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Periods of extreme ankle displacement during one-legged standing

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Abstract

The goal of this study was to describe the movement patterns of the ankle joint whilst standing on one leg. Ten healthy adult females (age 24 ± 5.3 years) performed a one-legged standing task with eyes closed. Force platform recordings and video analyses were used to describe the kinematic and kinetic characteristics of the ankle joint during this task. A rocking movement of the foot (heel inversion–eversion) was documented by examining instances of extreme ankle displacement. Extreme ankle displacement was defined as any instant when the ankle position was more than ± 2 SD away from the mean ankle joint position. Extreme values of lateral and medial ankle joint displacement were 14.8 and 9.2 mm, correspondingly. These instances of extreme foot inversion–eversion were characterized by large medial–lateral displacement of the gravity line (GLP) and center of pressure (COP) and large horizontal forces. Comparing instances of extreme ankle joint displacement to periods of non-extreme ankle displacement, the ankle joint moment remained fairly constant, averaging 8.4 ± 4 and 6.9 ± 3.5 Nm, respectively. The moment about the 'body-minus-foot' center of mass generated by the ankle joint reaction force, however, was on average over four times larger during instances of extreme ankle displacement (3.4 ± 2.8 Nm), than during periods of non-extreme ankle displacement (0.8 ± 0.4 Nm). In utmost situations, the moment due to the joint reaction force was up to 73% of the ankle joint moment. These results suggest that at least two different techniques are used to maintain balance during one-legged standing. The first technique, termed the *ankle torque technique*, involves a large restorative moment at a stationary ankle joint for balance maintenance. The other technique, the *shear force technique*, involves a large horizontal force at a moving ankle joint for balance maintenance. During non-extreme periods, balance was maintained primarily through the ankle torque technique. During extreme instances, a combination of both techniques was observed.

Keywords: Standing; Balance; Ankle movement; One-legged standing

1. Introduction

One-legged standing is a popular test of balance. Postural tests incorporating one-legged standing predominantly involve clinical evaluations of the elderly and ankle injuries [1–17]. One-legged standing has also been used to study topics such as: developmental disabilities [18,19], recovery from hip fracture [20], fear of falling [21,22], mental retardation [23], anterior cruciate ligament injury [24,25], cardiorespiratory system [26], and sensory interaction [27]. However, limited research has been conducted on the mechanics or balance techniques used in one-legged standing.

Davis and Grabiner [28] developed an inverted pendulum model of one-legged standing to model the effects of fatigue on postural control during one-legged standing. An assumption critical to the development of this model was that balance during one-legged standing is maintained by a restorative torque at the ankle generated by the ankle muscles. Davis and Grabiner [28] were able to successfully implement their model to reproduce expected COP ranges, root mean square values, and frequencies in the anterior–posterior and medial–lateral directions during eyes open one-legged standing of healthy adults.

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In a study of 15 healthy subjects and 15 subjects with functional ankle instability (defined as a feeling of giving away or recurring ankle sprains), Tropp and Odenrick [29] recognized two movement strategies for maintaining balance during one-legged standing. During periods of equilibrium, Tropp and Odenrick [29] found that balance was maintained by a large torque generated at the ankle joint, an ankle torque technique; and during periods of disequilibrium, balance was maintained through movements at the hip joint creating large restoring horizontal forces. A notable finding from Tropp and Odenrick's work was the existence of ankle inversion–eversion as a critical component to balance maintenance during one-legged standing [29].

Hoogvliet et al. [30] further examined the relationship between medial–lateral ankle displacement and COP migration during one-legged standing of healthy adults. These investigators modeled one-legged standing on a rocker bottom shaped foot. They concluded that medial–lateral motions of the COP are good indicators of tilting motions of the foot observed during one-legged standing, and that foot motion is an important mechanism of postural control.

