Relation between Postural Stability and Plantar Flexors Muscle Volume in Young Males

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ABSTRACT

USHIYAMA, J., and K. MASANI. Relation between Postural Stability and Plantar Flexors Muscle Volume in Young Males. *Med. Sci. Sports Exerc.*, Vol. 43, No. 11, pp. 2089–2094, 2011. **Purpose**: It has been generally assumed that muscle volume is not a limiting factor of balance for young populations. To verify this assumption, this study investigated the relationship between postural stability and muscle volume of the plantar flexors, which have been regarded as the major agonist muscles for postural control, of healthy young male individuals. **Methods**: Forty-five healthy young males were requested to maintain quiet standing on a force platform for 60 s in eyes-open and eyes-closed conditions. Various time and frequency domain measures of the center of pressure were calculated. Muscle volume of the plantar flexors was estimated from muscle thickness in the lateral gastrocnemius and soleus muscles measured by ultrasonography and was divided by body mass to yield the normalized muscle volume. **Results**: Many time domain center of pressure measures such as mean distance, root mean square, and mean velocity were negatively correlated to normalized muscle volume ($P < 0.05$). **Conclusions**: It is suggested that, even among the young males, the muscle volume of their plantar flexors can act as a limiting factor for postural stability. **Key Words:** FORCE PLATFORM, CENTER OF PRESSURE, ULTRASONOGRAPHY, MUSCLE

Posturography, i.e., the use of measures of postural steadiness, is a commonly applied and accepted technique for assessing postural stability and control during upright standing. Because the end goal of postural control is to maintain the whole-body center of mass (COM) within the base of support, COM is the first parameter to assess for postural stability. However, because the procedure for measuring COM is time consuming and complex, many studies have used the center of pressure (COP), which is easier to measure than COM. Because COP approximately coincides with COM during quiet standing as a dynamic consequence (3,29), COP measurements contain valuable information on the overall COM fluctuation. Furthermore, because it is believed that COP summarizes the neuromuscular control response to the imbalances of the body's COM (29), COP measurements also contain valuable information on the used balance control strategy. Therefore, many studies have used COP time series as a posturography measure in an attempt to characterize the neural mechanisms responsible for balance control during quiet standing (5–7,16).

Recently, many researchers have also focused on the contribution of the maximal force-generating capacity of the muscles to the control mechanism responsible for balance control, especially within the elderly population. Because COP is proportional to the exerted ankle torque during quiet standing (13), the maximal force generation capacity of the muscles around the ankle should relate to the characteristics of ankle torque exertion, which can influence COP measure. For elderly individuals, it has been suggested that muscle strength is associated with the incidence of falls (22,24) and that falling is directly related to impaired postural stability (18,23). In support of such suggestions, Melzer et al. (19) recently demonstrated that muscle strength of plantar flexors is correlated with the postural stability of elderly persons. Furthermore, numerous studies have reported that resistance trainings on leg muscles have positive effects on improving COP measure as well as muscle strength and muscle volume among the elderly (2,12,26). Thus, as indicated in a guideline published by the American Geriatrics Society (1), muscle weakness is regarded as a risk factor for insufficient balance in the elderly population.

However, in the young population, the positive effect of resistance training has been rejected (9,28). For example, Kouzaki et al. (9) demonstrated that in the young population, resistance training of the leg muscles did not prevent diminishing postural stability due to bed rest for 20 d despite the maintenance of muscle volume of the plantar flexors

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(MV). Those studies may indicate that the maximal force generation capacity of the muscles is not related to COP measures. However, to date, no effort has been taken to investigate the relationship between the maximal force-generating capacity of the muscle and COP measures among a homogeneous subject population.

On the basis of the dynamics, during quiet standing, the calf muscles are dominant in controlling the ankle torque, and hence COP measure (13). Therefore, the purpose of this study was to investigate the relationship between COP measures and the maximal force generation capacity of the muscles responsible for the ankle torque generation among healthy young subjects. In the present study, we used the MV to represent the maximal force-generating capacity of the muscle. For the COP measure, we adopted various time and frequency domain measures reported by Prieto et al. (25). Postural stability evaluations have often included both eyes-open (EO) and eyes-closed (EC) conditions to estimate the role of the visual system in maintaining standing balance in the literature. Further, Prieto et al. (25) reported that especially in frequency domain measures, differences between young and elderly adults were prominent in the EC condition. Because there is a possibility that the relationship between the maximal force-generating capacity of the muscles and postural stability may be affected by eye condition, we therefore examined COP fluctuations during quiet standing in both EO and EC conditions.

