

Reciprocal Angular Acceleration of the Ankle and Hip Joints during Quiet Standing in Humans

Yu Aramaki,^{1,2*} Daichi Nozaki,^{3*} Kei Masani,⁴
Takeshi Sato,³ Kimitaka Nakazawa,³ and Hideo Yano³

* These authors equally contributed to this study.

Research Institute of National Rehabilitation Center for the Disabled,
Tokorozawa, Saitama 359-8555, Japan

¹ Educational Physiology Laboratory, Graduate School of Education,
The University of Tokyo, 7-3-1 Hongo, Bunkyo-ku, Tokyo 113-0033, Japan;

² Department of Sensory and Communicative Disorders,
Research Institute of National Rehabilitation Center for the Disabled,
4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan;

³ Department of Motor Dysfunction,
Research Institute of National Rehabilitation Center for the Disabled,
4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan;

⁴ Department of Life Sciences, Graduate School of Arts and Sciences,
The University of Tokyo, 3-8-1 Komaba, Meguro-ku, Tokyo 153-8902, Japan.

Correspondence to:
Daichi Nozaki PhD
Department of Motor Dysfunction
Research Institute of NRCD
4-1 Namiki, Tokorozawa
Saitama 359-8555, Japan
Phone: 81-42-995-3100
Fax: 81-42-995-3132
Email: dnozaki@rehab.go.jp

to be appeared in *Experimental Brain Research*

Abstract

Human quiet standing is often modeled as a single inverted pendulum rotating around the ankle joint, under the assumption that movement around the hip joint is quite small. However, several recent studies have shown that movement around hip joint can play a significant role in the efficient maintenance of the center of body mass (COM) above the support area. The aim of this study was to investigate how coordination between the hip and ankle joints is controlled during human quiet standing. Subjects stood quietly for 30 sec with their eyes either opened (EO) or closed (EC), and we measured subtle angular displacements around the ankle (θ_a) and hip (θ_h) joints using three highly sensitive CCD laser displacement sensors. Reliable data were obtained for both angular displacement and angular velocity (first derivative of the angular displacement). Further, measurement error was not predominant even among the angular acceleration data, which were obtained by taking the second derivative of the angular displacement. The angular displacement, velocity and acceleration of the hip were found to be significantly greater ($P < 0.001$) than those of the ankle, confirming that hip joint motion cannot be ignored even during quiet standing. We also found that a consistent reciprocal relationship exists between the angular accelerations of the hip and ankle joints, namely positive or negative angular acceleration of ankle joint is compensated for by oppositely directed angular acceleration of the hip joint. Principal component analysis revealed that this relationship can be expressed as: $\ddot{\theta}_h = \gamma \ddot{\theta}_a$ with $\gamma = -3.15 \pm 1.24$ and $\gamma = -3.12 \pm 1.46$ (mean \pm S.D.) for EO and EC, respectively, where ‘ $\ddot{\theta}$ ’ is the angular acceleration. There was no significant difference in the values of γ for EO and EC, and these values were in agreement with the theoretical value calculated assuming the acceleration of COM was zero. On the other hand, such a consistent relationship was never observed for angular displacement itself. These results suggest that the angular motions around the hip and ankle joints are not to keep the COM at a constant position, but rather to minimize acceleration of the COM.

Key words

Postural control, Center of pressure, Posturography, Degree of freedom, Kinematic constraint

Introduction

From a mechanical viewpoint, the bipedal posture of the standing human is inherently unstable. Because this instability is unique to the human postural system, it has attracted the attention of many researchers interested in the mechanisms responsible for posture control. One approach that has proved useful for investigating human postural control is analysis of the responses of a quietly standing body to perturbations applied to various sub-systems, such as the mechanical (Bloem et al. 2000; Fitzpatrick et al. 1992a, b, 1996; Gurfinkel et al. 1995; Horak and Nashner 1986; Nashner 1976), proprioceptive (Johansson et al. 1988), visual (van Asten et al. 1988; Dijkstra et al. 1994; Peterka and Benolken 1995), and vestibular (Fitzpatrick et al. 1996; Johansson et al. 1995; Pavlik et al. 1999) systems. These perturbation studies have played significant role for clarifying the contribution made by each sub-system to overall postural control (Dietz 1992).

Another approach is to analyze the spontaneous sway that occurs during quiet standing as the trajectory of the center of pressure (COP) (Collins and De Luca 1993, 1994; Goldie et al. 1989), the center of body mass (COM) (Winter et al. 1998), or the ankle joint motion (Fitzpatrick et al. 1994). For example, the contributions of the somatosensory, vestibular and visual systems to the stabilization of the posture have been evaluated by investigating the changes in the magnitude of the angular ankle motion when each system was selectively blocked (Fitzpatrick et al. 1994). It should be noted that in these studies, the sway of the posture was evaluated by a single measure COP, COM or ankle joint motion. In other words, the human body was modeled as a single inverted pendulum, which in many cases can rotate only around the ankle joint. On the other hand, quantitative analysis has shown that hip joint movement may not be negligible (Day et al. 1993), and several more recent studies have pointed out that hip joint motion likely plays a significant role in the efficient maintenance of human standing posture. For example, movement of the head was found not merely to be a scaled reflection of the motion of the hip, and the inconsistency between them, which represents the flexibility of the hip joint, is larger for healthy young subjects than for elderly subjects (Accornero et al. 1997). In addition, when hip joint motion is restricted by a stiff splint during quiet standing in healthy young subjects, the magnitude of the ankle angle sway increases considerably (Fitzpatrick et al. 1994). These studies clearly show that control around the hip joint may be important for stabilizing human

posture and should not be ignored. To our knowledge, however, the complete time-dependent profiles of the movement of hip and ankle joints have not been fully elucidated. The aim of this study was to investigate how coordination between the two joints is controlled during human quiet standing.

The most serious challenge to measuring the angular motion around the hip and ankle joints during quiet standing is almost certainly the level of accuracy that can be attained, as angular displacement around ankle joint is quite subtle below 0.01 rad (0.573 deg) (Fitzpatrick et al. 1992a, 1994). Furthermore, because in the frequency domain almost all of the power of fluctuation is distributed below 1 Hz (Fitzpatrick et al. 1992b, 1994, 1996), a more accurate method than has been conventionally used is needed to reliably record its fast fluctuating component. A high degree of accuracy in the data is also important for measuring the angular velocity and angular acceleration, which are obtained by digital differentiation. In the present study, this data recording problem was overcome by using CCD (Charge Coupled Device) laser displacement sensors with very high accuracy (resolution $\approx 1 \mu m$). We first show that using these sensors, we were able to make sufficiently reliable measurements of angular motion to enable calculation the angular acceleration (i.e., the second derivative of the angular displacement); we then show that there was a consistent pattern between the motions of the hip and ankle joints, and this relationship might be generated by a constraint minimizing acceleration of the COM.

Materials and methods

Subjects

Six healthy male subjects (age 26 – 32 years; height 170 – 180 cm) participated in this study; all gave informed consent which was approved by the ethical committee of our research institute.

Experimental protocol

The barefoot subjects were requested to stand quietly for 30 seconds on a force platform (Kistler 9281B) with their eyes either opened (EO) or closed (EC). The trials of each condition were alternately repeated and ten trials were conducted under each condition. Narrow stance width (≤ 8 cm) increases lateral sway in the frontal plane (Day et al. 1993); therefore, to enable the analysis to be focussed on the motion in the sagittal plane, the intermalleolar distance was set to 10 cm. Knee and head-neck-trunk movements were restricted by three stiff wooden splints strapped to the back of the subjects at the forehead and pelvis, for upper body, and above and below each knee for the legs, for the lower body (Fig. 1). These splints restricted motion to around the hip and ankle joints only. As illustrated in Fig. 1, three CCD laser displacement sensors (Keyence LK-2500; resolution 1 μm ; size $12 \times 9 \times 5$ cm; weight 700 g) were fixed on a stable rack. The support surface of the rack was separated from the force plate, avoiding the subject movement being transmitted to the CCD sensors. The displacement of three locations of the body (l_1 , l_2 , and l_3 in Fig. 1) was measured by these CCD laser sensors, while the position of the COP was simultaneously recorded by the force platform. It was confirmed that the output of three sensors did not correlate with each other when measuring the location of the stable calibration plate (Fig. 1).