Based on the aforementioned data, it appears that motion of the foot for certain tasks and populations is an important factor in maintaining balance on one leg. Thus, it is possible that one-legged standing may belong to a specialized subgroup of postures characterized by a moving distal segment. Such a movement pattern could identify a new class of balance techniques that have only begun to be studied for human movement [30]. In this type of situation, balance is maintained through a relatively large displacement of the COP made possible by a rocking motion of the foot creating a combination of ankle torque and horizontal forces. The essential premise of this paper is that during different instances of one-legged standing, the foot can be classified as stationary or rocking. When the foot is stationary, balance is maintained by developing a large restorative torque at the ankle joint. However, during periods of foot rocking, balance is maintained through both a restorative ankle torque and a large shear force.

The goal of this study was to further investigate the role of foot movement in maintaining balance during one-legged standing by examining five variables simultaneously: the rocking motion of the foot, ankle joint torque, shear force, center of pressure displacement, and gravity line displacement. A particular interest was to examine instances of extreme disequilibrium or extreme medial–lateral ankle displacement, since the ability of the body to return to equilibrium from these extreme positions is an important feature of balance control. Knowledge obtained from this study could lead to the development of strategies to improve balance control in people with impaired balancing ability.

2. Methods

2.1. Subjects

Ten healthy adult females (age 24 ± 5.3 years; height 1.61 ± 0.05 m; body mass 54.9 ± 5.7 kg) volunteered to participate in this study. Exclusion criteria for subjects consisted of a history of lower limb injury (or current lower limb injury) and/or participation in balance-oriented activities such as gymnastics and dance. Since there is some evidence that gender is a determinant of one-legged standing ability amongst healthy adults, this study was delimited to female subjects [31]. After a brief description of the study, subjects gave their written informed consent, in concordance with the policies of The Pennsylvania State University Human Subjects Office of Regulatory Compliance. All subjects wore tight fitting black shorts and a dark T-shirt, and were tested in bare-feet.

2.2. Protocol

The eyes closed one-legged standing task (ECOL) utilized in this study was part of a larger study incorporating eight different standing conditions, two of which involved bipedal standing. The eight conditions were randomly assigned to the subjects and rest periods were provided between each condition. The ECOL task required the subjects to stand with the longitudinal axis of their support foot aligned with the anterior posterior axis of their trunk. Since leg dominance does not appear to affect one-legged balancing ability [1,2,32,24,33], the subjects were instructed to choose their preferred leg for standing. The non-support leg was held at ankle height above the ground, just adjacent to the support leg. The subject was instructed to maintain a vertical head posture as if they were gazing at a mark placed eye level 1.5 m in front of the subject. The subject was to maintain this starting posture to the best of her ability throughout a 45 s trial.

2.3. Equipment

Two cameras operating at 60 fields/s (model AG450, Panasonic Industrial Company, Matsushita Electric Corporation of America, NJ, USA) were used to obtain a three-dimensional analysis of the foot and lower leg. The field of view for each camera was approximately 45 cm wide by 35 cm high, which was just large enough to capture the movement of the lower leg and foot segments.

Seven retro-reflective markers were placed on the foot and lower leg of the subject (Fig. 1). The locations of the markers relative to the medial and lateral malleoli were recorded for later use in data analysis. Photographic quartz lights were aligned along the optical axes of each camera to illuminate retro-reflective markers placed on the subject. A light emitting diode connected to a synchronization unit was placed near the edge of the force platform in the field of view of the cameras. Two markers positioned at exact locations on the force platform created an origin and y-axis for the video coordinate system. This coordinate system was later transformed to the force platform coordinate system.