METHODS

Subjects

Forty-five healthy young males participated voluntarily in this study. Table 1 summarizes the participants' anthropometric information. They had no history of neurological disorders. All subjects provided informed consent to participate in the study after receiving a detailed explanation of the purpose, potential benefits, and risks involved. The experimental procedures used in this study were approved in the ethical standards of the Committee on Human Experimentation at Keio University.

Anthropometric Estimation of Body Dimensions

When we approximate the human body as a single-joint inverted pendulum that rotates about the ankle joint, the dynamics equation of that inverted-pendulum model is described as:

$$I\ddot{\Theta} = mgh\sin\Theta - T \quad [1]$$

where Θ is the sway angle, m is the body mass excluding the feet, I is the moment of inertia of the body, h is the distance between COM and the ankle, g is the gravitational acceleration, and T is the total ankle torque (14). Because COM is located in front of the ankle joint, backward ankle torque, i.e., plantarflexion torque, is mainly responsible for preventing falling forward. On the basis of Masani et al. (14), we estimated h as $0.547H$, where H indicates the subject's height, and m as $0.971M$, where M indicates the subject's body mass. The amount of plantarflexion torque fluctuates at around $mgh\sin\Theta$ (29). Because $\sin\Theta$ does not vary much during quiet standing, the amount of $mgh\sin\Theta$ primarily depends on mgh. Thus, the amount of plantarflexion torque required to maintain the upright posture mainly depends on the subjects' m and h. To discuss which parameter between these can affect the required amount of plantarflexion torque more, we estimated the coefficient of variation (CV) (SD $\times$ 100/mean) of mg and h. If h has a much smaller variation than mg, we can consider that the effect of mg on the between-subject variation of the amount of plantarflexion torque is dominant compared with h. The lower leg length (LL) was also determined as the distance from the articular cleft between the femur and tibia to the lateral malleolus.

TABLE 1. Physical characteristics of the subjects (n = 45).

Age (yr)	20.5 ± 1.60
H (cm)	173.6 ± 4.85
h (cm)	94.9 ± 2.66
M (kg)	65.2 ± 7.03
m (kg)	63.3 ± 6.82
MT (cm)	6.85 ± 0.51
LL (cm)	40.4 ± 2.10
MV (cm^2)	977.4 ± 129.7
normalized-MV ($cm^2 \cdot kg^{-1}$)	15.6 ± 1.84

Mean ± SD values of all physical characteristics are shown.

Experimental Protocol

Subjects were requested to maintain quiet standing on a force platform (Type 9281C; Kisler, Zürich, Switzerland) for approximately 70 s in EO and EC conditions. They were instructed to maintain an upright standing posture with their arms at the sides of their bodies. The distance between their heels was 18 cm, and the angle between their feet was 12° opening laterally; these values are based on the mean values of the natural stance position of healthy young adults reported in a previous study (17). In EO condition, subjects were requested to look straight ahead at a visual reference. In EC condition, they were asked to close their eyes after looking at the visual reference. One trial was conducted for each condition, and sufficient resting time was allowed between trials. The order of the trials was randomized among the subjects. All signals from the force platform were stored with a sample frequency of 100 Hz on a personal computer using a 16-bit analog-to-digital converter (PH-701; DKH, Tokyo, Japan). After 10 s of steady-state quiet standing, the data from the latter 60 s were used for analysis.

COP Measures

The outputs of the force platform allow us to compute the COP time series in anterior–posterior (AP) and medial–lateral (ML) directions. On the basis of Prieto et al. (25), the data were low-pass filtered using a second-ordered zero-phase Butterworth digital filter with a cutoff frequency at 5 Hz. In addition, we subtracted the mean value from the COP time series in each window to remove the offset. The resultant distance (RD) time series, which is the vector distance from the mean COP to each pair of points in the AP and ML time series, was calculated as follows:

$$RD[n] = [AP[n]^2 + ML[n]^2]^{1/2} n = 1, 2, \ldots, N \qquad [2]$$

where N is the number of data points included in the analysis. The time and frequency domain COP measures were calculated to evaluate the age-related changes in the postural stability on the basis of Prieto et al. (25), using MATLAB (The MathWorks, Inc., Natick, MA). Each measure was calculated by averaging the measures estimated from three 20-s windows of the recorded data according to Prieto et al. (25).

