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Motor mechanisms of balance during quiet standing

David A. Winter *, Aftab E. Patla, Milad Ishac, William H. Gage

Department of Kinesiology, University of Waterloo, Waterloo, Ont. N2L 3G1, Canada

Abstract

The purpose of this paper is to highlight the motor mechanisms involved in balance as the human, as a biped, continuously defends against gravitational and internal forces to maintain a safe posture. The search for these mechanisms needs precise and valid 3D measurements including both limbs plus valid biomechanical models. The literature shows the need for two force platforms to separate the mechanisms at the ankle and hip (load/unload mechanism). Also, precise measures (≈0.03 mm) of markers on a multi-segment 3D bilateral model are required to record the minute trajectories of all segments and joints. The controlled variable, center-of-mass, is seen to be virtually in phase with the controlling variable, the center-of-pressure, which suggests a 0th order system where a simple series elastic spring could maintain balance. The first model involves a mass/spring/damper of medial/lateral balance: the stiffness was varied with stance width and the predicted sway from a spring controlled inverted pendulum closely matched the experimentally measured stiffness and sway. The second was a non-linear model of the plantarflexor series elastic elements which resulted in three closely validated predictions of anterior/posterior balance: the locus of the gravitational load line, the predicted ankle moment and the ankle stiffness at the operating point.

Keywords: Quiet standing; Balance; Measurements; Mechanisms; Biomechanical models

1. Measurement challenges

Quiet standing as a human "movement" has been confounded at the measurement level for two major reasons. First, the vast majority of studies have been limited to force plate studies using only one force platform [1]. Second, because the displacements, velocities and accelerations of the individual segments and joints are so small that the vast majority of optical systems are not precise enough to yield meaningful kinematic data. The major kinetic variable measured from a force platform is the centre of pressure (COP) and for a single platform produces a two-dimensional "spaghetti" diagram that was the spatial/temporal summation of the anterior/posterior (A/P) COP and medial/lateral (M/L) COP. The control of both these COPs was considered to be the ankle muscles. Not until two platforms were used to separate the contributions of the individual limbs was it possible to identify the fact that the "spaghetti" plot was nothing more than the spatial/temporal summation of two completely separate mechanisms [2]. The major kinematic variable reported was the body centre of mass (COM) and this has varied from crude single point one-dimensional estimates [3] to a 3D, 14 segment bilateral estimate [4]. Standard TV based optical systems lack the precision; however, IRED based systems, such as OPTOTRAC, have a precision of 0.03 mm which is sufficient to record displacements of the leg (≈0.2–0.5 mm) to the head (≈2–4 mm). Thus accurate estimates of the total body COM are possible (≈1.5 mm in the M/L direction and 3 mm in the A/P direction).

2. Methods and theory

For the series of separate studies reported here the measurement systems and anthropometric model were identical and are now summarized. Ground reaction forces and COP measures from each foot were recorded from two Advanced Mechanical Technology force platforms. Both A/P and M/L COPs were recorded: the left COP (COP_l) and right COP (COP_r) plus the COP_{net} as calculated from a weighted average of the two COPs:

$$COP_{net} = COP_l * R_{vl}/(R_{vl} + R_{vr}) + COP_r * R_{vr}/(R_{vl} + R_{vr}), \quad (1)$$

* Corresponding author. Fax: +1-519-747-1894.
E-mail address: dwinter@golden.net (D.A. Winter).

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where: R_{vl} and R_{vr} are the vertical reaction forces from the left and right feet respectively.

Eq. (1) identifies the two separate mechanisms of COP control. COP_l and COP_r are the control under the individual feet by the ankle muscles while the fraction of body weight taken by each foot, $(R_{vl}/(R_{vl} + R_{vr}))$, is controlled by the hip muscles. Depending on the individual study the subjects were asked to stand quietly with their feet at different spacings or different positions. The 14 segment 3D bilateral model to estimate the total body COM is presented in Fig. 1. It consists of legs (2), thighs (2), lower arms (2), upper arms (2), trunk (4), pelvis and head. The location of the 21 infrared emitting diodes (IREDs) are numbered and the segment definitions and mass fractions are presented in Table 1. This kinematic data was recorded using a 3-D optoelectronic camera system (OPTOTRAK®; Northern Digital, Waterloo, Ontario, Canada).