A calibration frame with 12 precisely known points was used to calibrate a space $40 \times 30 \times 30$ cm in dimension for the foot and ankle video analysis. The direct linear transformation (DLT), a method of calculating three-dimensional spatial coordinates from video data obtained from two or more arbitrarily placed cameras, was used to calculate the 3D coordinates of the foot and ankle markers [34]. The calculated error of the volume of the calibration frame from the DLT was less than 0.5% of the total volume, and the mean square error (MSE) along each dimension was 0.001 m or less. The accuracy of the camera set-up was 1 mm. By comparison, medial–lateral ankle joint displacement for eyes closed one-legged standing is on average between 9 and 10 mm. The signal-to-noise ratio for eyes closed one-legged standing with this camera set-up was determined to be 12:1, which was considered sufficient to perform a kinematic analysis of the leg and foot using this passive marker system.

A Bertec force platform and amplifier (force plate model 4060, amplifier model AND64, Bertec Corporation, OH, USA) were used to measure the ground reaction forces and moments. Signals from the force plate were collected with a 12 bit A/D board (model AT-NUO-64E3, National Instruments, TX, USA) controlled by software written in the LabVIEW graphical programming language (version 3.11, National Instruments, TX, USA). The sampling frequency of the force platform data was 60 samples/s. Data collection was triggered with the LED synchronization unit and data were collected for 45 s.

Fig. 1. Photo illustration of the marker set-up utilized for the analysis of the foot system.

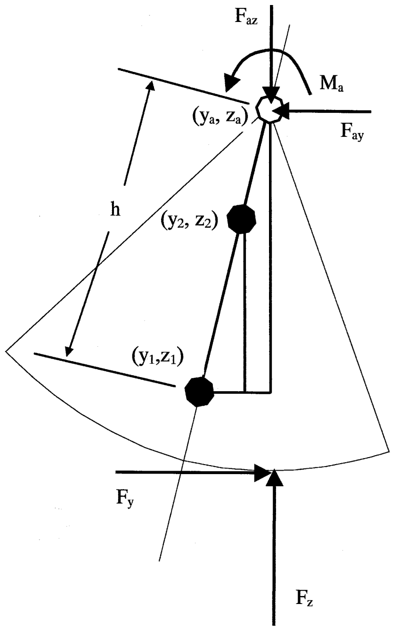
2.4. Data processing

The first 10 s of standing were separated from the beginning of each trial, and the following 30 s of data were used for analysis based on results from Carroll and Freedman [35] and King [36] concerning the stationarity of stabilographic time series. Stationarity of the data was a necessary assumption for cross correlation and power spectral analyses performed as part of the study. However, trials of 45 s in duration are approximately 15 s longer than the reported average time for eyes closed one-legged standing by healthy adults [1]. Two subjects had touchdowns on the force platform during testing. Both of these subjects touched down within the first 10 s, which were not included in the analysis, and one of these two subjects touched down between 38 and 39 s into the standing task. The data set analyzed for this subject was only 27 s in duration.

Customized software programs written in LabVIEW (version 3.11, National Instruments Corp., TX, USA) were used to process the force platform recordings. All force plate data were low pass filtered using a fourth order Butterworth filter with a cutoff frequency of 25 Hz. The video data were analyzed with a PEAK5 motion measurement system (Peak Performance Technologies, Inc., CO, USA). Each camera view was automatically digitized at 60 fields/s. The raw two-dimensional coordinates were then low passed filtered with a fourth order Butterworth filter at a cutoff frequency of 8 Hz. Three-dimensional coordinates were then calculated using the DLT.

2.5. Kinematic variable calculations

Heel and ankle inversion–eversion and medial–lateral ankle joint displacement were calculated from the three-dimensional kinematic data using PEAK5 and MATLAB (version 4.0, The MathWorks, Inc., MA, USA). The heel inversion–eversion and ankle inversion–eversion measurements were adjusted based on the subjects 'natural' inversion–eversion standing angle by subtracting the average inversion angle of the subject during eyes open bipedal standing from the angles obtained for one-legged standing. The ankle joint center was determined by assuming that the subtalar ankle joint is located at the intersection of the midline of the heel and a line connecting the tips of the medial and lateral malleoli. The distance of this point above the centroid of the bottom heel marker (h) was calculated from anthropometric measurements. The location of the subtalar ankle joint center was then calculated by using similar triangles (Fig. 2).