Time domain measures. The measures described here are the most commonly used measures for postural stability. We calculated these measures by referring to the equations described in Prieto et al. (25).

To evaluate COP fluctuations, we calculated 12 parameters from RD, AP, and ML time series. They are 1) the mean distances of COP ($MDIST_{RD}$, $MDIST_{AP}$, $MDIST_{ML}$), which are the average distances of COP from the mean; 2) the root mean square distances from the mean COP (RMS_{RD}, RMS_{AP}, RMS_{ML}), which are equivalent to the SD of the zero-meaned COP time series; 3) the mean velocities of the COP ($MVELO_{RD}$, $MVELO_{AP}$, $MVELO_{ML}$), which are calculated by normalizing the total length of the COP path by the period; and 4) the ranges of the COP displacement ($RANGE_{RD}$, $RANGE_{AP}$, $RANGE_{ML}$), which are the peak-to-peak ranges of COP values.

To evaluate the area of COP fluctuations, we calculated two parameters. They are 1) the area of the 95% confidence circle (AREA-CC), which is the area of a circle with a radius equal to the one-sided 95% confidence limit of RD time series, and 2) the area of the 95% confidence ellipse (AREA-CE), which is the area of the 95% bivariate confidence ellipse.

As for the time domain "hybrid" measures, we also calculated four measures. They are 1) the sway area (AREA-SW), which is the sum of the area of the triangles formed by two consecutive points of the COP path and the mean COP, and 2) the mean frequencies of RD, AP, and ML time series ($MFREQ_{RD}$, $MFREQ_{AP}$, $MFREQ_{ML}$), which are the sinusoidal oscillations derived from the mean distance and the mean velocity.

Frequency domain measures. Frequency domain measures were also analyzed to evaluate the qualitative characterization of the frequency distribution of the displacement of COP. The power spectral density was computed for the RD, AP, and ML time series, for the entire period of each window using the Welch method. The frequency domain measures were calculated for the frequency range from 0.15 to 5 Hz. The first two frequency points past the direct-current component (0.05 and 0.10 Hz) were not included in the analysis of the power spectrum densities because such low-frequency components represent events that occur once every 10 or 20 s and probably do not provide significant information about the postural control system (25). The following 12 measures were defined for the RD, AP, and ML time series on the basis of the equations reported in Prieto et al. (25). They are 1) the 50% power frequencies ($PF50_{RD}$, $PF50_{AP}$, $PF50_{ML}$), which are the median power frequencies or the frequencies below which 50% of the total power is found; 2) the 95% power frequencies ($PF95_{RD}$, $PF95_{AP}$, $PF95_{ML}$), which are the frequencies below which 95% of the total power is found; 3) the centroid frequencies ($CFREQ_{RD}$, $CFREQ_{AP}$, $CFREQ_{ML}$), which are the frequencies at which the spectral masses are concentrated; and 4) the frequency dispersion ($FREQD_{RD}$, $FREQD_{AP}$, $FREQD_{ML}$), which are unitless measures of the variability in the frequency content of the power spectral density.

Muscle Volume Estimation

Muscle thickness of the lateral gastrocnemius and soleus (MT) was measured using the B-mode ultrasonic apparatus (model SSD-900; ALOKA, Tokyo, Japan). During the ultrasonic measurements, subjects kept standing with their arms and legs relaxed in an extended position. According to Miyatani et al. (21), the measurement site was precisely determined on the posterior surface at 30% of LL. An ultrasonic probe (UST-579T; ALOKA) with a frequency of 7.5 MHz was attached perpendicular to the underlying muscle and bone tissue. The probe was coated with water-soluble transmission gel to provide acoustic contact without depressing the dermal surface. The interface between the subcutaneous adipose tissue and the lateral gastrocnemius muscle and that between the soleus muscle and the tibia bone were identified from ultrasonic images. The distance between each interface was measured as representative MT. The accuracy of the measurements was verified in a previous study (20).

Miyatani et al. (21) reported the multiple regression equation for predicting muscle volume from MT and LL for young males. In their reports, the estimated MV was highly correlated with the muscle volume measured by MRI ($r^2 = 0.832$, $P < 0.05$). Thus, we estimated the MV using the equation as follows:

$$MV = 219.9MT + 31.3LL - 1758.0 \qquad [3]$$

In addition, MV was divided by the body mass to yield the normalized muscle volume (normalized-MV).