University students with no known balance or gait pathology volunteered to participate in this study. Prior to their participation, each participant provided informed consent to the potential risks associated with their participation. The participants their normal footwear with shorts and an athletic shirt throughout the experiment. Tight fitting clothing was worn to reduce movement of markers placed on the body. Approval of this study was provided by the Office of Human Research, at the University of Waterloo, Waterloo, Ontario, Canada.

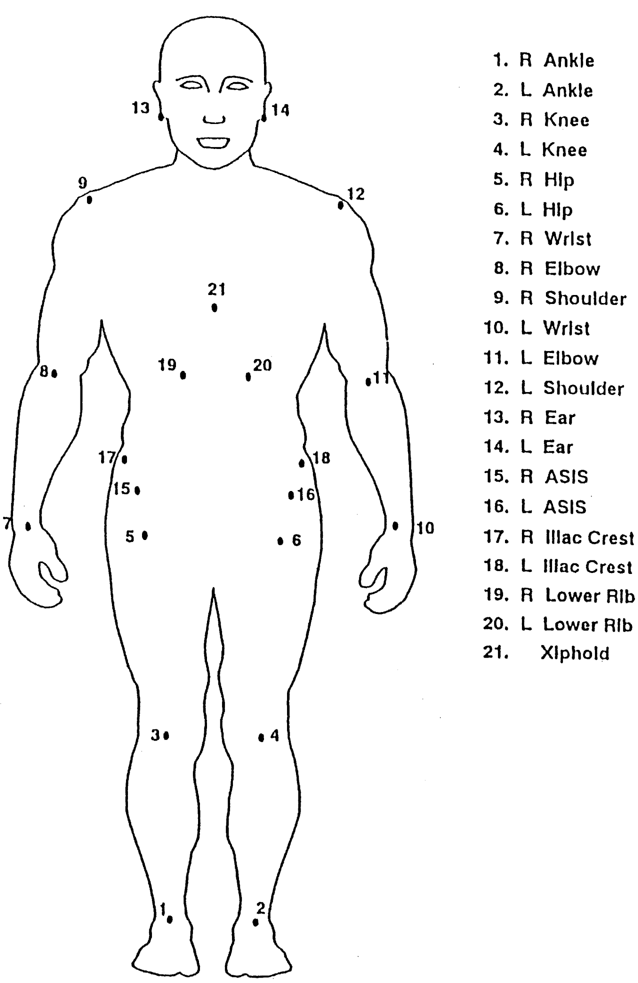

Fig. 1. Location of 21 IREDs in the 14 segment 3D bilateral model to estimate total body COM. Table 1 presents the definition of each segment and its mass fraction.

3. Results of study #1—separation of motor mechanisms that control COP

Ten young adults participated in this study [2]. They were required to stand quietly for two minutes in three separate foot positions: side-by-side with the ankle-to-ankle distance equal to their hip-to-hip distance, with the left foot forward of the right with the ankle-to-ankle angle = 45°, and with the feet in a tandem position. Figs. 2 and 3 demonstrate the input data that led to the identification of these separate mechanisms. In side-by-side standing Fig. 2 presents the COP_l and COP_r records in combination with COP_{net} as calculated from Eq. (1). It is quite apparent that COP_l and COP_r move forwards and backwards primarily under control of the plantarflexor muscles. A weighted average of COP_l and COP_r results in a COP that virtually matches the A/P COP but has negligible contribution to the M/L COP. Fig. 3 shows body weight taken by the feet and it is apparent that the M/L COP is virtually in phase with fraction of body weight taken by the individual feet. This mechanism of M/L has been described as a "load/unload" mechanism because it is controlled by the simultaneous loading of one limb and unloading of the opposite limb by the hip abd/adductor muscle. Fig. 4 combines the contribution of these separate mechanisms: the contribution of the load/unload mechanism due to vertical ground reaction forces, COP_v, moves the COP between the feet while the net effect of the ankle muscle control of the COPs under both feet, COP_c, moves the COP in an A/P direction. Thus we see two separate mechanisms at work and in the side-by-side position they are orthogonal, thus they are not only separate but they are independent. Thus the "spaghetti" plot is a temporal and spatial summation of two separate and independent mechanisms. In other stance positions, such as tandem and 45° these separate mechanisms have been documented [2]. In the 45° position the trajectories of the load/unload mechanism were not orthogonal to the ankle mechanism so these separate mechanisms were not independent.