y_1, z_1 — y, z coordinates of bottom center of heel, known
y_2, z_2 — y, z coordinates of top center of heel, known
y_a, z_a — y, z coordinates of ankle joint, unknown
h — height of ankle joint above bottom center heel, known (this was calculated from anthropometric and marker placement measurements)

$$\frac{y_2 - y_1}{\sqrt{(y_2 - y_1)^2 + (z_2 - z_1)^2}} = \frac{y_a - y_1}{h} \quad (1)$$

Fig. 2. Schematic of the method used to calculate ankle joint center.

2.6. Inverse dynamics

Using the vertical and medial–lateral stabilographic and kinematic data, the two-dimensional ankle joint moment and joint reaction force were calculated in the frontal plane using inverse dynamic procedures, based on Newtonian mechanics. The segment masses, COM locations, and moments of inertia were determined using data from Zatsiorsky et al. [37]. The linear and angular accelerations of the COM of each segment were calculated from the video data. The equation used to calculate the ankle joint moment was

$$M_a = -F_z d_{z1} - F_y d_{y1} - F_{az} d_{z2} - F_{ay} d_{y2} + I\alpha \quad (2)$$

where M_a is the moment at ankle, F_z, F_y are vertical and medial–lateral ground reaction forces, F_{az}, F_{ay} are vertical and medial–lateral ankle joint forces, d_{z1}, d_{yl} are moment arms for F_z, F_y about the COM of the foot, d_{z2}, d_{y2} are moment arms for F_{az}, F_{ay} about the COM of the foot, I is the moment of inertia of the foot about the COM in the frontal plane, and α is the angular acceleration of the foot in the frontal plane.

In addition to the ankle joint moment and joint reaction force, a moment about the COM of the 'body-minus-foot' system, generated by the ankle joint reaction force, was calculated. The moment of the joint reaction force was calculated by multiplying the medial–lateral joint reaction force by the moment arm about the COM of the 'body-minus-foot' system. The existence of this moment follows from the basic dynamic equations for the multi-link serial chains [38,39].

$$\begin{aligned} F_a - mg - ma = 0 \\ M_a + r \times F_a - \dot{H}_i = 0 \end{aligned} \quad (3)$$

where F_a is the force acting on the leg at the ankle joint, m is the mass of the body above the foot, a is the acceleration of the COM of the 'body-minus-foot' system, M_a is the joint moment at the ankle, r is the radius from the center of mass to the ankle joint center, and $\dot{H}$ is the time rate of change of the angular momentum of the 'body-minus-foot' system. Note that F_a and gravity are the only external forces that are acting on the 'body-minus-foot' system. Because the mass of the foot is only 1.3% of the total body mass [37], the centers of mass for the entire body and for the 'body-minus-foot' system practically coincide. The equation for this moment is

$$M(F_{ay}) = r \cos\theta \times F_{ay} \quad (4)$$

where $M(F_{ay})$ is the moment of joint reaction force, F_{ay} is the medial–lateral joint reaction force, r is the distance from the COM to the ankle joint center, and θ is the inclination of r from vertical in the frontal plane. Eq. (4) was linearized, based on the assumption of small angles of θ (Eq. (5)), which was considered a valid assumption for one-legged standing, even in instances of extreme heel inversion.

$$M(F_a) = r \times F_a \quad (5)$$

2.7. Stabilographic variable calculations

Center of pressure (COP), gravity line position (GLP), and horizontal force were calculated for each trial. The GLPs were calculated using a zero-crossing algorithm developed by Zatsiorsky and King [40]. Horizontal force zero-crossings were determined by finding the instances when the horizontal force switched from negative to positive values. The medial–lateral acceleration of the COM was then integrated twice from a zero-crossing to zero crossing in a piecewise process to calculate the medial–lateral displacement of the COM.