Data processing

The signals from the displacement sensors and force platform were passed through A/D converters (sampling frequency=100 Hz) and then digitally low-pass filtered (cut off frequency=10 Hz) using a fourth-ordered Butterworth filter with zero phase lag (Winter 1990). The changes

in position measured with the laser sensors were then converted to angular displacements as

$$\theta_a = (l_1 - l_{1cal})/h_1, \quad (1)$$

$$\theta_h = \{(l_3 - l_{3cal}) - (l_2 - l_{2cal})(h_3 + h_4)/h_2\}/h_4, \quad (2)$$

where θ_a and θ_h respectively represent the ankle and hip joint angles¹ defined in Fig. 1, within which the other parameters are also depicted. It should be noted that Eqs. 1 and 2 give the angles in radian.

In principle, the two joint angles could be estimated using only two sensors, eliminating the need for sensor #1. In that case, the ankle angle can be estimated as

$$\theta_a^* = (l_2 - l_{2cal})/h_2. \quad (3)$$

However, this estimation could misrepresent the relationship between the angular motions of the hip and ankle because θ_h and θ_a^* are not mathematically independent of one another. This is a significant issue that will be discussed in the later sections, where we will show that the independent measurement of both angles is essential when the signal-to-noise ratio of the data is limited.

The angular displacements were digitally differentiated to obtain the angular velocity ($\dot{\theta}_a$ and $\dot{\theta}_h$) and angular acceleration ($\ddot{\theta}_a$ and $\ddot{\theta}_h$) as: $\dot{\theta}(i) = \{\theta(i+1) - \theta(i-1)\}/2\Delta t$ and $\ddot{\theta}(i) = \{\theta(i+1) - 2\theta(i) + \theta(i-1)\}/\Delta t^2$, where Δt is the sampling time (in this study $\Delta t = 0.01$ s) and $\theta(i)$ etc. are the digitized data.

Data analysis

Because the knee joints were immobilized by splints (Fig. 1), measurements of the angle of the ankle joint made with sensor #1 should agree with those from sensor #2. Therefore, the reliability of the data was quantitatively assessed by calculating the correlation coefficient for the respective measurements in each trial. The root mean square value (r.m.s) was then calculated for the angular displacements, velocities and accelerations to evaluate the magnitude

¹The ankle joint we obtained should be called the shank angle relative to vertical axis in the strict sense, because it would contain the foot deformation (Gurfinkel et al. 1994).

of the motion of each joint. To quantify the respective contributions made by ankle and hip motions to the control of body balance, the cross-correlation function between the COP and each joint angle was calculated (Gatev et al. 1999). However, because the COP and the angular displacement around the ankle joint possess almost all the power in the range of frequencies below 1 Hz (Fitzpatrick et al. 1992b, 1994), their cross-correlation function might ignore the relationship when frequencies are relatively high. We therefore adopted the squared coherence spectrum to investigate the relationship between fast movements of both joints in detail:

$$\text{Coh}^2(f) = \frac{|S_{xy}(f)|^2}{S_x(f)S_y(f)}, \quad (4)$$

where $S_x(f)$ and $S_y(f)$ are the power spectral densities of each time series and $S_{xy}(f)$ is the cross power spectral density between them (Bloomfield 2000). Using the coherence spectrum, the relationship of both joint motions can be analyzed in the frequency domain; the power spectra were thus estimated using a fast Fourier transform. The critical value s for the null hypothesis of zero coherence for the significance level α is given by

$$s = 1 - (1 - \alpha)^{\frac{g^2}{1-g^2}}, \quad (5)$$

where g^2 is determined from the spectral window function and the number of data subsets when estimating the power spectrum (Bloomfield 2000).

The time-dependent motions of the ankle and hip joints were expressed as a trajectory on the (θ_a, θ_h) , $(\dot{\theta}_a, \dot{\theta}_h)$, or $(\ddot{\theta}_a, \ddot{\theta}_h)$ planes, and principal component analysis was used to identify the pattern of the trajectory on each plane (Bianchi et al. 1998; Borghese et al. 1996; Kuo et al. 1998). The principal component analysis was first applied to standardized data with a mean of 0 and unit variance, and by calculating the proportion of the variance (%Variance) that the first principal component was able to explain, whether or not a preferential axis existed was investigated. We judged that a preferential axis existed if the %Variance was larger than 0.6 (this value was arbitrarily selected). The principal component analysis was next applied to the original data whose %Variance was larger than 0.6, and the eigenvector $(1, \gamma)$ of the first principal component was calculated (the value of γ represents the slope of the preferential axis on each plane).

Differences between EO and EC for each variable and between the two joints were compared

using unpaired and a paired t-tests, respectively. Values of $P < 0.05$ were considered statistically significant.

Iso-COM relationship between ankle and hip joint angles

In this subsection, the theoretical relationship between the two joint angles are derived when the position of the COM is held constant, enabling it to serve as a reference on the (θ_a, θ_h) plane. The anterior-posterior position of the COM (X_{COM}), relative to the location of the ankle joint, can be expressed as a linear summation of θ_a and θ_h :

$$X_{\text{COM}} = k_1\theta_a + k_2\theta_h, \quad (6)$$

where k_1 and k_2 are constants determined from standard anthropometric data of the human body, including the length and mass of upper and lower body and so on. The constants k_1 and k_2 were estimated to be approximately 102 and 37, respectively, using the anthropometric data (Winter 1990). The straight lines with slope $-k_1/k_2 \approx -2.8$ are the iso-COM lines on the (θ_a, θ_h) plane. They indicate that to keep the COM at a constant position, the hip joint angle should move approximately three times as much as the ankle joint angle. In that case, the trajectory on the (θ_a, θ_h) plane should draw paths along the iso-COM line and the eigenvector parallel, and that line should be extracted by the principal component analysis.

Results

Reliability of data recording

Typical examples of the time course of the angular motions of the joints are illustrated in Fig. 2. The angular motion around the ankle joint is displayed in two ways: one based on the signal from sensor #1 (Eq. 1) and the other on the signal from sensor #2 (Eq. 3). With respect to angular displacement, the two time series appeared quite similar (Fig. 2A), and the correlation coefficient between them was very high (Table 1). Digital differentiation generally increased the noise level, especially at higher frequencies. Nonetheless, the angular velocities were also quite similar (Fig. 2B), and again the correlation coefficient was high (0.914 ± 0.077 ; Table 1). Even after calculating the second derivative, some similarity could be observed (Fig. 2C), and the correlation coefficient was still reasonably high (0.595 ± 0.164 ; Table 1), confirming that we were able to reliably record the data, even for angular acceleration.

The values of the correlation coefficient were very stable among the subjects. Still, in two subjects the correlation coefficients for the angular velocity and angular acceleration were relatively low (subject 1 and 2; Table 1), indicating that the measurements made on these subjects may have been more influenced by random effects (e.g., the fixation of knee joint by splints was not sufficient) than the others. When these two subjects were excluded, the values of the correlation coefficient were 0.997 ± 0.003 , 0.961 ± 0.020 and 0.684 ± 0.092 for the angular displacement, angular velocity and angular acceleration, respectively (Table 1).

Magnitude of the joint motion and the COP displacement

The r.m.s values for each quantity are shown in Fig. 3. The hip joint consistently exhibited significantly ($P < 0.001$) greater angular displacement, velocity and acceleration than the ankle joint. In addition, a significant difference was observed between the EO and EC conditions for the angular velocity (ankle joint, $P < 0.001$; hip joint, $P < 0.05$), and the r.m.s of the COP displacement was larger for EC than for EO ($P < 0.001$).

Relationship between the displacement of COP and the angular motion

Figure 4 shows representative examples of the squared coherence spectra comparing the COP and the angular displacements in two subjects. At frequencies below 1 Hz, the coherence level between the COP and the ankle joint angle was very high, indicating synchronization of their slow fluctuations. On the other hand, at lower frequencies, coherence between the COP and the hip joint angle did not reach the level of statistical significance. At higher frequencies, by contrast, coherence reached significance for both the ankle and hip joints. Similar characteristics in the coherence spectra were also observed in the four other subjects.