Statistical Analysis

The effect of eye conditions on each COP measure was analyzed using a paired Student's t-test. The Pearson correlation coefficients between each COP measure and normalized-MV were also calculated. The first significant level for all statistical comparisons was chosen at $P < 0.05$.

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All statistical analyses were performed using PASW statistics software (SPSS Japan, Inc., Tokyo, Japan).

RESULTS

Measurements of the physical characteristics. Table 1 summarizes mean ± SD values of all physical characteristics of the subjects. CV value of *mg* (10.78%) was larger than that of *h* (2.80%).

Influences of eye conditions on each sway measure. Table 2 shows mean ± SD values of all COP measures under EO and EC conditions. As the results of paired Student's *t*-tests, significant differences between EO and EC conditions were detected for all of the time domain COP measures such as $MDIST_{RD}$, $MDIST_{AP}$, $MDIST_{ML}$, RMS_{RD}, RMS_{AP}, RMS_{ML}, $MVELO_{RD}$, $MVELO_{AP}$, $MVELO_{ML}$, $RANGE_{RD}$, $RANGE_{AP}$, $RANGE_{ML}$, AREA-CC, AREA-CE, AREA-SW, $MFREQ_{RD}$, $MFREQ_{AP}$, and several frequency domain measures such as $PF95_{RD}$, $PF95_{ML}$, $CFREQ_{ML}$, and $FREQD_{RD}$ (all, $P < 0.05$).

Influences of estimated MV on each sway measure. Table 3 represents the Pearson correlation coefficient, *r*, of the relationship between each COP measure and normalized-MV under EO and EC conditions. Significant negative correlations with normalized-MV were observed in many time domain measures such as $MDIST_{RD}$, $MDIST_{ML}$, RMS_{RD}, RMS_{ML}, $MVELO_{RD}$, $MVELO_{AP}$, $MVELO_{ML}$, $RANGE_{RD}$, $RANGE_{ML}$, AREA-CC, AREA-CE, and AREA-SW in both EO and EC conditions (all, $P < 0.05$). $FREQD_{ML}$ were positively correlated with normalized-MV in EO and EC conditions ($P < 0.05$). Furthermore, $MDIST_{AP}$, RMS_{AP}, and $RANGE_{AP}$ showed significant negative correlation (all, $P < 0.05$), and $FREQD_{RD}$ showed significant positive correlation with normalized-MV only in EO condition ($P < 0.05$).

TABLE 2. Time and frequency domain measures of COP fluctuation during quiet standing under EO and EC conditions.

Measure	EO	EC	Difference
$MDIST_{RD}$ (mm)	3.40 ± 0.93	3.88 ± 1.06	***
$MDIST_{AP}$ (mm)	2.53 ± 0.79	2.85 ± 0.76	**
$MDIST_{ML}$ (mm)	1.73 ± 0.57	2.03 ± 0.75	***
RMS_{RD} (mm)	3.87 ± 1.03	4.43 ± 1.18	***
RMS_{AP} (mm)	3.12 ± 0.90	3.54 ± 0.91	***
RMS_{ML} (mm)	2.17 ± 0.72	2.57 ± 0.92	***
$MVELO_{RD}$ (mm·s^{-1})	8.78 ± 2.57	11.3 ± 3.05	***
$MVELO_{AP}$ (mm·s^{-1})	5.71 ± 1.65	8.00 ± 2.13	***
$MVELO_{ML}$ (mm·s^{-1})	5.44 ± 1.78	6.30 ± 2.05	***
$RANGE_{RD}$ (mm)	8.91 ± 2.38	10.4 ± 2.83	***
$RANGE_{AP}$ (mm)	14.2 ± 3.35	17.0 ± 4.11	***
$RANGE_{ML}$ (mm)	10.8 ± 3.56	13.1 ± 4.65	***
AREA-CC (mm^2)	143 ± 71.8	188 ± 94.7	***
AREA-CE (mm^2)	125 ± 61.7	174 ± 93.3	***
AREA-SW (mm^2·s^{-1})	9.88 ± 4.32	14.5 ± 6.77	***
$MFREQ_{RD}$ (Hz)	0.44 ± 0.12	0.49 ± 0.12	***
$MFREQ_{AP}$ (Hz)	0.44 ± 0.15	0.53 ± 0.15	***
$MFREQ_{ML}$ (Hz)	0.60 ± 0.16	0.60 ± 0.16	
$PF50_{RD}$ (Hz)	0.41 ± 0.16	0.44 ± 0.11	
$PF50_{AP}$ (Hz)	0.31 ± 0.10	0.30 ± 0.08	
$PF50_{ML}$ (Hz)	0.39 ± 0.11	0.38 ± 0.09	
$PF95_{RD}$ (Hz)	1.54 ± 0.47	1.64 ± 0.42	*
$PF95_{AP}$ (Hz)	1.18 ± 0.37	1.13 ± 0.34	
$PF95_{ML}$ (Hz)	1.20 ± 0.28	1.11 ± 0.28	*
$CFREQ_{RD}$ (Hz)	0.85 ± 0.23	0.88 ± 0.20	
$CFREQ_{AP}$ (Hz)	0.68 ± 0.20	0.63 ± 0.16	
$CFREQ_{ML}$ (Hz)	0.74 ± 0.17	0.68 ± 0.15	**
$FREQD_{RD}$	0.72 ± 0.05	0.71 ± 0.05	*
$FREQD_{AP}$	0.74 ± 0.05	0.72 ± 0.06	
$FREQD_{ML}$	0.70 ± 0.07	0.69 ± 0.06	