2.8. Movement pattern analysis

Typically, maxima and minima are not variables of great importance in most biomechanical studies. The maxima and minima represent only the most extreme values of the observed variables; and thus they do not provide a robust measure for statistical purposes. However, when studying balance, it is exactly these extreme measures which are of interest. Being able to maintain balance, or return to equilibrium, from extreme positions is an important characteristic of balance control. Thus, to study different movement patterns, a strategy of examining the kinematic and stabilographic parameters in extreme situations was developed.

Two different balancing patterns were identified during the one-legged standing tasks. Instances during which the standing foot underwent an extreme rocking motion were separated from periods where the standing foot was relatively stationary by identifying points in time at which the ankle joint displacement was greater than ± 2 SD from the mean. During these instances in time, the medial/lateral forces, COP and GLP locations, several kinematic variables, and ankle joint moments were determined. Comparisons of these extreme instances were then made to periods when the ankle joint was not in extreme displacement (less than ± 2 SD).

3. Results

At instances of extreme displacement, the ankle joint could either be displaced medially, placing the foot in an everted position, or displaced laterally, placing the foot in an inverted position. A typical time history of heel inversion–eversion is illustrated in Fig. 3. The majority of subjects maintained a slightly inverted foot during periods of non-extreme ankle displacement, with an average inversion ankle of $4.5 \pm 2.1°$. The maximax medial displacement from the mean ankle position over all subjects was 9.2 mm, and maximax lateral displacement from the mean ankle position over all subjects was 14.8 mm. Maximax refers to the maximum value from all subjects.

Histograms of heel inversion–eversion revealed that five of the ten subjects had a bimodal type distribution of heel inversion–eversion while the other five did not (Fig. 4). Bimodal distributions suggest use of both a foot rocking technique and a stationary or stable foot technique to maintain balance during ECOL. Unimo-