4. Results of study #2—validation of the inverted pendulum model

A number of researchers have predicted that the difference between the COP and COM should be directly

Table 1
Segment definitions

Segment	Mass fraction	Marker definition of COM
Head	0.081	(13 + 14)/2
Trunk 4	0.136	(9 + 12 + 21)/3
Trunk 3	0.078	((19 + 20)/2 + 21)/2
Trunk 2	0.065	(17 + 18 + 19 + 20)/4
Trunk 1	0.078	(17 + 18 + 15 + 16)/4
Pelvis	0.142	(15 + 16)/2
Thighs	0.100 (2)	0.433 × 3 + 0.567 × 5 and 0.433 × 4 + 0.567 × 6
Legs and feet	0.060 (2)	0.606 × 1 + 0.394 × 3 and 0.606 × 2 + 0.394 × 4
Upper arms	0.028 (2)	0.436 × 8 + 0.564 × 9 and 0.436 × 11 + 0.564 × 12
Forearms	0.022 (2)	0.682 × 7 + 0.318 × 8 and 0.682 × 10 + 0.318 × 11
Total	1.000	

$$COP_{net}(t) = COP_l(t) \bullet \frac{R_{vl}(t)}{R_{vl}(t)+R_{vr}(t)} + COP_r(t) \bullet \frac{R_{vr}(t)}{R_{vl}(t)+R_{vr}(t)}$$

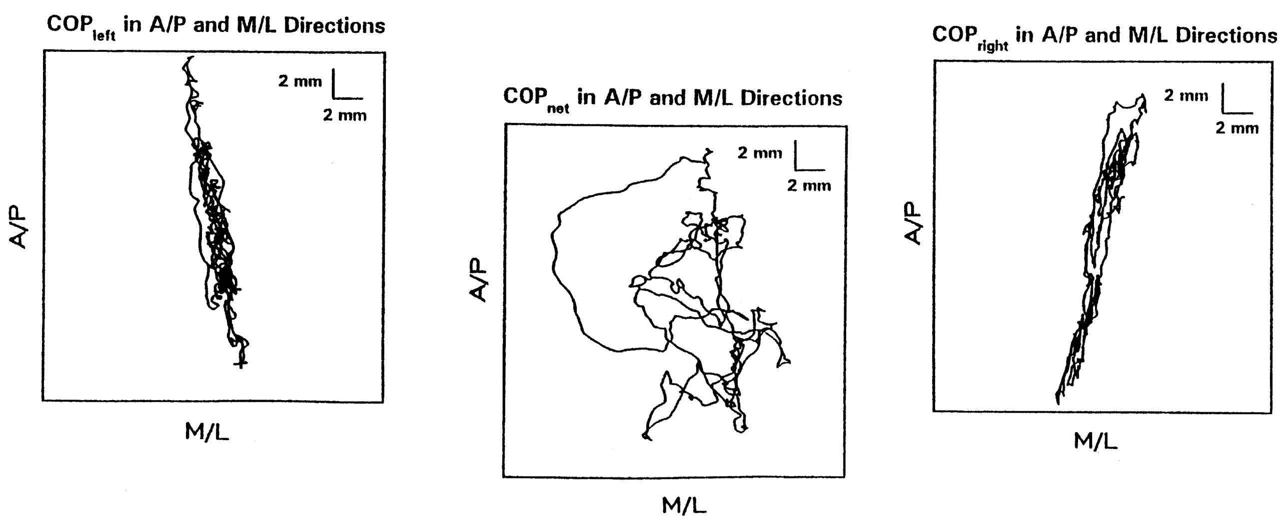

Fig. 2. Centres of pressure recorded separately from left and right feet during quiet standing in the side-by-side position. COP_{net} is calculated from the formula and is a weighted average of COP_l and COP_r. It is evident that the average of COP_l and COP_r would result in an A/P movement of the COP_{net} but negligible M/L displacement. Thus it is evident that the fractional loading term in the equation is responsible for the M/L displacement of COP_{net}. See Fig. 3 for confirmation of the load/unload mechanism in controlling the M/L displacement.

correlated with the horizontal acceleration of the COM in either the A/P or M/L directions [4–7]. With precise and accurate measures of COP and COM it now possible to develop a valid biomechanical model of quiet standing: the inverted pendulum. In both A/P and M/L directions a simple inverted pendulum model [4] showed that:

$$COP - COM = -K \bullet \ddot{COM} \quad (2)$$

where: $K = I/Wh$, an individual anthropometric variable; I is the total body moment of inertia about the ankles, W is body weight and h is height of COM above the ankles.

