Neuroscience Letters 450 (2009) 27–31

Contents lists available at ScienceDirect

Neuroscience Letters

journal homepage: www.elsevier.com/locate/neulet

Effect of the hip motion on the body kinematics in the sagittal plane during human quiet standing

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ARTICLE INFO

Article history:
Received 3 September 2008
Received in revised form 1 November 2008
Accepted 12 November 2008

Keywords:
Quiet standing
Postural control
Inverted pendulum model
Ankle joint
Hip joint

ABSTRACT

Human quiet stance is often modeled as a single-link inverted pendulum pivoting only around the ankle joints in the sagittal plane. However, several recent studies have shown that movement around the hip joint cannot be negligible, and the body behaves like a double-link inverted pendulum. The purpose of this study was to examine how the hip motion affects the body kinematics in the sagittal plane during quiet standing. Ten healthy subjects were requested to keep a quiet stance for 30 s on a force platform. The angular displacements of the ankle and hip joints were measured using two highly sensitive CCD laser sensors. By taking the second derivative of the angular displacements, the angular accelerations of both joints were obtained. As for the angular displacements, there was no clear correlation between the ankle and hip joints. On the other hand, the angular accelerations of both joints were found to be modulated in a consistent anti-phase pattern. Then we estimated the anterior–posterior (A–P) acceleration of the center of mass (CoM) as a linear summation of the angular acceleration data. Simultaneously, we derived the actual CoM acceleration by dividing A–P share force by body mass. When we estimated CoM acceleration using only the angular acceleration of the ankle joint under the assumption that movement of the CoM is merely a scaled reflection of the motion of the ankle, it was largely overestimated as compared to the actual CoM acceleration. Whereas, when we take the angular acceleration of the hip joint into the calculation, it showed good coincidence with the actual CoM acceleration. These results indicate that the movement around the hip joint has a substantial effect on the body kinematics in the sagittal plane even during quiet standing.

From a mechanical viewpoint, human bipedal stance is inherently unstable, because a large body mass is located high above a relatively small base of support. Hence, an advanced facility of the postural-control system is required for maintaining upright posture. The generally accepted idea is that humans are able to select distinct strategies depending on task requirements [12]. For small disturbances, the ankle motion alone is believed to be sufficient to maintain balance (ankle strategy) [10]. For large disturbances, the body behaves like a double-link inverted pendulum (DIP), displaying multiple coordination patterns between the hip and ankle joints (hip strategy) [10,19]. Other study has also indicated that in the narrow stance, anterior–posterior (A–P) balance is predominantly under ankle control (plantar/dorsiflexor), whereas medio-lateral (M–L) balance is under hip control (abd/adductors) [23].

On the basis of the background mentioned above, as for the sagittal plane, it is generally assumed that human quiet stance can be approximated as a single-link inverted pendulum (SIP) pivoting only around the ankle joints. Such approximation has been validated experimentally by the evidence that the difference between the center of mass (CoM) and center of pressure (CoP) is proportional to the horizontal acceleration of the body [5,16,24]. In addition, Gage et al. [8] demonstrated that the individual segments and lower limb angles temporally and spatially synchronize with the whole body CoM. Moreover, Gatev et al. [9] reported that only the ankle motion correlated with the CoP in the sagittal plane, and concluded that ankle mechanisms dominate in the balance control during quiet standing.

However, several recent studies have shown that the movement around the hip joint cannot be negligible even in the sagittal plane during quiet standing [4,6,7,25], and the body behaves like a DIP. For example, Aramaki et al. [4] reported a consistent reciprocal relationship between the angular accelerations of the hip and ankle joints during quiet standing. In addition, Creath et al. [6] and Zhang et al. [25] recently examined angular relationship between the leg and trunk segments during quiet standing using frequency domain technique. They demonstrated co-existence of

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doi:10.1016/j.neulet.2008.11.027

in-phase and anti-phase relationships between the leg and trunk angles, i.e., the angular motion of both segments were in-phase below 1 Hz and anti-phase for frequencies above 1 Hz. However, how the movement around the hip joint affects the whole body kinematics is still unknown. The purpose of this study was to examine the effect of the hip motion on the body kinematics in the sagittal plane during quiet standing, by quantifying the complete time-dependent profiles of both joint movements.

Ten healthy active male subjects participated voluntarily in this study. Their age, height, and body mass were 25.2 ± 2.1 years, 170.2 ± 5.1 cm, and 69.7 ± 8.0 kg, respectively. They had no history of neurological disorders. The experimental procedures used in this 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 consent to participate in the study after receiving a detail explanation of the purpose, potential benefits, and risks involved.

All the experiments were performed 14:00–16:00 h. The barefoot subjects were requested to keep a quiet stance for 30 s on a force platform (Type 9281B, Kistler, Switzerland) with eyes open (EO) and closed (EC). The subjects had their arms hanging along the sides of body, and their feet were kept parallel 15 cm apart between centers of the heels. Five trials were conducted for each eye condition, and 1-min break was provided between the trials. The order of application of EO and EC condition was randomized among the subjects. Upper back and shank displacements in the A–P direction were measured by two highly sensitive charge coupled device laser sensors (resolution: 1 μm, LK-2500, Keyence, Japan). The lower and upper sensors were placed at the level of 20 and 75% of subject's height, respectively. All electric signals were stored with a sample frequency of 100 Hz on the hard disk of a personal computer using a 16-bit A/D converter (PowerLab/16SP, ADInstruments, Australia).