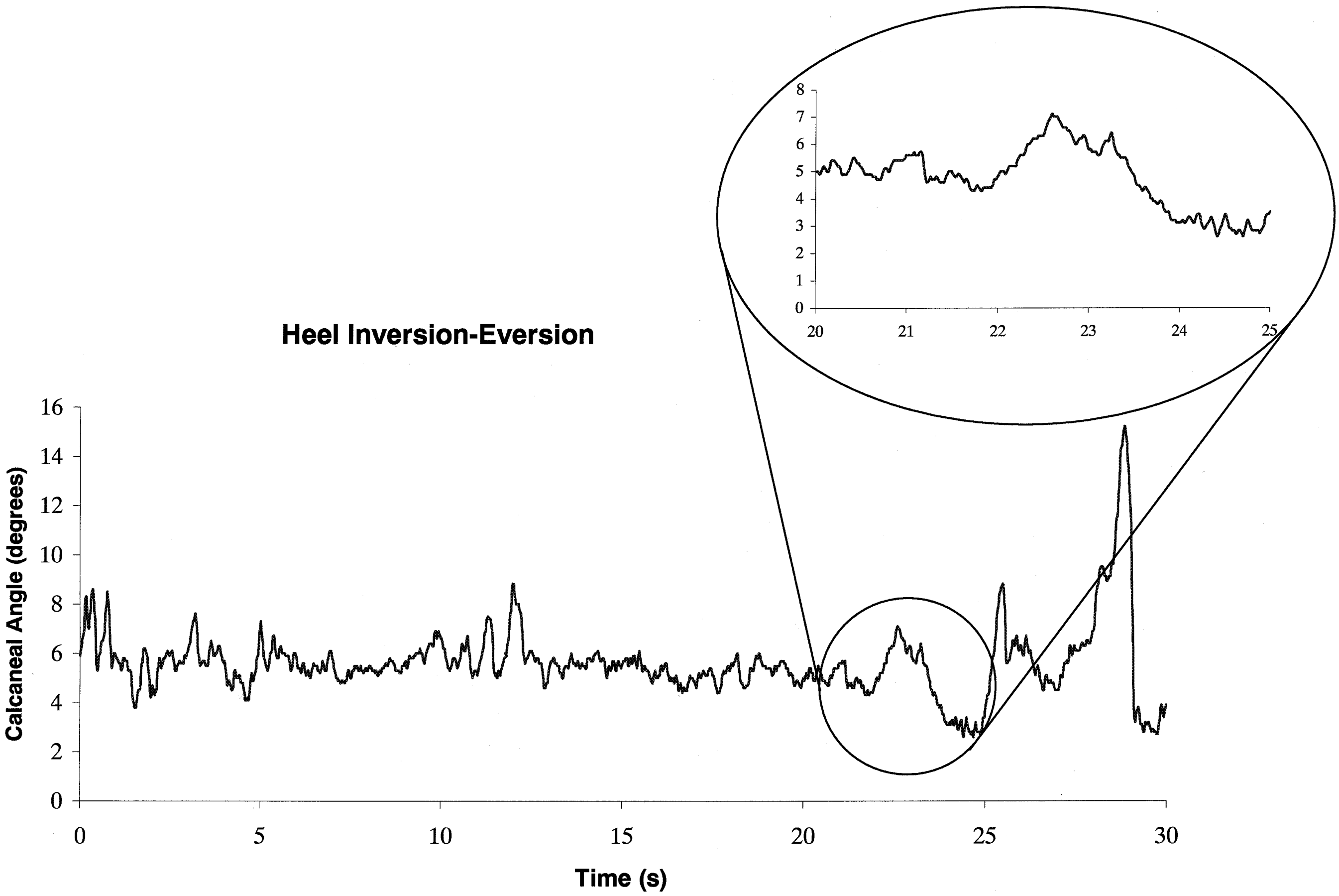


Fig. 3. Example of a typical COP and GLP migration pattern for eyes closed one-legged standing from subject 8.

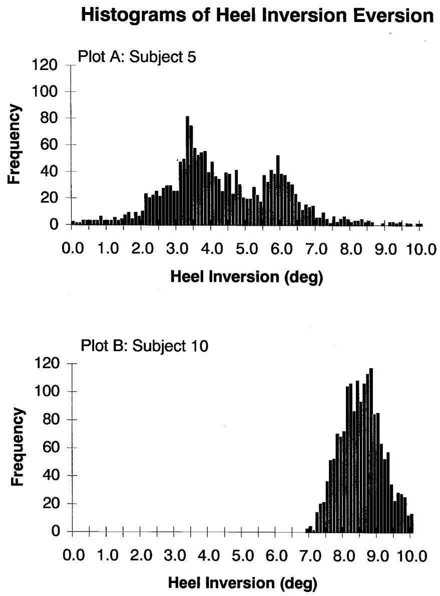


Fig. 4. Histograms of heel inversion and eversion for two different subjects. Plot A is subject 5 that illustrates a bimodal histogram. Plot B is subject 10 illustrating a unimodal histogram. The histogram was calculated over 30 s and each bar represents the number of occurrences of the heel angular position for every 0.1°.

dal distributions suggest a predominately stable foot technique for maintaining balance. During foot rocking, or extreme ankle displacement, the average horizontal (shear) force (7.0 ± 3.2 N) was approximately double the horizontal force during non-extreme ankle positions (3.5 ± 1.7 N). During extreme medial positions of the ankle joint, the horizontal force was directed laterally. During extreme lateral positions of the ankle joint, the horizontal force was directed medially.