$\ddot{COM}$ is the horizontal acceleration of the COM in either the A/P or M/L directions.

A regression of COP−COM vs $\ddot{COM}$ for ten subjects yielded an average r of −0.91 in the A/P direction and −0.78 in the M/L direction [4]. The lower M/L score was due to the fact that COP–COM was only twice the noise level while the A/P measures were five times the noise level [4]. More recently for 11 young adults standing quietly in the side-by-side position [8] by more precise placement of the IREDs and the reduction of drift in the COP records gave a higher r of −0.954 in the A/P direction and −0.84 in the M/L direction. Eq. (2) gives us the framework to understand the horizontal accelerations of the COM, those acceleration which could be either destabilizing or stabilizing. COM is the passive controlled variable while COP if the active con-

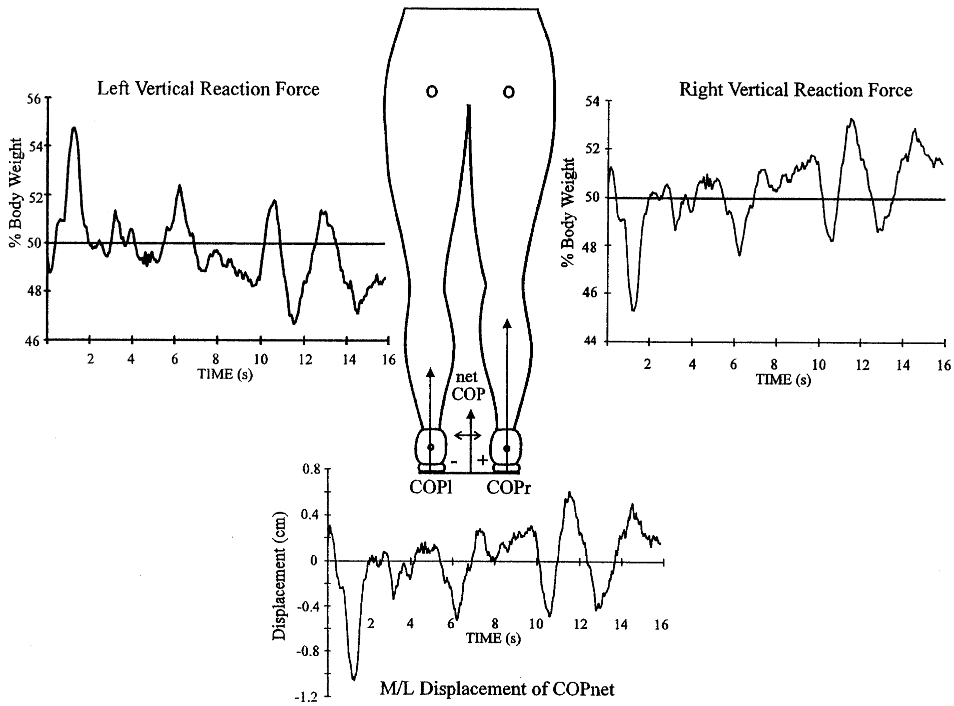

Fig. 3. For the same trial as presented in Fig. 2 the left and right vertical ground reaction forces (as a % of body weight) are plotted along with the M/L displacement of COP_{net}. These waveforms demonstrate virtually 100% control of M/L COP by the “load/unload” mechanism. An inverse dynamics analysis of the kinetics in the frontal plane shows that this load/unloading is controlled by the hip abd/adductor muscles.