Relationship of the angular motion between both joints

If increases (or decreases) in the angular displacements of the hip joint are compensated by decreases (or increases) in those of the ankle joint, the relationship between them should be negatively correlated. Although there was no clear correlation between the angular displacements of the hip and ankle joints (Fig. 2A), we found that the angular velocity of the hip seemed to move anti-phase to that of the ankle (Fig. 2B), and this reciprocal relationship could be seen even more clearly in the angular acceleration (Fig. 2C). The relationship between the angular motions of the hip and ankle joints can also be depicted as trajectories (Fig. 5). The trajectories reflecting the relationship between angular displacements were clearly not negatively correlated and varied considerably from trial to trial (Fig. 5A). On the other hand, compensatory relationships were evident from the trajectories of the angular velocity and angular acceleration (Fig. 5B, C).

Table 2 summarizes the results of the principal component analysis, which quantified the relationship between the joint motions. For the angular displacement plane, the number of EO and EC trials in which %Variance was larger than 0.6 was 46 and 48, respectively, and the direction of the eigenvector (γ) were quite variable (0.19 ± 3.17 and 1.32 ± 2.90 for EO and EC, respectively). This suggests that angular displacements of the ankle and hip lack a consistent relationship, which is consistent with Fig. 5A. For the angular velocity plane, the number of trials in which %Variance was larger than 0.6 was almost the same as for angular displacement

(47 and 45 for EO and EC, respectively), but the value of γ was much more stable (-2.99 ± 0.85 and -3.40 ± 1.37 for EO and EC, respectively). That γ was negative suggests that the angular velocities of the two joints were in anti-phase with each other. For the angular acceleration plane, %Variance was larger than 0.6 in 58 of 60 trials under both EO and EC conditions, indicating the stable presence of a preferential axis. The remaining two trials were with a subject whose angular acceleration data were not reliable (subject 2, see Table 1); nevertheless, even in these trials the value of %Variance for the angular velocity plane was larger than 0.6. Thus, there was no trial in which %Variance was smaller than 0.6 for both the angular velocity and angular acceleration planes. The value of γ on the angular acceleration plane was -3.15 ± 1.24 and -3.12 ± 1.46 for EO and EC, respectively. The distribution of the value of γ was very stable, and there was no trial in which γ was positive (Fig. 6). It can therefore be concluded that the angular acceleration of the ankle and hip joints fluctuate consistently in anti-phase with each other. There was no statistically significant difference in the value of γ between the EO and EC conditions.

Discussion

We should begin by acknowledging the possibility that our results were affected by the unnatural standing posture adopted in this study, where body movements were possible only around the ankle and hip joints. The reason why we dared to use splints is that it is impossible to measure the complete body movement during normal standing. In fact, in our experimental setup where only three laser sensors are available, the movements of more than two segments cannot be recorded. Even if additional laser sensors are available, the movement of upper body might be impossible due to the flexibility of the human spine. This incomplete description of the body movement without splints would consequently spoil the high sensitivity of the laser sensors. On the other hand, in case of splinted standing, the movements of two segmented body can be recorded without any ambiguity.

Although the subjects did not feel any difficulty in stabilizing the posture with splints, it should be recognized that there are several drawbacks to use the splints. Considering the result by Fitzpatrick et al. (1994) that the postural sway increases when the whole body is splinted, it would be possible that the characteristics of the natural quiet standing were partly lost in our experimental procedure. Although further researches will be needed to investigate to what extent our results hold true for the normal standing, we believe that even the analysis of two-segmented body movement can give us deeper insight into the postural control of a multi-linked body than the previous studies in which the human quiet standing is modeled as only one segment body.

Reliability of data

The magnitude of the angular displacement of the ankle joint was in the order of 0.5 deg (Fig. 3). This level of fluctuation corresponds to ≈ 1 cm movements of the great trochanter, assuming the length between ankle joint and great trochanter is 1 m. Furthermore, since most of the power of fluctuation is distributed below 1 Hz (Fitzpatrick et al. 1992a, 1994), the magnitude of the fast fluctuation is even smaller, necessitating the use of very accurate method of measurement to perform a detailed analysis of the fast fluctuation.

The most accurate conventionally-employed method makes use of a highly sensitive force transducer connected to a very weak spring, the stiffness of which was carefully chosen (Fitzpatrick et al. 1992a, b, 1994, 1996; Gurfinkel et al. 1995). The noise level of the force transducer used by Fitzpatrick et al. (1994) was $< 2 \times 10^{-5}$ rad $\approx 1 \times 10^{-3}$ deg. We achieved the same low noise level in our CCD displacement sensors, thereby guaranteeing that the accuracy of the present data is comparable to that in earlier studies.

Because the noise level of the recording device does not directly correspond to its sensitivity, data reliability should be of paramount importance. We therefore tested the reliability of our data by correlating angular displacements of the ankle measured using two different sensors (Table 1). The values of the correlation coefficients for the angular displacements measured with sensors #1 and #2 were quite high (Table 1). Moreover, because digital differentiation generally increases the noise level at higher frequencies, the fact that the data remained highly reliable after the taking the first derivative (Table 1) serves as further evidence of the accuracy and reproducibility of our measurements. Indeed, even after taking the second derivative, the noise did not dominate the resultant angular acceleration data (Table 1). Furthermore, the magnitude of $\ddot{\theta}_a$ agreed well with that of $\ddot{\theta}_a^*$ (Fig. 2C) (i.e., $\ddot{l}_2 \approx h_2 \ddot{l}_1 / h_1$). This result indicates that $\ddot{\theta}_a$ and $\ddot{\theta}_a^*$ are the exact angular acceleration of the ankle joint, because this agreement would be destroyed if the knee movement and/or the skin deformation affect significantly these variables. Excluding the data from subjects 1 and 2, the correlation coefficient for the angular acceleration was ≈ 0.7 . This means that half of the variance ($0.7^2 = 0.49$) can be explained by common fluctuation, which is sufficient to clarify the relationship between the angular accelerations of the two joints, as shown in the principal component analysis.

Angular motion around hip joint

Movement around hip joint is usually assumed to be much smaller than that around the ankle joint. This assumption is part of the rationale for adopting the single inverted pendulum, rotating around the ankle joint, as the model with which to study the mechanics of human posture (Fitzpatrick et al. 1996; Johansson et al. 1995). However, our findings clearly show that

the motion of hip joint is not negligible, but is in fact larger than the motion of the ankle joint: the angular velocity and acceleration of the hip is approximately twice that of the ankle.

Larger angular motion around the hip joint has also been shown in several earlier studies (e.g., Day et al. 1993; Gatev et al. 1999). But based on the cross-correlation function, Gatev et al. (1999) concluded that ankle mechanisms still dominate in standing posture control, because only angular ankle motion correlated with the excursions of the COP in the sagittal plane. The strong correlation between the COP and the angular motion of the ankle might be due to their high degree of coherence at frequencies below 1 Hz (Fig. 4). This is because slow fluctuations have more power than fast ones and can more strongly affect the correlation structure in the time domain. Hence, on this slower time scale, balance control around the ankle joint is important, as Gatev et al. (1999) concluded. This does not necessarily indicate that the contribution of the hip joint can be disregarded, however. In fact, the coherence spectra clearly show that non-zero coherence existed between the COP and the angular displacements of both joints within the range of frequencies > 1 Hz (Fig. 4). It should be noted that in order to clarify this faster correlation structure, it is essential not only to perform the analysis in the frequency domain, but also to use a highly reliable method for measuring subtle, fast fluctuations. In addition, if the postural system uses any automatic compensation mechanisms, such as a stretch reflex system (Dietz 1992), the time scale of the control may be much faster than 1 Hz. Recent study (Bloem et al. 2000) has given some evidences that the hip joint movement rather than ankle joint movement triggers the automatic postural response to perturbation, suggesting that more attention should be paid to the faster hip joint movement.

Pattern of angular motions around ankle and hip joints

If the location of the COM is kept constant, the relationship between the motions of the ankle and hip joints should be represented by a single line with a negative slope on the (θ_a, θ_h) plane, as shown by Eq. 6. In that case, any rotation of the hip joint would be completely compensated for by an oppositely directed rotation of ankle joint. It is well known that this type of balance control is effective when a relatively large perturbation is applied to a support surface (Horak

and Macpherson 1996; Runge et al. 1999). On the other hand, for smaller perturbations, ankle motion alone is believed to be sufficient to maintain body balance (Nashner 1976; Horak and Macpherson 1996). Thus, if spontaneous sway is considered to be continuous reactions to small internal perturbations, one might guess that ankle joint motion is predominant during quiet standing. While in the previous section, we showed that this is not the case and that hip joint motion cannot be ignored, we did not mean that hip joint motion necessarily compensates the angular displacement of the ankle joint as the previous studies using a large perturbation has reported. In fact, we observed that the angular motion of the ankle and hip were never reciprocal. Instead, both angles fluctuated randomly, and the pattern between them was highly variable (Fig. 5A).

