibular information, and relatively high frequency (approximately 2–10 Hz) COP fluctuations are under postural control based on proprioceptive information. Moreover, our recent study (Kimura et al., 2012) demonstrated that applied noise-like vibration, which enhances cutaneous sensitivity, to the touching surface of the index finger during quiet standing decreases only the high frequency COP fluctuations. Therefore, there is a possibility that the diminished cutaneous sensory input from foot sole has an effect on the COP fluctuations; this possibility was addressed in this study.

We hypothesized that the characteristics of body sway would vary between TS and QS and that there would be greater tuning of muscle activity to COP sway during TS compared with QS. To test these hypotheses, the present study examined (1) the amplitude of body fluctuation and the diffusion and scaling coefficients of stabilogram-diffusion, (2) the frequency features of COP fluctuations, and (3) the contributions of the muscles that participate in erect posture to COP sway investigated with a cross-spectral analysis.

2. Materials and methods

2.1. Subjects

Eight healthy male subjects were included in this study. The subjects' mean age, height, and body mass were (mean ± SD) 25.0 ± 6.2 years, 171.3 ± 5.3 cm, and 66.5 ± 6.3 kg, respectively. The subjects had no significant medical history or signs of gait, postural, or neurological disorders, and no subject had vision problem. Informed consent was obtained from each subject prior to participation. All procedures used in this study were in accordance with the Declaration of Helsinki and were approved by the Ethics Committee of the Graduate School of Human and Environmental Studies at Kyoto University.

2.2. Experimental protocol and measurements

The basic setup and measurements of postural sway during bipedal standing have been described in our previous studies (Kouzaki and Masani, 2012; Kouzaki and Shinohara, 2010). Each subject was instructed to stand quietly on a force platform (EFP-A-1.5kNSA13B, Kyowa, Tokyo, Japan) with their eyes open. They looked at a fixed point on a plain wall in front of them. The subjects stood barefoot with their arms resting comfortably at their sides and their feet together and parallel. TS can be considered a model in which the joints have more degrees of freedom compared to QS; this increase in the degrees of freedom partially leads to the inherent unstableness of TS. Each subject performed the following two kinds of standing for approximately 40 s: (1) standing quietly with their heels on the platform, which we refer to as quiet standing (QS), and (2) tiptoe standing (TS). Ten trials were conducted for each condition, and a sufficient rest period of at least 2 min was allowed between the trials. The order of the trials was randomized between the two conditions.

Surface electromyograms (EMGs) from the skin surface over the soleus (SOL), flexor hallucis brevis (FHB), medial gastrocnemius (MG), lateral gastrocnemius (LG), and tibialis anterior (TA) muscles were recorded with Ag-AgCl electrodes with a diameter of 5 mm and an interelectrode distance of 20 mm. After careful shaving and abrasion of the skin, the electrodes were placed over the belly of the appropriate muscles. The reference electrode for the EMGs was placed over the lateral portion of the knee. The electrodes were connected to a preamplifier and a differential amplifier with a bandwidth of 5–1 kHz (MEG-6116M, Nihon-kohden, Tokyo, Japan). All signals were stored with a sampling frequency of 2 kHz on the hard disk of a personal computer using a 16-bit analog-to-digital converter (PowerLab/16SP, ADInstruments, Sydney, Australia).

2.3. Data analysis

For all recorded signals, the data from a 30-s period during the middle portion of the collection interval (~40 s) were selected for analysis of the individual trials. We first calculated the mean amplitude of the EMG (mEMG) from each muscle and computed the ratio of the mEMGs from the SOL and the TA to investigate whether the co-contraction of these muscles occurred during standing. The center of pressure (COP) in both the anteroposterior and mediolateral directions was calculated by the vertical component of a force platform. To assess the body sway amplitude, the path length and rectangular area of the COP displacement were calculated. Before the calculation of the path length and rectangular area of the COP, a time series of the COP was passed through a 15 Hz Butterworth low-pass filter (Kouzaki and Masani, 2012; Kouzaki et al., 2007). Next, the rectangular area of the trajectory of the COP was calculated.

The path length of the COP trajectory is related to the amount of regulatory balancing activity; i.e., the low frequency component of COP displacement is under the control responding to visual and vestibular information, and the high frequency component is the outcome of the control that reflects somatosensory information. The COP trajectory is the sum of these whole frequency compo-