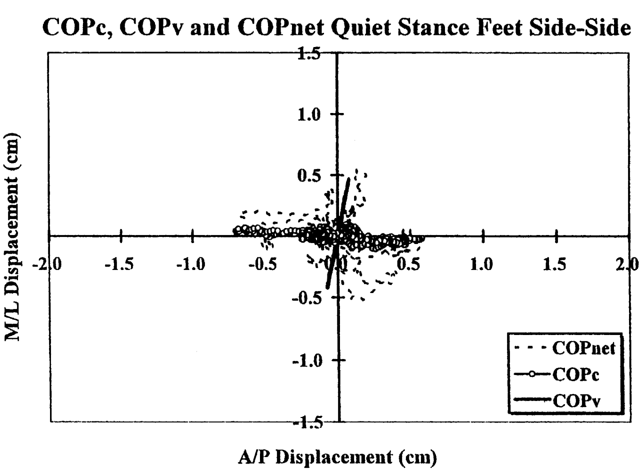

Fig. 4. Partitioning of the COP_{net} into contributions by the ankle muscles to the centre of pressure under each foot (COP_c) and the load/unload mechanism due to the vertical ground reaction forces (COP_v). The “spaghetti” plot of COP_{net} is seen to be a spatial and temporal summation of these two separate and independent mechanisms.

trolling variable. In quiet standing COP oscillates either side of COM to keep COM in a safe and fairly constant position between the two feet. If COP is ahead of COM it accelerates COM posteriorly; if COP is to the right of COM it accelerates it to the left, etc.

Detailed segment and joint kinematics of quiet standing have not yet been reported. COM measures are a weighted summation of the trajectories of all body segments in both A/P and M/L planes but do not necessarily reflect the fact that all segments and joints are moving in unison. A recent study [8] using the 14 segment bilateral model revealed that all segments move in synchronization with the total body COM and that the angular movement at the ankle is highly correlated with the total body COM. For one of the subjects Fig. 5 plots all the segment rms trajectories vs height of each segment above the ankle yielded a straight line passing though the ankle at (0,0). For an ideal rigid inverted pendulum the regression should yield a perfectly straight line with an $R^2 = 1$. For 11 subjects standing quietly for two minutes the average R^2 for this linear regression averaged 0.966 in the A/P direction and 0.944 in the M/L direction. The correlation between the individual segment and the total body COM trajectories (shown beside each segment) for this trial ranged from $r^2 = 0.770$ to 0.999 and averaged 0.972. Thus it is quite evident that quiet standing all segments from the legs up to the head move in unison. Also, a correlation between the individual ankle joint angular displacements and the total body COM gave an average r^2 of 0.876 for the left ankle, 0.891 for the right ankle and 0.896 for the average left and right ankles. Thus in the simplified inverted pendulum model (Eq. (2)) COM is a valid representation of all segments and reinforces the assumption that COM is the single controlled variable.