By themselves, the respective angular displacements of the ankle and hip suggest that the two joints move independently of one another, but when angular acceleration is analyzed, a hidden constraint is revealed. In the angular acceleration plane, almost all trials exhibited a preferential axis whose slope was negative (Table 2). This is somewhat surprising because the reliability of these data was not especially high, owing to the taking of the second derivative. Moreover, in the angular velocity plane, some trials lacked a preferential axis despite the relatively high data reliability. It thus appears that the fundamental constraint during quiet standing is the reciprocal relationship between the angular acceleration of the ankle and hip and that in some cases, the reciprocal relationship is partially maintained on the angular velocity plane. In a broad sense, such a strategy for maintaining balance can be regarded as a mixed strategy (Horak and Nashner 1986).

There was no significant difference in the values of γ for angular acceleration under EO or EC conditions. This suggests that the contribution of the visual system to the reciprocal relationship of angular acceleration is small. In fact, reciprocal oscillations faster than 1 Hz (Fig. 2C) are too fast for the visual system to influence standing posture. This is because vision stabilizes postural sway primarily below 0.1 Hz (Dichgans and Brandt 1978), and for visual input the postural system acts as a low-pass filter with a cutoff frequency below 1 Hz (van Asten et al. 1988; Peterka and Benolken 1995). Nevertheless, visual information can modify the relationship between the angular displacement and/or the angular velocity of the two joints.

Accornero et al. (1997) reported that head movements became more like a magnified reflection of the hip movements when the eyes were closed, suggesting that body rigidity increases upon closing the eyes. Indeed, one result of such increased body rigidity was observable in our study (Fig. 3): when eyes were closed, the increase in the angular velocity of the ankle joint was more prominent than that of the hip joint. It is of course possible that our analysis was unable to detect changes in the eigenvector direction (if any) when the eyes were closed, because the angular acceleration data contained a considerable error component. Further investigation is needed to clarify the capacity of our analysis to quantify changes in the postural pattern when other sensory conditions are altered, as Kuo et. al (1998) have attempted to do.

False negative relationship

We found that under certain conditions, the procedure used to process the data was capable of generating a reciprocal relationship between the motions of the ankle and the hip that was entirely artifactual. For example, the data depicted in Fig. 7 are essentially the same as those shown in the top row of Fig. 5, though for the former, ankle joint motion was recorded using sensor #2 instead of sensor #1 (i.e., using only sensors #2 and #3). It should be noted that Eq. 2 can be easily rewritten as the subtraction of the ankle angle from the trunk angle with respect to the vertical line. This procedure has been widely used to obtain the relative angle of one segment from an adjacent one (Day et al. 1993; Gatev et al. 1999). Note that in Fig. 7A the relationship between the angular displacements of the ankle and hip look quite similar to those shown in Fig. 5A. When the first derivatives are taken, however, the reciprocal relationship becomes more apparent (Fig. 7B), and finally when the second derivative is taken, the trajectory on the angular acceleration plane exhibits an unnaturally strong negative correlation (Fig. 7C). It must be emphasized that this clear reciprocal relationship is an artifact. Since when using only two sensors, both Eq. 1 and Eq. 2 use the same l_2 term, θ_h in Eq. 2 is not independent of θ_a^* in Eq. 3. This lack of independence between θ_a^* and θ_h allows for the possibility that an erroneously negative correlation will be generated.

This is not a serious problem when the signal-to-noise ratio is high, as was the case with angular

displacement. In fact, the structures of the relationship in Fig. 5A are well maintained in Fig. 7A. However, as the magnitude of the random noise component increases, as in the case of the angular acceleration, the false negative correlation may be more evident. This is easily understood by imagining a simple case in which $\theta_a = \xi_1$ and $\theta_h = \xi_2 - \xi_1$, where ξ_1 and ξ_2 are independent random variables with unit variance. This mimics the situation where both sets of angle measurements are completely dominated by errors from the two sensors (i.e., the signal-to-noise ratio is zero). Under those conditions, the correlation coefficient between θ_a and θ_h can be calculated to be $-1/\sqrt{2}$, though there is no actual deterministic correlation between them. To our knowledge, this point has not been recognized in earlier studies, and it points up the necessity of taking care to avoid such errors when investigating subtle fluctuations. In our experiments, three CCD sensors were used so that measurements of the angular motion of the ankle and hip could be obtained independently. This means that even if the data are degraded by random error, their independence eliminates the possibility of generating false negative correlations. In contrast to the aforementioned case, this is equivalent to a simple case in which $\theta_a = \xi_1$ and $\theta_h = \xi_2 - \xi_3$, where ξ_1 , ξ_2 , and ξ_3 are independent random variables. These conditions dictate that the correlation coefficient between θ_a and θ_h is always zero, as it should be, and are the reason we used three sensors to measure only two angles. That we observed a reciprocal relationship despite the independence of our measurements shows that the relationship is bonafide.

Time scale of the reciprocal relationship of angular acceleration between both joints

In the present study, we set the cutoff frequency of the digital low-pass filter to 8 Hz. One might think that this value is high, considering the mechanical characteristics of the human body such as inertia. One criticism to our results might be that, if the reciprocal angular accelerations between both joints is generated by the high-frequency component compared to the eigen frequency of the human body (≈ 1 Hz), it is not a consequence of the active postural control. To investigate this, we calculated the cross-power spectrum between the angular accelerations of both joints (Fig. 8A). Figure 8A shows that the phase is approximately $-\pi$ in almost all frequency and that there is a large peak in the cross-power spectrum at around 2 Hz. These results indicate that the angular accelerations of both joints oscillate in anti-phase with each

other and that this reciprocal oscillation is mainly generated at around 2 Hz. In fact, the reciprocal relationship of the angular acceleration is preserved even when the cutoff frequency is reduced to 3 Hz (Fig. 8B). The frequency ≈ 2 Hz would not be too high compared with the eigen frequency of the human body.

Figure 8A also indicates that the respiration might not affect directly the reciprocal relationship of the angular acceleration between both joints. This is because the frequency of the respiration is usually approximately 0.3 Hz, while there is no cross-power below 1 Hz (Fig. 8A).

Implication of reciprocal relationship of angular acceleration between both joints

Borghese et al. (1996) used the principal component analysis to show that the angular motions of the ankle, knee and hip joints during locomotion are kinematically constrained: their time-dependent changes drew a circular trajectory on a plane in 3-D space (i.e., ankle, knee and hip joint angle space); specifically, the number of degrees of freedom of the joint angles of the leg was reduced from 3 to 2 during locomotion. Furthermore, Bianchi et al. (1998) demonstrated that this kinematic constraint changes systematically with the locomotion speed. Those investigators speculated that a neuronal control system generates the kinematic constraint by minimizing or maximizing some unknown performance criteria.

The criterion generating kinematic constraint discovered in the present study can be partly deduced as follows. In the METHODS, we calculated the iso-COM line (Eq. 6). Similarly, the acceleration of the COM (\ddot{X}_{COM}) can be readily obtained by differentiating Eq. 6 with time:

$$\ddot{X}_{\text{COM}} = k_1\ddot{\theta}_a + k_2\ddot{\theta}_h. \quad (7)$$

When the \ddot{X}_{COM} is held at zero, the trajectory described by Eq. 7 is a straight line with slope $-k_1/k_2 \approx -2.8$ and an intercept 0 on the $(\ddot{\theta}_a, \ddot{\theta}_h)$ plane. Interestingly, the value of $-k_1/k_2$ is very similar to the value of γ calculated by the principal component analysis on the angular acceleration plane (Table 2). Therefore, human quiet standing is apparently constrained so as to minimize the acceleration of the COM, rather than to keep the position of the COM constant. Kuo (1995) demonstrated, using a mathematical model, that the relationship between both joint

accelerations during standing would be constrained under the condition that the feet remain in place on the support surface. However, since the angular accelerations during quiet standing are too small to disturb that condition, the stronger constraint than his speculation would be needed to explain our result.