Means ± SD values of all COP fluctuation measures are shown.
Significant differences in each measure between EO and EC conditions (Difference) are also listed: * $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$.

TABLE 3. Relationship between each COP measure and normalized-MV.

Measure	EO	EC
$MDIST_{RD}$	−0.533***	−0.387**
$MDIST_{AP}$	−0.426**	−0.253
$MDIST_{ML}$	−0.514***	−0.455**
RMS_{RD}	−0.538***	−0.394**
RMS_{AP}	−0.436**	−0.277
RMS_{ML}	−0.530***	−0.454**
$MVELO_{RD}$	−0.516***	−0.398**
$MVELO_{AP}$	−0.541***	−0.342*
$MVELO_{ML}$	−0.441**	−0.405**
$RANGE_{RD}$	−0.503***	−0.372*
$RANGE_{AP}$	−0.482***	−0.272
$RANGE_{ML}$	−0.584***	−0.505***
AREA-CC	−0.524***	−0.380*
AREA-CE	−0.557***	−0.411**
AREA-SW	−0.617***	−0.459**
$MFREQ_{RD}$	−0.033	0.023
$MFREQ_{AP}$	−0.103	−0.055
$MFREQ_{ML}$	0.070	0.144
$PF50_{RD}$	−0.171	−0.016
$PF50_{AP}$	−0.034	0.007
$PF50_{ML}$	0.024	0.028
$PF95_{RD}$	−0.158	−0.048
$PF95_{AP}$	−0.076	−0.020
$PF95_{ML}$	−0.160	0.039
$CFREQ_{RD}$	−0.073	0.036
$CFREQ_{AP}$	−0.050	0.035
$CFREQ_{ML}$	0.153	0.116
$FREQD_{RD}$	0.314*	0.144
$FREQD_{AP}$	0.160	0.023
$FREQD_{ML}$	0.323*	0.326*

Pearson correlation coefficients, *r*, are shown in the table.
Significant correlations between each COP measure and normalized-MV are also listed: * $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$.

DISCUSSION

The purpose of the present study was to investigate the relationship between various time and frequency domain COP measures during quiet standing and the maximal force-generating capacity of the plantar flexors in the young males. We demonstrated that many time domain COP measures showed significant negative correlations with normalized-MV under both EO and EC conditions.

As shown in Table 3, significant negative correlations were observed between many time domain COP measures and normalized-MV in both eye conditions. Because the results concerning ML series might be indirectly derived from the correlation between the volume of adductor or abductor muscles around the hip joint and that of the plantar flexors, we focused on the results concerning RD and AP series that may be directly influenced by the MV in this section.