The ankle joint moment was of similar value during instances of extreme (8.4 ± 4.9 Nm) and non-extreme (6.9 ± 3.5 Nm) ankle positions. However, the moment due to the joint reaction force was over four times greater during instances of extreme ankle position (3.4 ± 2.8 Nm) as compared to non-extreme ankle positions (0.8 ± 0.4 Nm). Maximal moments due to the joint reaction force, during extreme ankle displacement, were as high as 73% of the maximal ankle moment. Accordingly, the moment due to the ankle joint reaction force cannot be ignored.

As expected, the COP had a greater range of migration than the GLP, although, the two variables were grossly in phase (Figs. 5 and 6). The average range of the COP migration was 54.4 ± 14.3 mm and the average range of the GLP migration was 44.2 ± 11.3 mm from the mean position. Gravity line movement in the lateral–medial direction was roughly four times larger than the medial–lateral movement of the ankle joint center.

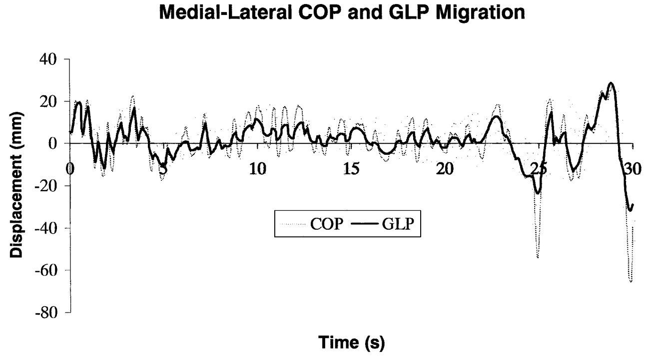


Fig. 5. Example of a typical heel inversion–eversion pattern for eyes closed one-legged standing from subject 8.

4. Discussion

A unique emphasis of this paper is the study of the extreme positions encountered during one-legged postural control whilst standing with eyes closed. The healthy young adult subjects in this study were able to return from extreme heel inversion angles of over 17° of inversion and 6° of eversion. During these instances of extreme medial–lateral ankle displacement, the average horizontal force was approximately double the average horizontal force experienced in periods of non-extreme ankle joint displacement and the moment due to the ankle joint reaction force was more than four times higher than during periods of non-extreme displacement. However, the ankle joint moment did not vary

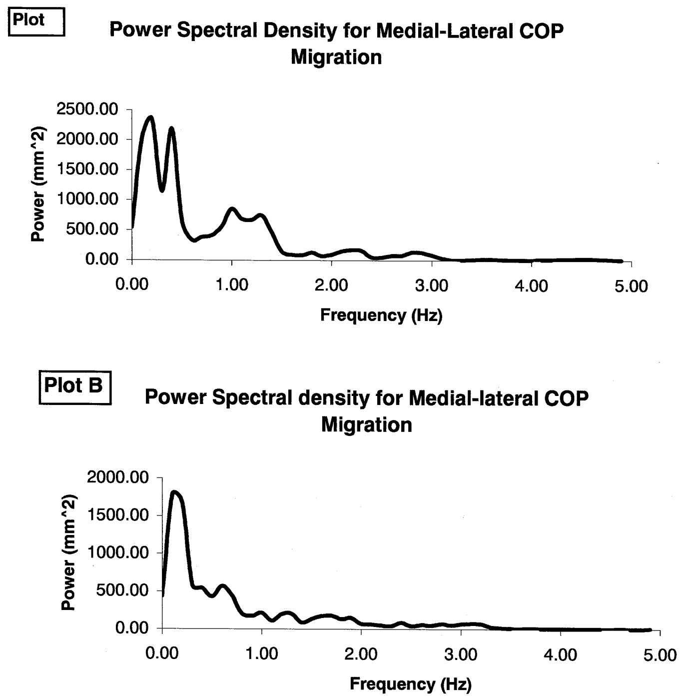


Fig. 6. Plots of power spectral density for medial-lateral COP migration.

substantially with ankle joint position. Accordingly, to recover balance from extreme positions the subjects utilized both a torque and a force technique at the ankle joint. In summary, it was demonstrated that: (a) a foot rocking motion was utilized by some subjects to maintain equilibrium; (b) the rocking motion was characterized by larger than average horizontal forces and greater than average GLP displacements, (c) the ankle joint moment was approximately of the same magnitude during extreme and non-extreme ankle displacement, and (d) the moment due to the reaction force in the ankle joint was over four times greater during instances of extreme ankle displacement as compared to non-extreme ankle displacement.