The constraint minimizing the acceleration of the COM is not trivial, because the human postural system could have adopted an alternative form of constraint: for example, it could have given priority to keep the COM at a constant position by modulating the angular accelerations of both joints without any reciprocal relationship between them. Some evidence suggests that the position of the COM itself is not strictly controlled. Collins and De Luca (1993, 1994) showed that the trajectory of the COP can be described as a kind of one-dimensional Brownian motion. They also pointed out that the location of the COP is controlled somewhat sloppily on a faster time scale. Gurfinkel et al. (1995) speculated that the human postural control system does not adopt the position of the COM as a reference because the COM never returns to its initial position after slow ramp tilting of the support surface. Our result might partly explain their results: the COM itself appears uncontrolled because the human postural control system gives higher priority to zero acceleration of the COM than to its constant position.

It is known that the central nervous system has to integrate all available sensory input to construct the information about the position of the COM, although the details of that process remain unknown. Such a process would necessarily be very complex, involving an internal model of the body (Wolpert and Kawato 1998). One might imagine that COM acceleration is also calculated by a similarly complex process, but information about the acceleration of the COM can be obtained more easily than about its position. For example, according to Newton's law of motion, the acceleration of the COM is equivalent to the external force exerted on the body. The only possible location that could be acted upon by an external force during standing is the sole of the foot. Therefore, the shear force acting on sole of the foot would be a possible candidate to provide information about anterior-posterior COM acceleration (Morasso and Schieppati 1999), and would thus be an important factor for maintaining COM acceleration at zero. In fact, studies of cats entailing perturbation of a support surface have shown that controlling the force acting on the soles plays a significant role in posture control (Jacobs and Macpherson 1996), and in

humans, anesthetizing the feet increases the amount of sway in the COP (Magnussen et al. 1990) and in the angular motion of the ankle (Fitzpatrick et al. 1994). The significance of the receptors in the sole of the foot for stabilizing the human stance has also been clearly demonstrated by using vibration as a stimulus (Kavounoudias et al. 1998), by electrically stimulating the sural nerve (Aniss et al. 1992), and by evaluating the effect of foot pressure on the postural reflex (Wu and Chiang 1997). The neuronal origin(s) of kinematic constraint cannot be discerned from the present study. Further research focusing on the time-dependent changes in kinetic data, or the patterns of muscle activation obtained by electromyography, would be needed for a detailed analysis of the strategy used to control a multi-segment body. Since the analysis of such kinetic data as the torque acting on the joints requires determination of the second derivative of the angular displacement, our method using CCD laser sensors would be helpful to these future studies.

Acknowledgment

This work was supported by the Japan Space Forum and the Japan Science and Technology Agency. Address for reprint requests: D. Nozaki, Research Institute of NRCD, 4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan.

References

- Accornero N, Capozza M, Rinalduzzi S, Manfredi GW (1997) Clinical multisegmental posturography: age-related changes in stance control. *Electroencephalogr Clin Neurophysiol* 105:213-219
- Aniss AM, Gandevia SC, Burke D (1992) Reflex responses in active muscles elicited by stimulation of low-threshold afferents from the human foot. *J Neurophysiol* 67:1375-1384
- Asten WN van, Gielen CC, Gon JJ van der (1988) Postural movements induced by rotations of visual scenes. *J Opt Soc Am [A]* 5: 1781-1789.
- Bianchi L, Angelini D, Orani GP, Lacquaniti F (1998) Kinematic coordination in human gait: Relation to mechanical energy cost. *J Neurophysiol* 79:2155-2170
- Bloem BR, Allum JHJ, Carpenter MG, Honegger F (2000) Is lower leg proprioception essential for triggering human automatic postural responses? *Exp Brain Res* 130:375-391
- Bloomfield P (2000) *Fourier Analysis of Time Series: An Introduction*. 2nd Ed John Wiley & Sons, New York
- Borghese NA, Bianchi L, Lacquaniti F (1996) Kinematic determinants of human locomotion. *J Physiol* 494:863-879
- Collins JJ, De Luca CJ (1993) Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories. *Exp Brain Res* 95:308-318
- Collins JJ, De Luca CJ (1994) Random walking during quiet standing. *Phys Rev Lett* 73:764-767
- Day BL, Steiger MJ, Thompson PD, Marsden CD (1993) Effect of vision and stance width on human body motion when standing: implications for afferent control of lateral sway. *J Physiol* 469:479-499
- Dichgans J, Brandt T (1978) Visual-vestibular interaction: effects on self-motion perception and postural control. In: Held R, Leibowitz H, Trauber H (eds) *Handbook of Sensory Physiology*, vol VIII. Springer, Berlin, pp 756-804.
- Dietz V (1992) Human neuronal control of automatic functional movements: interaction between central programs and afferent input. *Physiol Rev* 72:33-69
- Fitzpatrick RC, Taylor JL, McCloskey DI (1992a) Ankle stiffness of standing humans in response to imperceptible perturbation: reflex and task-dependent components. *J Physiol* 454:533-547

- Fitzpatrick RC, Gorman RB, Burke D, Gandevia SC (1992b) Postural proprioceptive reflexes in standing human subjects: bandwidth of response and transmission characteristics. *J Physiol* 458:69-83
- Fitzpatrick RC, Rogers DK, McCloskey DI (1994) Stable human standing with lower-limb muscle afferents providing the only sensory input. *J Physiol* 480:395-403
- Fitzpatrick RC, Burke D, Gandevia SC (1996) Loop gain of reflexes controlling human standing measured with the use of postural and vestibular disturbances. *J Neurophysiol* 76:3994-4008
- Gatev P, Thomas S, Kepple T, Halett M (1999) Feedforward ankle strategy of balance during quiet stance in adults. *J Physiol* 514:915-928
- Goldie PA, Bach TM, Evans OM (1989) Force platform measures for evaluating postural control: reliability and validity. *Arch Phys Med Rehab* 70: 510-517
- Gurfinkel VS, Ivanenko YP, Levik YS (1994) The contribution of foot deformation to the changes of muscular length and angle in the ankle joint during standing in man. *Physiol Res* 43:371-377
- Gurfinkel VS, Ivanenko YP, Levik YS, Babakova IA (1995) Kinesthetic reference for human orthograde posture. *Neuroscience* 68:229-243
- Horak FB, Nashner LM (1986) Central programming of postural movements: adaptation to altered support-surface configurations. *J Neurophysiol* 55:1369-1381
- Horak FB, Macpherson JM (1996) Postural orientation and equilibrium. In: *Handbook of Physiology. Exercise: Regulation and Integration of Multiple Systems*. Am Physiol Soc, pp 255-292
- Jacobs R, Macpherson JM (1996) Two functional muscle grouping during postural equilibrium tasks in standing cats. *J Neurophysiol* 76:2402-2411
- Johansson R, Magnusson M, Åkesson M (1988) Identification of human postural dynamics. *IEEE Trans Biomed Eng* 35:858-869
- Johansson R, Magnusson M, Fransson PA (1995) Galvanic vestibular stimulation for analysis of postural adaptation and stability. *IEEE Trans Biomed Eng* 42:282-292
- Kavounoudias A, Roll R, Roll J-P (1998) The plantar sole is a 'dynamometric map' for human balance control. *Neuroreport* 9:3247-3252

- Kuo AD (1995) An optimal control model for analyzing human postural balance. *IEEE Trans Biomed Eng* 42:87-101
- Kuo AD, Speers RA, Peterka RJ, Horak FB (1998) Effect of altered sensory conditions on multivariate descriptors of human postural sway. *Exp Brain Res* 122:185-195
- Morasso PG, Schieppati M (1999) Can muscle stiffness alone stabilize upright standing? *J Neurophysiol* 83:1622-1626
- Nashner LM (1976) Adapting reflexes controlling the human posture. *Exp Brain Res* 26:59-72
- Pavlik AE, Inglis JT, Lauk M, Oddson L, Collins JJ (1999) The effects of stochastic galvanic vestibular stimulation on human postural sway. *Exp Brain Res* 124:273-280
- Peterka RJ, Benolken MS (1995) Role of somatosensory and vestibular cues in attenuating visually induced human postural sway. *Exp Brain Res* 105:101-110
- Runge CF, Shupert CL, Horak FB, Zajac FE (1999) Ankle and hip postural strategies defined by joint torques. *Gait Posture* 10:161-170
- Winter DA (1990) *Biomechanics and Motor Control of Human Movement*. Wiley, New York
- Winter DA, Patla AE, Prince F, Ishac M, Gielo-Periczak K (1998) Stiffness control of balance in quiet standing. *J Neurophysiol* 80:1211-1221
- Wolpert DM, Miall RC, Kawato M (1998) Internal models of the cerebellum. *Trends Cog Sci* 2:338-347
- Wu G, Chiang J-H (1997) The significance of somatosensory stimulations to the human foot in the control of postural reflexes. *Exp Brain Res* 114:163-169