The COP measures, which showed significant negative correlations with normalized-MV (Table 3), have been used to evaluate the amount of COP fluctuations in other studies. Thus, the present findings indicate that the more MV relative to the body mass, the less the COP fluctuates during quiet standing in the young male subjects. In addition, it was demonstrated that CV values were larger in *mg* (10.78%) than in *h* (2.80%), indicating that there was a larger individual variation in *mg* than in *h*. Thus, it is reasonable to assume that *mg* had greater influence on the between-subject variation of the plantarflexion torque for maintaining balance during quiet standing. Because Fukunaga et al. (8) reported that muscle volume is a major determinant of maximal isometric joint torque in humans, it is suggested that the more MV relative to the body mass, the less relative plantarflexion torque (percent of maximal voluntary contraction torque) we need to exert to control balance during quiet standing. Furthermore, it was demonstrated that, in the plantar flexors, the amount of force fluctuation linearly increases with increment of the contraction level during isometric voluntary force matching task (30). In addition, Kouzaki and Shinohara (10) demonstrated that fluctuations in exerted force during low-intensity plantarflexion were correlated with postural sway during quiet standing. Thus, it is suggested that the subjects who had less MV relative to their body mass need to produce a higher percent of maximal voluntary contraction torque for plantarflexion to keep balance during quiet standing, such that it would create larger fluctuations in their COP. The present findings suggest a possibility that increasing muscle volume via resistance training would have positive effects toward improving the postural stability even for young males.

Moreover, it is possible that postural stability is determined not only by the precision of the exerted plantarflexion torque but also by the mechanical stiffness of the ankle joint (4,11). Recently, Seynnes et al. (27) demonstrated that muscle volume correlates with tendon stiffness, by representing that training-induced increase in muscle volume shows significant positive correlation with increment of tendon stiffness. If so, we could assume that negative correlations between time domain COP measures and normalized-MV observed in the present subjects reflect the influences of the ankle stiffness, which is indirectly led from MV, on the postural stability. To verify this assumption, further investigation will be needed regarding the relationship between muscle volume, ankle stiffness, and postural stability.

In addition, our results represent a significant positive correlation between $FREQD_{RD}$ and normalized-MV in the EO condition. According to Maurer and Peterka (16), FREQD is a measure showing large positive correlations with the gain of the derivative part of postural control systems, K_D, which is the component to control body sway velocity. Considering this matter with observed negative correlations between time domain COP measures and normalized-MV, it is assumed that the subjects who have larger normalized-MV adopted a control strategy relying on the velocity information and, consequently, the amount of their COP fluctuations becoming smaller during quiet standing. This assumption supports our previous proposition indicating the importance of body sway velocity information on controlling balance during quiet standing (13,15).

CONCLUSIONS

We demonstrated that postural stability is attributable to the muscle volume of the plantar flexors, which have been regarded as a major agonist muscle group for quiet standing, even in young males. Although it has been generally assumed that muscle volume is not a limiting factor of balance for young individuals, this study suggests that such an assumption may not be true.

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REFERENCES

1. American Geriatrics Society; British Geriatrics Society; American Academy of Orthopaedic Surgeons Panel on Falls Prevention. Guideline for the prevention of falls in older persons. *J Am Geriatr Soc*. 2001;49(5):664–72.
2. Buchner DM, Cress ME, de Lateur BJ, et al. The effect of strength and endurance training on gait, balance, fall risk, and health services use in community-living older adults. *J Gerontol A Biol Sci Med Sci*. 1997;52(4):M218–24.
3. Caron O, Faure B, Brenière Y. Estimating the centre of gravity of the body on the basis of the centre of pressure in standing posture. *J Biomech*. 1997;30(11–12):1169–71.
4. Casadio M, Morasso PG, Sanguineti V. Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application. *Gait Posture*. 2005;21(4):410–24.
5. Collins JJ, De Luca CJ. Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories. *Exp Brain Res*. 1993;95(2):308–18.
6. Collins JJ, De Luca CJ. The effects of visual input on open-loop and closed-loop postural control mechanisms. *Exp Brain Res*. 1995;103(1):151–63.
7. Collins JJ, De Luca CJ, Burrows A, Lipsitz LA. Age-related changes in open-loop and closed-loop postural control mechanisms. *Exp Brain Res*. 1995;104(3):480–92.
8. Fukunaga T, Miyatani M, Tachi M, Kouzaki M, Kawakami Y, Kanehisa H. Muscle volume is a major determinant of joint torque in humans. *Acta Physiol Scand*. 2001;172(4):249–55.
9. Kouzaki M, Masani K, Akima H, et al. Effects of 20-day bed rest with and without strength training on postural sway during quiet standing. *Acta Physiol (Oxf)*. 2007;189(3):279–92.