In 1997, Hoogvliet and colleagues determined that healthy adults performing eyes open standing demonstrated a foot rocking motion whilst standing. Results from their study indicated that ankle displacement and COP migration are highly related. In the frontal plane, COP migration did not represent body sway but rather the foot rocking motion utilized for balance during one-legged standing. Furthermore, the amplitude and or velocity of COP migration corresponded to an increased reliance on foot motion as a mechanism of balance control suggesting that in more challenging situations such as eye closure, lower limb injuries, and increased age, foot motion would become an increasingly important factor in balance maintenance.

Results from Hoogvliet [30] and the current study extend the results of Tropp and Odenrick [29] who studied healthy adults and adults with functional ankle instability in a one-legged standing task. Tropp and Odenrick found that frontal plane motions were predominately controlled via corrections at the ankle joint and that small angular displacement of the supporting leg created relatively large displacements of the COP. The large migration of the COP should then create a large torque about the ankle joint and restore the center of gravity to a more balanced position. However, the subjects in this study exhibited relatively large horizontal forces in addition to the ankle torque during periods of large ankle displacement.

As follows from the basic dynamic equations (Eq. (3)), two techniques principally exist to maintain equilibrium, assuming movement only at the ankle joint. The first technique is the *ankle moment technique*, described by a large torque about a fixed ankle joint used to maintain balance. The second possible technique is the *shear force technique*. At this point in time, the selection of the two techniques to maintain balance is not well understood. Anthropometric characteristics such as foot length, foot width, ankle height, arch, height, and body stature were not statistically different for those subjects who readily utilized the shear force technique to those who did not. Other factors that may differentiate the two groups are inversion and eversion ankle strength, ankle stability, and a history of balance training.

Previous research has shown improvements in balance with resistance training, primarily in eyes closed bipedal standing ability in the elderly [41,42]. However, little data are available on the effects of strength training on one-legged standing, particularly in healthy adults. Thus, while it is possible that ankle strength may be a determining factor in the selection of the two strategies during one-legged standing, more work needs to be done on the influence of lower extremity muscle strength and the selection of balance strategies.

Functional instability of the ankle has also been shown to affect the ability to maintain equilibrium in a group of soccer players [17]. However, results from this same study revealed that mechanical instability of the ankle was not a factor in maintaining equilibrium. Moreover, Tropp and Odenrick [29] reported that subjects with functional instability relied on increased motion of the hip to maintain equilibrium as compared to control subjects. These results may suggest that subjects with functional instability utilize hip motion either in addition to or in lieu of the shear force strategy.

Lastly, it has been well established that specific balance training is an effective method for improving balancing ability [10,11,7]. In particular, Gauffin and colleagues [7] demonstrated that ankle disk training in a group of soccer players with functional instability of the ankle improved postural control whilst standing on either the affected or non-affected ankles. Furthermore, the players demonstrated decreased hip motion following ankle disc training along with an increased reliance on the ankle for balance control.

While people who participated in activities that involve balancing tasks as fundamental skills, such as gymnastics and martial arts, were excluded from this study, subjects were not screened for their participation in other physical activities. It is possible that participation in activities such as soccer or tennis, for example, indirectly train neural and musculoskeletal systems utilized during one-legged standing, which may result in different utilization of ankle torque and shear force techniques during one-legged standing. However, further research is needed concerning the effects of ankle strength and physical activity on one-legged standing postural control.

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