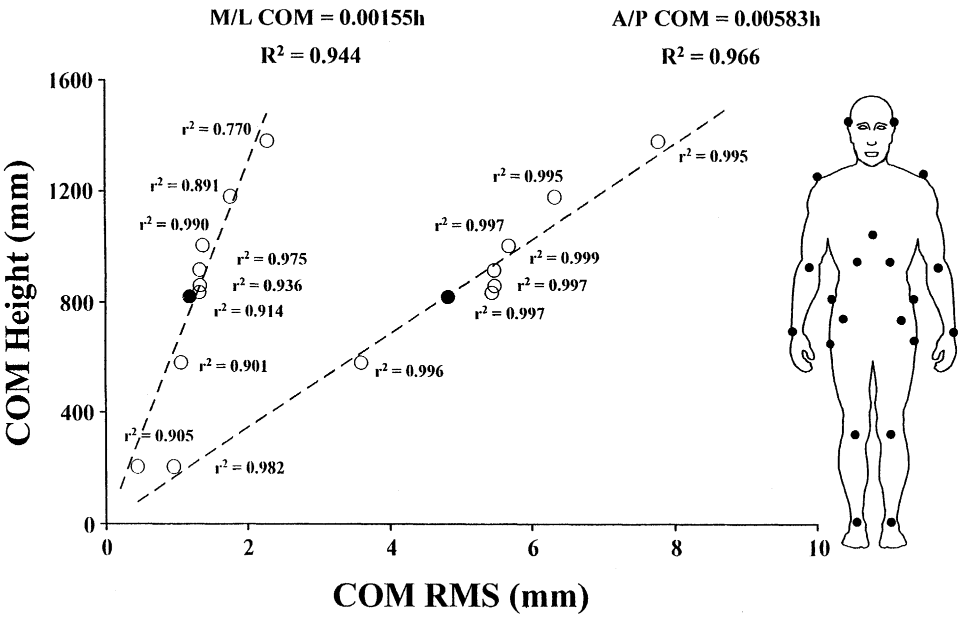

Fig. 5. Height of the COM of each segment vs its horizontal displacement in the A/P and M/L planes. The origin is at the ankle and a linear regression is drawn through the segment COMs. The correlation of this regression is $R^2 = 0.966$ in the A/P direction and 0.944 in the M/L indicating that all segments are moving virtually in unison. The total body COM (solid circle) was correlated with each segment's COM and their correlations are shown; they vary from $r^2 = 0.770$ to $r^2 = 0.999$ with an average of 0.972. This gives further evidence that COM_{net} is the single controlled variable and supports the simplified inverted pendulum model in Fig. 6.

5. Results of study #3—reactive vs passive stiffness control

The debate that now arises as to whether COP is under active reactive control or passive stiffness control. If reactive control were involved some form of sensory input must be available at the joint, muscle or vestibular levels. Our precise measures of joint angle displacements (0.1° in M/L, 0.25° in A/P), angular velocities (0.05°/s in M/L, 0.16°/s in A/P) and head accelerations (1.1 cm/s^2 in M/L, 1.9 cm/s^2 in A/P) are all at or below reported sensory threshholds [4]. Also, if reactive control of some sensory system were present it would track some segment or joint kinematic variable, all of which are highly correlated with the COM. Then there would be afferent and efferent latencies (≈50–70 ms) plus delays in build up of muscle tension (≈70–100 ms) as predicted by the newly recruited muscle twitches. Thus A/P COP, which is controlled by the plantarflexors, would be predicted to lag the COM by 120 to 170 ms. Similarly the M/L COP, which is controlled by the hip abd/adductors, would be predicted to have a similar lag behind the COM. However, our measures of COM show that the COP lags by only 4 ms [4]. Such a close relationship suggests a near 0th order control, i.e. a simple passive spring system with a small amount of damping. As the body swayed forward the ankle spring increases it tension causing the COP to increase and move ahead of the COM, thus causing a posterior acceleration of the COM. Then as the body swing backwards the spring tension decreases causing the COP to decrease and move behind the COM and decelerate its backward movement.

5.1. Stiffness model #1—the tuned mass/spring/damper system

The theoretical relationship between M/L sway and stiffness were closely matched by experimental results [4]. M/L sway was the only direction where sway could be varied experimentally and that was by standing in three separate stance widths; 50, 100 and 150%, where the 100% had the ankle-to ankle distance equal to the distance between the hip joints. The inverted pendulum model showed the relationship between sway angle as:

$$\theta_{sw} = c(K - \mathrm{mgh})^{-0.5} \quad (3)$$

where: c is an anthropometric constant for the inverted pendulum, K is the spring constant (N.m/rad.), m is the mass of the body above the ankle in kg, g is the gravitational constant and h is the height of the COM above the ankles in m.

The combined term "mgh" represents the gravitational load which acts like a spring because it varies linearly with θ_{sw}. Thus K must exceed the gravitational spring for viable balance and it is the difference between the muscle spring and the gravitational spring that drives the inverted pendulum. The spring constant, K, was estimated from frequency response curve of COP–COM which was seen to represent a second order tuned mechanical system [4]. Knowing the resonant frequency, ω_n, of the system it was possible to estimate K from:

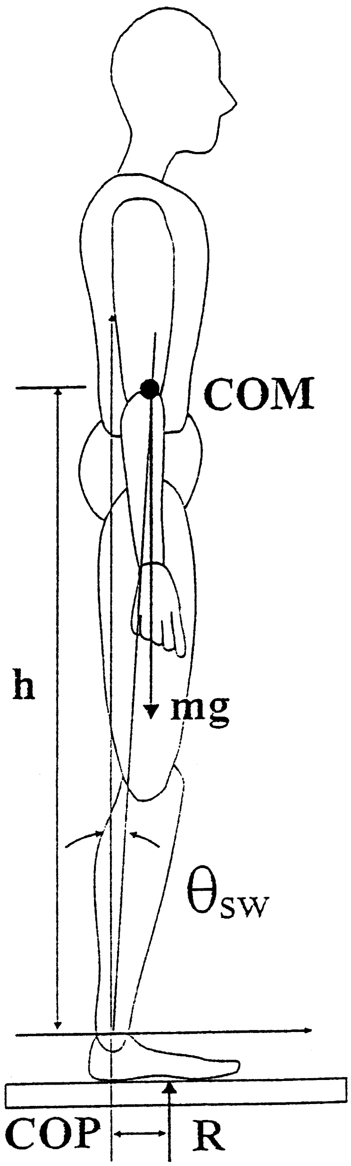

Fig. 6. Simplified inverted pendulum model used for the determination of ankle stiffness. The total body sway angle, θ_{sw}, is calculated from COM and its height, h, above the ankle and is related to the ankle moment, M_a, calculated from the COP and the vertical ground reaction force, R. Since h and R are constants in quiet standing M_a and θ_{sw} fluctuate directly as COP and COM respectively.

$$K = I\omega_n^2 \tag{4}$$

where: I is the moment of inertia of the body about the ankle joint.

For ten subjects standing quietly for two minutes in the three stance widths a plot of sway vs K revealed a curve that closely matched the theoretical Eq. (3):

$$\theta_{sw} = 7.66(K-\text{mgh})^{-0.55}$$

5.2. Stiffness model #2—simple regression of ankle moment and sway angle

More recently a second and more direct technique has been introduced that avoids the somewhat difficult curve fitting of the tuned mechanical system and also negates the need for a long data record (≈2 min) to get a valid FFT of COP–COM. Fig. 6 presents an inverted pendulum model where independent COP and COM measures have been taken. With the COM trajectory as an input the sway angle, θ_{sw}, is predicted from:

$$\theta_{sw} = \text{COM}/\text{h}. \tag{5}$$

The ankle moment, M_a, is calculated using inverse dynamics:

$$M_a = \text{COP}\bullet\text{R}-m_f g x_f \tag{6}$$

where: R is the ground reaction force, COP is the distance of COP anterior of the ankle, m_f is the mass of the foot, g is the gravitational acceleration and x_f is horizontal distance of the foot COM to the ankle.

In quiet standing R is a constant equal to body weight and $m_f g x_f$ is a constant for each subject; thus the fluctuations in M_a are directly in phase with the COP fluctuations. With time records of θ_{sw} and M_a we can estimate the ankle stiffness from the slope of a simple linear regression of short duration ($<$ 10 sec) of M_a vs θ_{sw}. The average ankle stiffness for ten subjects using this technique [9] always exceeded the gravitational load, mgh, and the average r^2 for the regression was 0.918 indication the in-phase relationship as predicted by a simple spring mechanism.

5.3. Anatomical rationale for A/P stiffness control

More recent evidence is now available as to why the ankle A/P stiffness is always higher than gravitational load (which varies linearly with sway angle) is suggested by the non-linear series elastic plantarflexor characteristics reported by Winters and Stark [10]. They report a non-linearity that is close to a square law, thus the tension in the plantarflexors would increase as the square of the sway angle. Thus at the operating point of the system (the intersection of the gravitational load line and the non-linear series elastic curve) the plantarflexor ankle moment would increase twice as fast as the gravitational load. The operating point is decided by the plantarflexor muscle tone which could drift over time causing the operating point also to drift but the changes in the plantarflexion moment about that changing operating point would always be twice that of the gravitational load. A recent model assumes a non-linearity where the ankle moment, M_a is a function of θ_{sw} to the power n:

$$M_a = k_1\theta_{sw}^n \tag{7}$$

Fig. 7 presents this non-linear model in conjunction with a representative gravitational load line, mgh. The intersection of the series elastic non-linearity with the gravitational load line is the operating point at a sway angle, θ_o and an ankle moment, M_o. At that instant in time θ_{sw} is oscillating about θ_o and M_a can be predicted from:

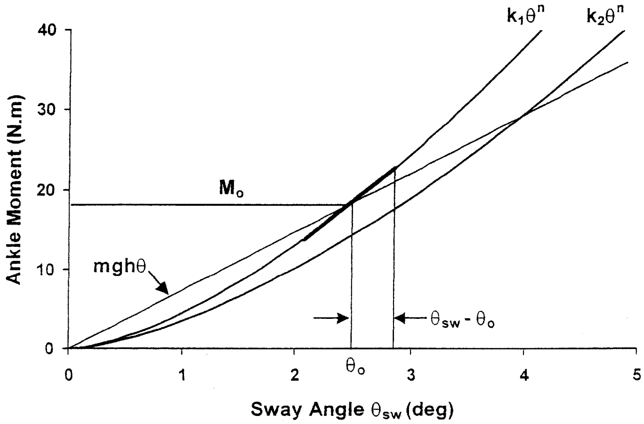

Fig. 7. Model of plantarflexor non-linear series elastic element, $K_1\theta^n$, showing its operating point with the gravitational load line, mgh2. The operating point, θ_o and M_o will move slowly up and down the load line as the muscle tone changes slowly during quiet standing. Input to this model is θ_{sw} which fluctuates either side of the operating point as COM fluctuates primarily due to mass shifts in the thoracic/lumbar region as a result of respiration. Fluctuations of θ_{sw} around θ_o are used to predict fluctuations of M_a (Eq. (8)); "*n*" is varied for an optimal comparison with independently measured M_a (Eq. (6)).

$$M_a(t) = mgh(\theta_o + n(\theta_{sw}(t) - \theta_o)). \tag{8}$$

COM fluctuates because it is perturbed internally by respiratory and cardiac mass shifts [11]. Thus θ_{sw} fluctuates in phase with COM and from Eq. (8) was used to predict M_a. For 11 subjects standing quietly for two minutes "n" was varied until the predicted M_a (Eq. (8)) most closely agreed with the independently calculated M_a (Eq. (6)). A perfect agreement between the predicted and measured M_a would result in an rms difference of 0 N.m and a regression of the predicted vs calculated M_a would yield an $r^2 = 1$ and a slope of 1. The results from our model gave an average rms difference of 0.61 N.m, an average r^2 of 0.975 with an average slope of 0.994. The non-linearity power "*n*" ranged from 1.55 to 2.11 with an average of 1.65. The locus of the operating point was also plotted and it produces a straight line which very closely agreed with the gravitational load line for each subject. Finally, the estimated stiffness from the non-linear model (Eq. (8)) is nmgh N.m/rad. The non-linear model estimates of the ankle stiffness at the operating point agreed within 6% of the experimental estimates.

6. Discussion

The need for a complete 3D bilateral multisegment model of quiet standing is now obvious. Without precise measures of segment and joint angles it would not be possible to validate the inverted pendulum model presented in Fig. 6. This model allowed a simple calculation of M_a from COP (Eq. (1)) and θ_{sw} from COM (Eq. (2)). These precise measures also validated that $\theta_{sw} \approx \theta_a$ and thus permitted the development of several models of balance control. This precision demonstrated that COM was virtually in phase with COP thus indicating that the control was a 0th order control suggesting a simple spring control of balance. This same precision enabled quantification of the extremely small displacements, velocities and accelerations at sensory sites which might be involved in reactive control. These potential sensory inputs were all at or below reported thresholds further reinforcing the concept of a simple stiffness control. Three different techniques were presented to estimate and model the joint stiffness; all resulted in stiffness that were safely greater than the gravitational stiffness as defined by the slope of the gravitational load line, mgh N.m/rad. The simple non-linear passive model at the ankle with nothing more than θ_{sw} as input yielded an extremely close prediction of M_a. The inherent simplicity of such a balance control system is that the sensory system is not involved any more than setting the muscle tone and even this can be somewhat "sloppy" because changes in muscle tone would predict the operating point to drift up and down the gravitational load line. This slow drift of the operating point up and down that load line was experimentally validated.

Morasso and Schieppati [12] have criticized such a stiffness model and claim that the in-phase relationship between COP and COM is a consequence of physical laws. In their model they claimed that a PD (proportional + derivative) feedback controller could produce this in-phase relationship. However, such a neurological PD controller was pure speculation and more importantly, their model neglected normal afferent and efferent latencies which would have resulted in a motor delay of 50–70 ms. Thus, even with the benefit of their speculative PD controller the motor variable COP would have lagged COM by 50–70 ms which disagrees with the experimental lag of 4 ms.

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