10. Kouzaki M, Shinohara M. Steadiness in plantar flexor muscles and its relation to postural sway in young and elderly adults. *Muscle Nerve*. 2010;42(1):78–87.
11. Loram ID, Lakie M. Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for stability. *J Physiol*. 2002;545(3):1041–53.
12. Lord SR, Clark RD, Webster IW. Physiological factors associated with falls in an elderly population. *J Am Geriatr Soc*. 1991;39(12):1194–200.
13. Masani K, Popovic MR, Nakazawa K, Kouzaki M, Nozaki D. Importance of body sway velocity information in controlling ankle extensor activities during quiet stance. *J Neurophysiol*. 2003;90(6):3774–82.
14. Masani K, Vette AH, Kouzaki M, Kanehisa H, Fukunaga T, Popovic MR. Larger center of pressure minus center of gravity in the elderly induces larger body acceleration during quiet standing. *Neurosci Lett*. 2007;422(3):202–6.
15. Masani K, Vette AH, Popovic MR. Controlling balance during quiet standing: proportional and derivative controller generates preceding motor command to body sway position observed in experiments. *Gait Posture*. 2006;23(2):164–72.
16. Maurer C, Peterka RJ. A new interpretation of spontaneous sway measures based on a simple model of human postural control. *J Neurophysiol*. 2005;93(1):189–200.
17. McIlroy WE, Maki BE. Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. *Clin Biomech (Bristol, Avon)*. 1997;12(1):66–70.
18. Melzer I, Benjuya N, Kaplanski J. Postural stability in the elderly: a comparison between fallers and non-fallers. *Age Ageing*. 2004;33(6):602–7.
19. Melzer I, Benjuya N, Kaplanski J, Alexander N. Association between ankle muscle strength and limit of stability in older adults. *Age Ageing*. 2009;38(1):119–23.
20. Miyatani M, Kanehisa H, Fukunaga T. Validity of bioelectrical impedance and ultrasonographic methods for estimating the muscle volume of the upper arm. *Eur J Appl Physiol*. 2000;82(5–6):391–6.
21. Miyatani M, Kanehisa H, Ito M, Kawakami Y, Fukunaga T. The accuracy of volume estimates using ultrasound muscle thickness measurements in different muscle groups. *Eur J Appl Physiol*. 2004;91(2–3):264–72.
22. Perry MC, Carville SF, Smith IC, Rutherford OM, Newham DJ. Strength, power output and symmetry of leg muscles: effect of age and history of falling. *Eur J Appl Physiol*. 2007;100(5):553–61.
23. Piirtola M, Era P. Force platform measurements as predictors of falls among older people—a review. *Gerontology*. 2006;52(1):1–16.
24. Pijnappels M, van der Burg PJ, Reeves ND, van Dieën JH. Identification of elderly fallers by muscle strength measures. *Eur J Appl Physiol*. 2008;102(5):585–92.
25. Prieto TE, Myklebust JB, Hoffmann RG, Lovett EG, Myklebust BM. Measures of postural steadiness: differences between healthy young and elderly adults. *IEEE Trans Biomed Eng*. 1996;43(9):956–66.
26. Schlicht J, Camaione DN, Owen SV. Effect of intense strength training on standing balance, walking speed, and sit-to-stand performance in older adults. *J Gerontol A Biol Sci Med Sci*. 2001;56(5):M281–6.
27. Seynnes OR, Erskine RM, Maganaris CN, et al. Training-induced changes in structural and mechanical properties of the patellar tendon are related to muscle hypertrophy but not to strength gains. *J Appl Physiol*. 2009;107(2):523–30.
28. Torvinen S, Kannus P, Sievänen H, et al. Effect of four-month vertical whole body vibration on performance and balance. *Med Sci Sports Exerc*. 2002;34(9):1523–8.
29. Winter DA, Patla AE, Prince F, Ishac M, Gielo-Perczak K. Stiffness control of balance in quiet standing. *J Neurophysiol*. 1998;80(3):1211–21.
30. Yoshitake Y, Shinohara M, Kouzaki M, Fukunaga T. Fluctuations in plantar flexion force are reduced after prolonged tendon vibration. *J Appl Physiol*. 2004;97(6):2090–7.