All kinetic and kinematic signals were digitally low-pass filtered using a fourth-ordered Butterworth filter with zero phase lag [22]. Cut-off frequencies of 3.0 and 1.5 Hz were chosen for the kinetic and kinematic data, respectively [8]. The leg and trunk segments were assumed to lie on the line connecting the lateral malleolus with the great trochanter and the line connecting the great trochanter with the acromion, respectively. We defined the ankle angle as the angle between the leg segment and earth vertical, and the hip angle as the angle between the extension of the leg segment and the trunk segment (Fig. 1). After adding the thickness of the body to the data from laser sensors, the displacement signals were converted to angular displacement of the ankle (θ_a) and hip (θ_h) joints as follows (Fig. 1):

$$\theta_a = \frac{l_1 - l_{\mathrm{cal}}}{h_1} \tag{1}$$

$$\theta_h = \frac{(l_2 - l_{\mathrm{cal}}) - (l_1 - l_{\mathrm{cal}})h_3/h_1}{h_3 - h_2} \tag{2}$$

It should be noted that Eqs. (1) and (2) give the angles in radians. Based on the fact that the knee joint angle remains approximately stationary during A–P sway motions during quiet standing [2], the knee joints were ignored in the present study. Then the angular displacements were digitally differentiated to obtain the angular velocities ($\dot{\theta}_a$ and $\dot{\theta}_h$) and angular accelerations ($\ddot{\theta}_a$ and $\ddot{\theta}_h$).

The normalized cross-correlation function (CCF) gives the correlation between the two zero referenced signals ($x(t)$ and $y(t)$) for a variety of time shift (τ). The CCF ($R_{xy}(\tau)$) was defined as follows:

$$R_{xy}(\tau) = \frac{\overline{x(t+\tau)y(t)}}{\sqrt{\overline{x^2 y^2}}} \tag{3}$$

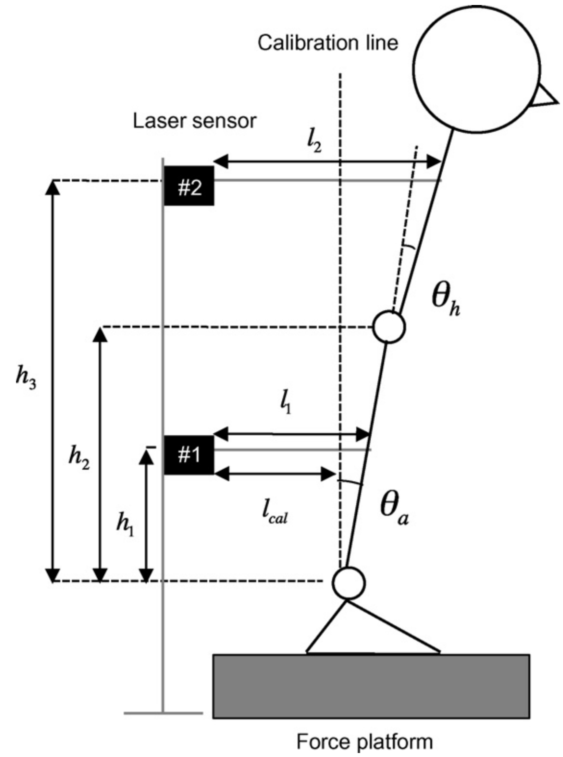

Fig. 1. Diagram of the experimental setup and definition of the angles of the ankle (θ_a) and hip (θ_h) joints. Where h_1, h_2, and h_3 denote the vertical distance from the ankle joints to the laser #1, to the great trochanter, and to the laser #2, respectively. l_1 and l_2 denote the horizontal distance from each sensor to the line connecting the landmarks. Before each subject's trials, the distance between the calibration line and each sensor were measured (l_{cal}).

A normalized CCF of +1 indicates that the two signals are identical, −1 indicates that the two signals are a perfect negative reflection of each other, and 0 indicates that there is no correlation between them.

Data are given as means ± S.D. To test the difference among the root mean squares (RMS) of estimated and actual CoM accelerations, a one-way ANOVA with repeated measures was used. The Tukey–Kramer test was used for post hoc analysis. Fisher's Z-transform was applied to correlation coefficient value to normalize the data for statistical analysis. Difference between EO and EC condition was compared using a paired t-test. $P < 0.05$ was used as a level of significance.

Fig. 2 illustrates typical examples of 10-s time series of the angular displacement (a), the angular velocity (b), and the angular acceleration (c) of the ankle and hip joints during EC condition. To evaluate how coordination between the ankle and hip motions is controlled during quiet standing, CCF for each time series was calculated (Fig. 3). There was no clear correlation between both joints in the angular displacement. Whereas, the negative correlation at zero time shift was observed in the angular velocity. This negative peak at zero time shift became more pronounced in the angular acceleration, indicating a consistent anti-phase relationship between the angular accelerations of the ankle and hip joints.

The A–P position of the CoM (X_{CoM}) relative to the ankle joint can be expressed as a linear summation of θ_a and θ_h [4]:

$$X_{\mathrm{CoM}} = k_1\theta_a + k_2\theta_h \tag{4}$$

where k_1 and k_2 are constants determined from individual anthropometric measurement (body height and segment lengths) and standard anthropometric data (mass distributions and mass centers) [22]. For example, k_1 and k_2 were calculated to be 93 and 25