	θ_a and θ_a^*	$\dot{\theta}_a$ and $\dot{\theta}_a^*$	$\ddot{\theta}_a$ and $\ddot{\theta}_a^*$
Subject 1	0.989 ± 0.009	0.819 ± 0.058	0.454 ± 0.082
Subject 2	0.992 ± 0.010	0.821 ± 0.064	0.381 ± 0.155
Subject 3	0.996 ± 0.002	0.941 ± 0.023	0.684 ± 0.053
Subject 4	0.997 ± 0.003	0.970 ± 0.005	0.803 ± 0.038
Subject 5	0.998 ± 0.003	0.970 ± 0.016	0.649 ± 0.066
Subject 6	0.996 ± 0.004	0.963 ± 0.064	0.599 ± 0.056
Mean \pm S.D.	0.995 ± 0.007	0.914 ± 0.077	0.595 ± 0.164
except sub. 1 and 2	0.997 ± 0.003	0.961 ± 0.020	0.684 ± 0.092

Table 1: Correlation coefficient between the data from sensor #1 and #2. Data are mean \pm S.D.

	%Variance			γ		
	θ	$\dot{\theta}$	$\ddot{\theta}$	θ	$\dot{\theta}$	$\ddot{\theta}$
EO	46	47	58	0.19 ± 3.17	-2.99 ± 0.85	-3.15 ± 1.24
EC	48	45	58	1.32 ± 2.90	-3.40 ± 1.37	-3.12 ± 1.46

Table 2: Summary of the results of the principal component analysis. The value in %Variance indicate the number of trials whose %Variance is larger than 0.6 (the number of trials in each category is 60). The value of γ is mean \pm S.D.

Figure legends

Figure 1. Schematic diagram of the experimental setup and definition of the angles of the ankle and hip joints. Three light wooden splints were used to restrict the motion of the subject to only around the ankle and hip joints. Laser beams from the CCD sensors were directed at the splint of the upper body (sensor #3) and at that of right leg (sensors #1 and #2) and the distance between each splint and sensor was measured. Using a tape measure, body thickness was measured as the distance between the splint and the central line of the body. Adding the thickness of the body to the data from sensors yielded l_1 , l_2 , and l_3 . Note that because we did not analyze the absolute values, but rather their fluctuation, the comparative imprecision of the measurement of body thickness had no influence on our results. For subsequent data processing, the length between ankle joint and each sensor (h_1 , h_2 , h_4) and the great trochanter (h_3) were recorded. Typically, h_1 , h_2 , and $h_3 + h_4$ were approximately 35, 60 and 135 cm, respectively. After each subject's trials, the distance between the calibration plate, which was placed vertically at the ankle joint (calibration line), and each sensor were measured (these values are l_{1cal} , l_{2cal} , and l_{3cal}). The depiction of the inclination of the body is exaggerated to more clearly show the definition of angles.

Figure 2. Representative examples of time series 3 seconds in duration. A, angular displacement; B, angular velocity; C, angular acceleration. Upward deflections represent forward movements. Horizontal lines in B and C represent zero. Ankle joint movements are displayed in two ways: Ankle 1 and Ankle 2 were calculated from Eq. 1 and Eq. 3, respectively. Angular acceleration of the ankle joint (solid line) is superimposed on that of the hip joint (dotted line) to more clearly show their reciprocal behavior (C, Ankle1).

Figure 3. Mean values of the root mean squares (r.m.s) of the angular displacement, angular velocity and angular acceleration averaged across all subjects. Error bars represent 1 S.D. ***, #, ### Note the statistically significant differences between both joints ($P < 0.001$) and between the EO and EC conditions (# $P < 0.05$; ### $P < 0.001$), respectively.

Figure 4. Representative examples of squared coherence spectra between the COP trajectory and the ankle (dotted line) and hip (solid line) angles calculated from two subjects. The spectra were the ensemble averages from 10 trials for each condition. The horizontal broken line denotes the estimated upper 95 % confidence limit for the zero squared coherence (Eq. 5).

Figure 5. Four representative relationships between the angles (A), the angular velocities (B), and the angular accelerations (C) of the ankle and hip joints of one subject (EC condition). The oblique lines in A are the iso-COM lines calculated in METHODS section (Eq. 6). Trajectories parallel to those lines are indicative of the COM being maintained at a constant position. In this case, the trajectories on the angle plane were not along those lines and varied considerably from trial to trial (A). On the other hand, reciprocal relationships were consistently observed on the angular velocity and angular acceleration planes (B, C).

Figure 6. Histograms showing the value of γ for the angular acceleration under EO and EC conditions. The distribution concentrates around at -3 under both conditions. The values of subjects 1 and 2 were relatively variable. For these subjects, the values of the correlation coefficients between the angular acceleration from sensors #1 and #2 were low (Table 1).

Figure 7. This figure depicts essentially the same data as the top row of Fig. 5, except θ_a^* was used instead of θ_a . The difference between Fig. 7 and Fig. 5 is not discernible from the relationship on the angle plane (A). The angular acceleration relationship, however, exhibits a stronger negative correlation (C), which is an artifact generated by the data processing procedure used in this figure (see text for the detail).

Figure 8. Representative examples of the cross power spectral density (solid line) and phase (dotted line) between the angular accelerations of both joints (A). The phase is consistently $\approx \pi$ and there is a large peak at around 2 Hz in the cross-power spectrum, suggesting the reciprocal angular accelerations of both joints might be mainly generated at this frequency range. The effect of the cutoff frequency on the reciprocal relationship of the angular acceleration between both

joints (B). Although the magnitude of the accelerations is reduced, the reciprocal relationship is preserved well even when the cutoff frequency is set to 3 Hz.

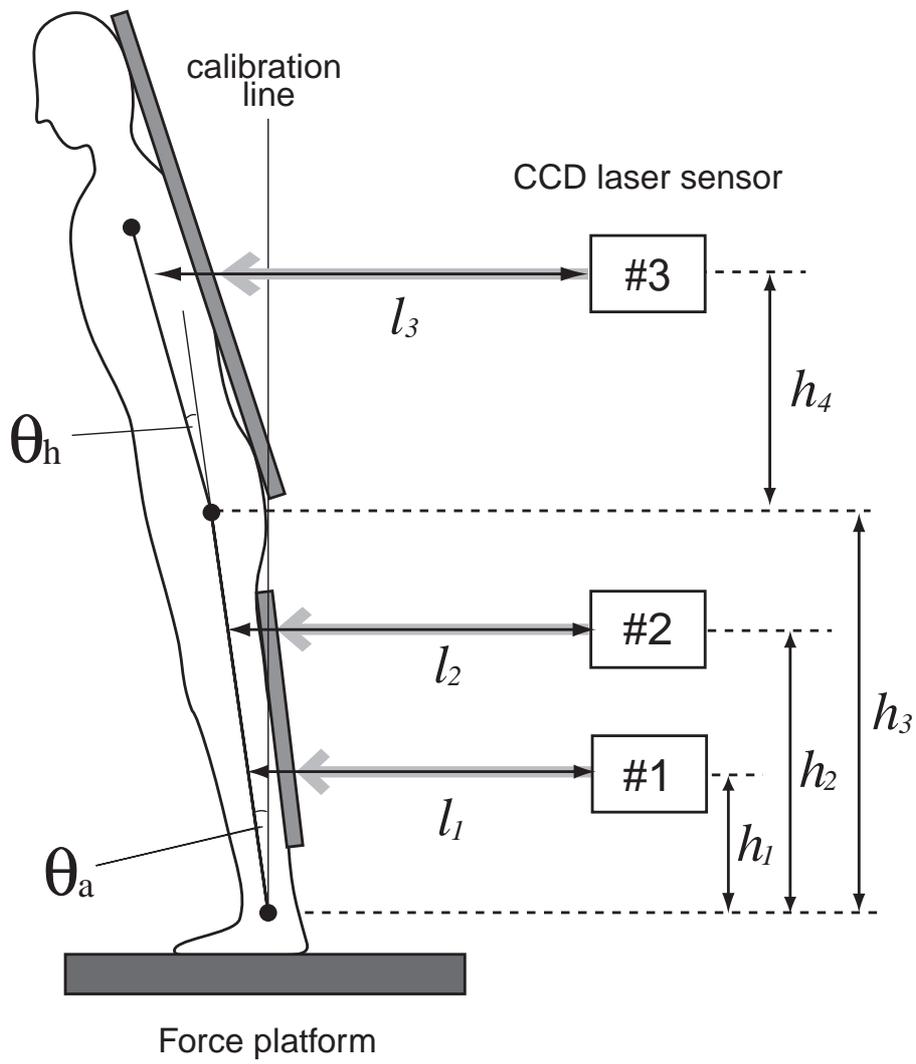


Figure 1:

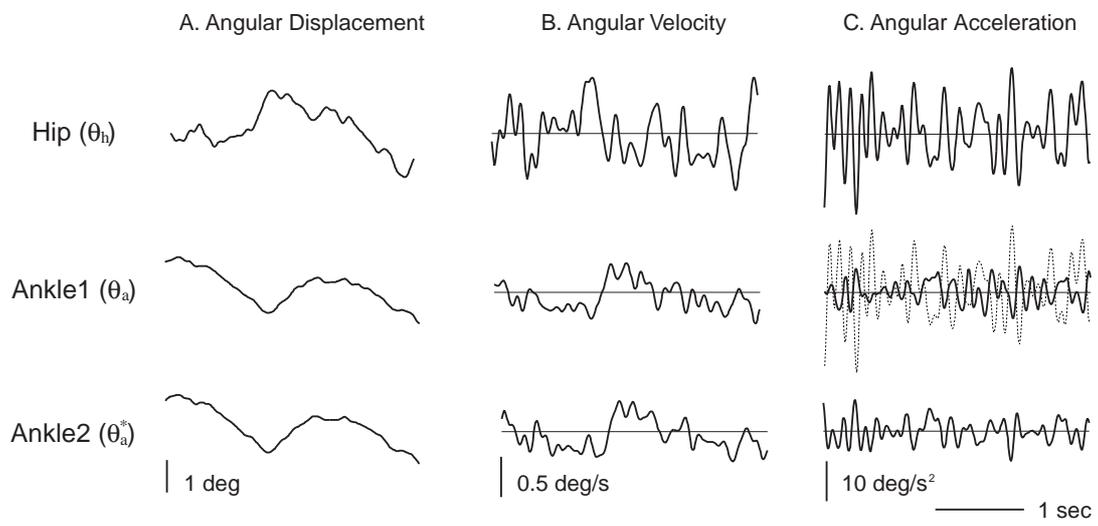


Figure 2:

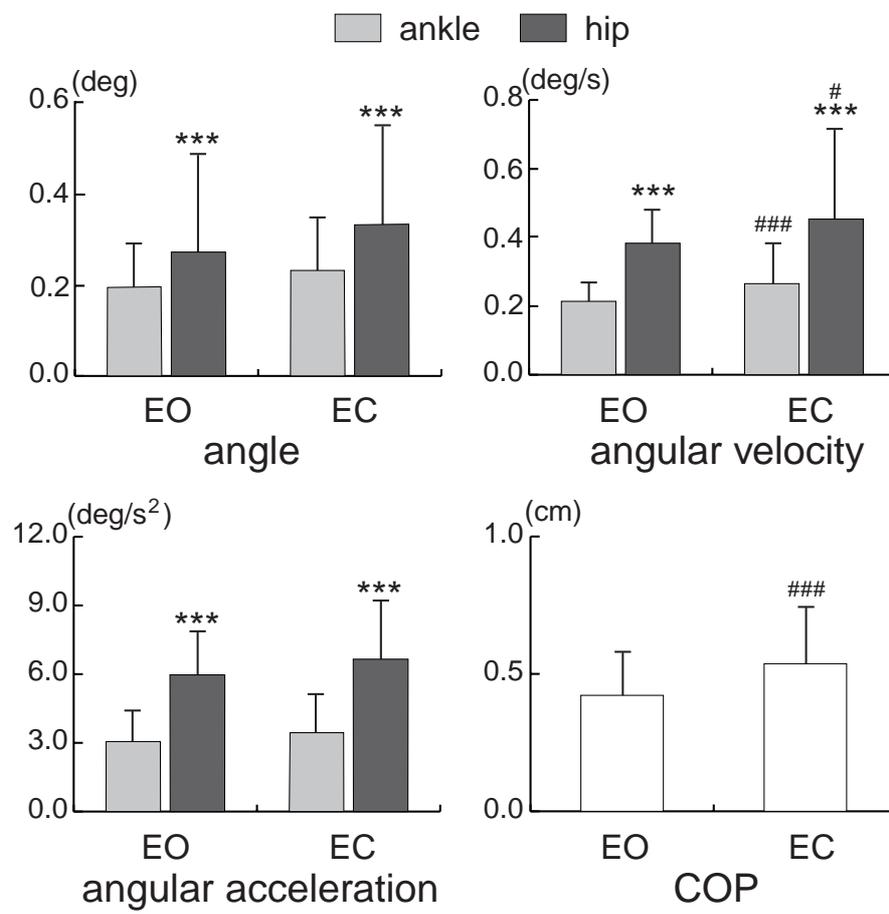


Figure 3:

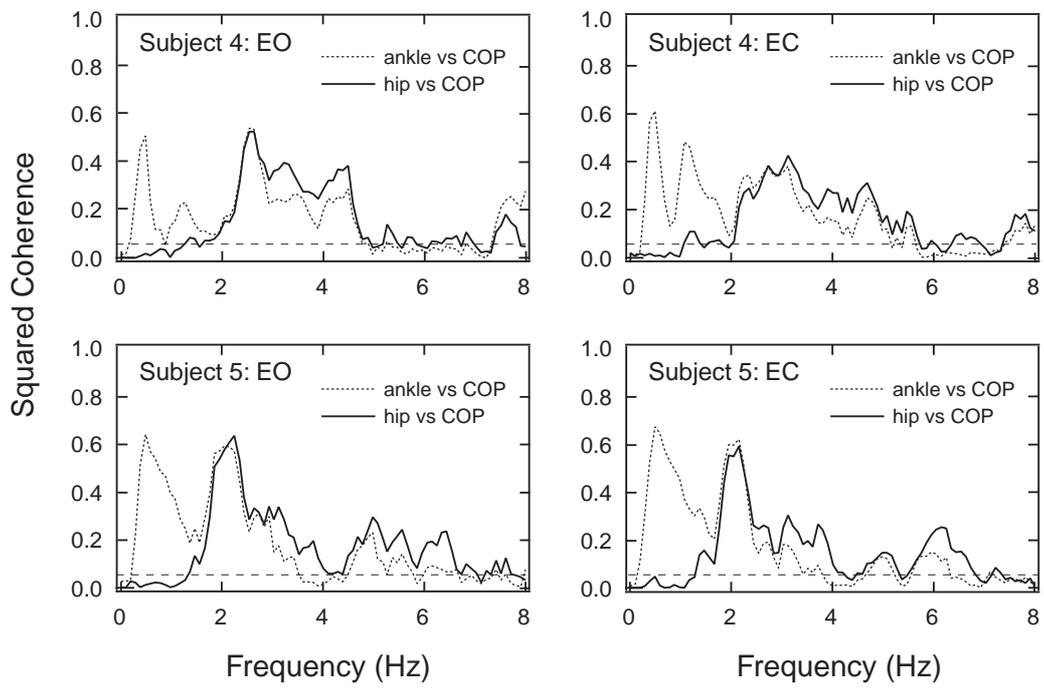


Figure 4:

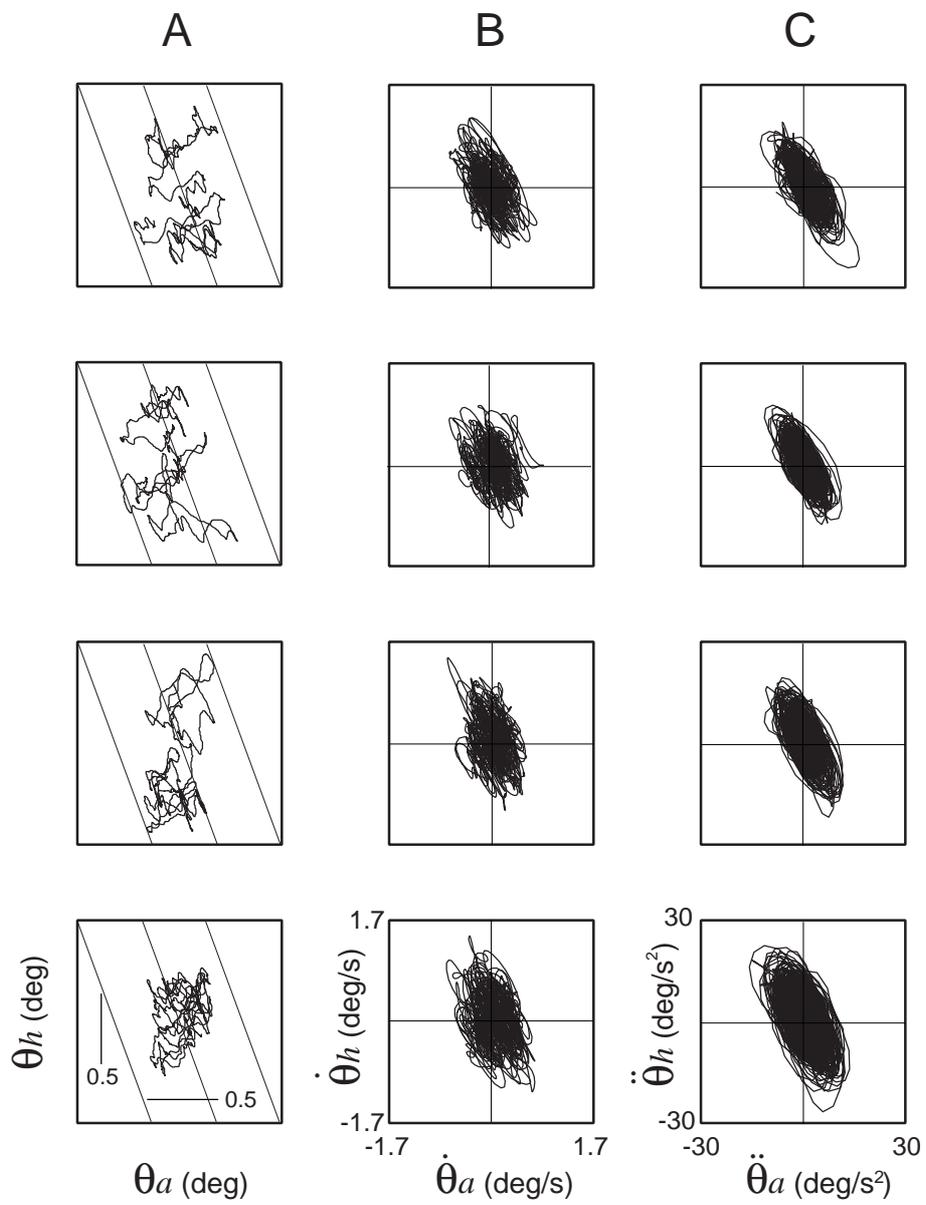


Figure 5:

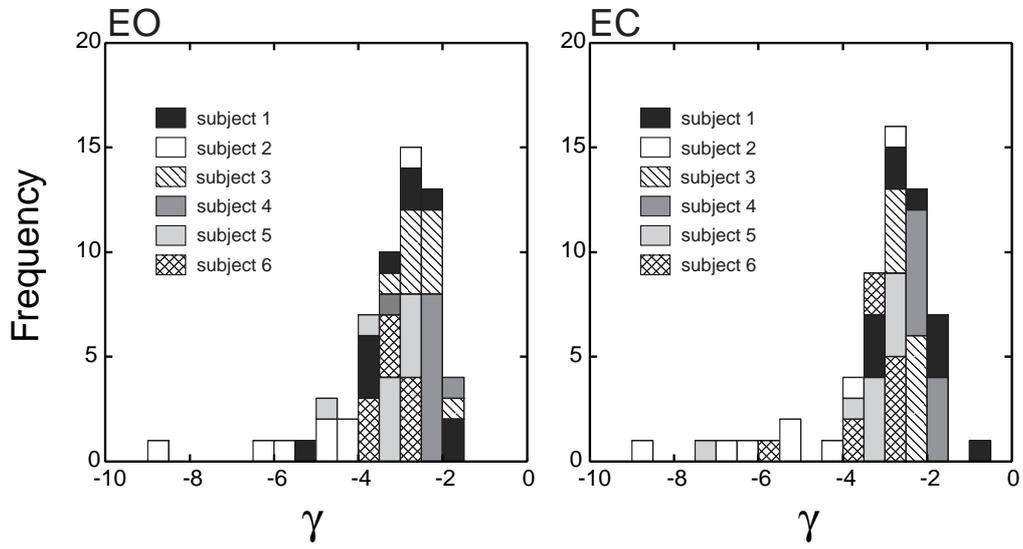


Figure 6:

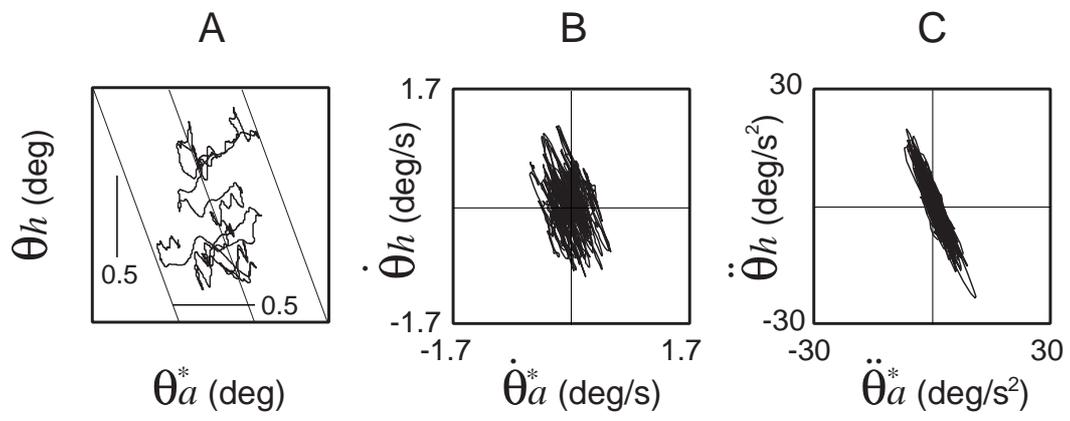


Figure 7:

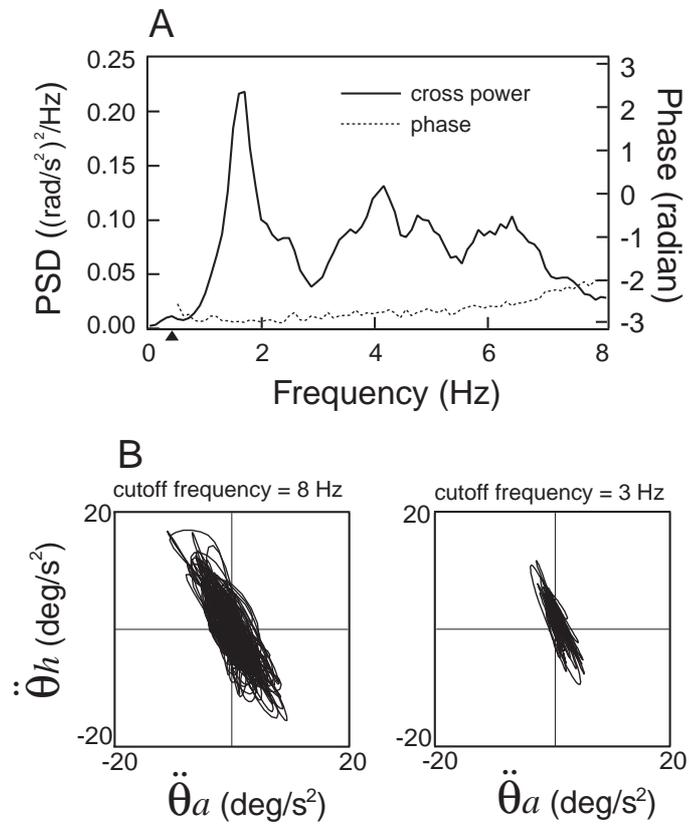


Figure 8: