Review article

Human balance and posture control during standing and walking

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Introduction

The fact that we as humans are bipeds and locomote over the ground with one foot in contact (walking), no feet in contact (running), or both feet in contact (standing) creates a major challenge to our balance control system. Because two-thirds of our body mass is located two-thirds of body height above the ground we are an inherently unstable system unless a control system is continuously acting. The purpose of this review is to introduce the reader to the basic anatomy, mechanics, and neuromuscular control that is seen in the current literature related to balance control.

The degeneration of the balance control system in the elderly and in many pathologies has forced researchers and clinicians to understand more about how the system works and how to quantify its status at any point in time. With the increase in our ageing population and with increased life expectancy of our elderly the importance of maintaining mobility is becoming ever more critical. Injuries and loss of life due to falls in the elderly is a very major factor facing them. Table 1 compares the deaths due to falls with the deaths due to motor vehicle accidents (MVA) as a function of age as reported by Statistics Canada in 1991. The startling fact is that the deaths due to falls in the 80+ population is almost as high as the MVA deaths in the accident-prone 15–29-year group. However, when we look at the deaths per 100 000 for these two causes the figures are even more startling. The young 15–29-year-old group had 21.5 MVA deaths per 100 000 while the elderly deaths due to falls was 185.6 per 100 000. That is almost nine times as many. Society's concern about the 'slaughter on the highways' of our young people should also be focused on the elderly and their exorbitant death rate due to falls.

Virtually all neuromusculoskeletal disorders result in some degeneration in the balance control system[1]. Because of the ability of the CNS to adapt for the loss of function a given pathology may not be apparent until the patient is temporarily deprived of the compensating system. Vestibular patients, for example, have excessive reliance on vision, so when they close their eyes or walk in a dark area they become very unstable. Pathologies that have special balance challenges would include the following: chronic ankle sprains, chronic degenerative low back pain, scoliosis, paroxysmal positional vertigo, head injury, stroke, cerebellar disease, Parkinson's disease, vestibular deficits, peripheral neuropathies, amputation, and cerebral palsy.

If we look at the epidemiology of falls we see reports that about 50% of the falls occur during some form of locomotion[2–5]. It is during walking that we challenge our system the most: initiating and terminating gait, turning, avoiding obstacles (altering step length, chang-

Table 1. Accidental deaths in Canada (1991) due to motor vehicle accidents versus falls as a function of age. (Stats Canada – 1991)

Age Group	MVA	Per 100 000	Falls	Per 100 000
0–14	253	4.4	14	0.2
15–29	1401	21.5	54	0.8
30–39	595	12.5	56	1.2
40–49	375	9.8	44	1.2
50–59	271	10.6	75	2.9
60–69	273	12.2	137	6.1
70–79	267	18.0	404	27.3
80+	157	23.0	1269	185.8
Total	3592	13.2	2053	7.5

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ing direction, stepping over objects, etc.), bumping into people and objects. Thus, much of the research has attempted to perturb the balance system in a large number of ways in order to quantify the human response. When the researcher is interested in reactive responses he/she introduces unexpected perturbations. On the other hand, proactive control will require either voluntarily initiated internal perturbations (such as raising an arm) or anticipatory well-learnt perturbations (such as is experienced many times over the walking cycle).

Three major sensory systems are involved in balance and posture. Vision is the system primarily involved in planning our locomotion and in avoiding obstacles along the way. The vestibular system is our 'gyro', which senses linear and angular accelerations. The somatosensory system is a multitude of sensors that sense the position and velocity of all body segments, their contact (impact) with external objects (including the ground), and the orientation of gravity. Neurophysiologists have devised a wide range of experiments to tease out the contribution of each of these systems and even to confuse the system by providing conflicting or false sensory inputs.

In spite of the epidemiological evidence that most falls occur during one form of locomotion or another, a large body of research into balance during quiet and perturbed standing has evolved. This research has yielded considerable information regarding the role of each of the three sensory systems and how the resultant redundancy can assist when one of the systems fails or is impaired.

Biomechanical models of balance are now emerging. Inverted pendulum models can be used to explore how the CNS controls balance. Especially in the kinetics of human movement we see the integrated control evident at each joint and in entire limbs. The CNS is totally aware of the problems of controlling a multisegment system and the interlimb coupling that can facilitate balance control. Here biomechanical analyses are extremely valuable in identifying the goals and synergies of the CNS and pinpointing the total limb or body synergies that accomplish those goals.

Basic definitions

Posture. This describes the orientation of any body segment relative to the gravitational vector. It is an angular measure from the vertical.

Balance. Balance is a generic term describing the dynamics of body posture to prevent falling. It is related to the inertial forces acting on the body and the inertial characteristics of body segments.

Centre of mass (COM). This is a point equivalent of the total body mass in the global reference system (GRS) and is the weighed average of the COM of each body segment in 3D space. It is a passive variable controlled by the balance control system. The vertical projection of the COM onto the ground is often called the centre of gravity (COG). Its units are metres (m).

Centre of Pressure (COP). This is the point location of the vertical ground reaction force vector. It represents a weighted average of all the pressures over the surface of the area in contact with the ground. It is totally independent of the COM. If one foot is on the ground the net COP lies within that foot. If both feet are in contact with the ground the net COP lies somewhere between the two feet, depending on the relative weight taken by each foot. Thus when both feet are in contact there are separate COPs under each foot. When one force platform is used only the net COP is available. Two force platforms are required to quantify the COP changes within each foot. The location of the COP under each foot is a direct reflection the neural control of the ankle muscles. Increasing plantarflexor activity moves the COP anteriorly, increasing invertor activity moves it laterally. Its units are metres (m). In the literature there is a major misuse of the COP when it is referred to as 'sway', thereby inferring that it is the same as the COG. Unfortunately some researchers even refer to the COP directly as the COG[6].

Quiet standing

Quiet standing has been the subject of scores of research papers. The major measure that has been recorded is COP_{net} from a single force plate. The excursions of the COP_{net} in both anteroposterior (A/P) and mediolateral (M/L) directions have been reported. However, before any review and discussion of these reports it is important to see the relationship between the COP_{net} and the COG during quiet standing. The most common position of the feet is a side-by-side position and the most commonly discussed control is in the A/P direction using an 'ankle-strategy'. More will be said later about different strategies of control.

The difference between the COG and COP has been recognized by a number of researchers[7,8]. To demonstrate this difference and at the same time introduce an inverted pendulum model of balance in the A/P direction we introduce Figure 1. Here we see a subject swaying back and forth while standing erect on a force plate. Each figure shows the changing situation at five different points over time. Time 1 has the body's COG ahead of the COP, and the angular velocity ω is assumed to be clockwise. Body weight W is equal and opposite to the vertical reaction force R, and this 'parallelogram of forces' acts at distance g and p respectively from the ankle joint. Both W and R will remain constant during quiet standing. Assuming the body to be an inverted pendulum, pivoting about the ankle, a counterclockwise moment equal to Rp and a clockwise moment equal to Wg will be acting.

$$Rp - Wg = I\alpha \qquad \text{(Eq. 1)}$$

where: I is the moment of inertia of the total body about the ankle joint ($kg.m^2$)
α is the angular acceleration of the inverted pendulum ($r\ s^{-2}$).

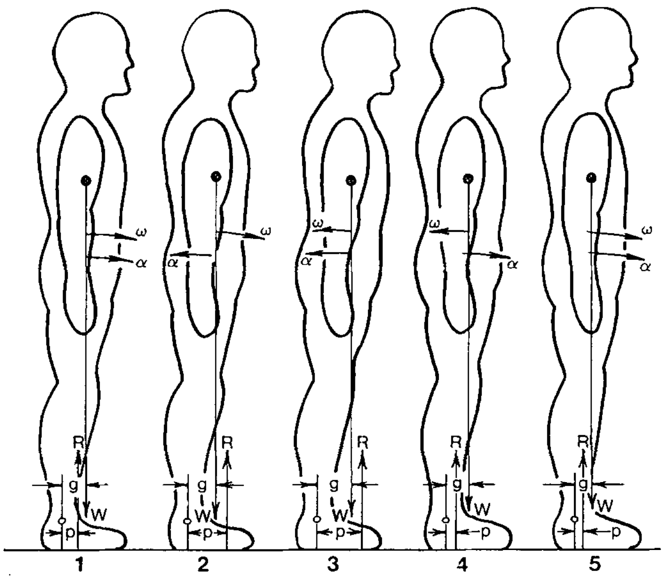


Figure 1. A subject swaying back and forth while standing quietly on a force platform. Five different points in time are described, showing the centre of gravity (g) and the centre of pressure (p) locations along with the associated angular accelerations (α) and angular velocities (ω). See text for detailed description.

If Wg > Rp, the body will experience a clockwise angular acceleration. In order to correct this forward 'sway', the subject will increase his or her COP (by increasing plantarflexion activation) such that at a time 2 of the COP will be anterior to the COG. Now Rp > Wg. Thus α will reverse and will start to decrease ω until, at time 3, the time integral of α will result in a reversal of ω. Now both ω and α are counterclockwise and the body is experiencing a backward sway. When the CNS senses that this posterior shift of the COG needs correcting, the COP decreases (by decreased plantarflexor activation) until it lies posterior to the COG. Thus α will reverse to become clockwise again at time 4, and after a period of time ω will again decrease and reverse, and the body will return to the original conditions, as seen for time 5. From this sequence of COG and COP conditions it can be seen that the plantarflexors–dorsiflexors in controlling the net ankle moment can regulate the body's COG. However, it is apparent that the dynamic range of the COP must be somewhat greater than that of the COG: the COP must be continuously moving anteriorly and posteriorly with respect to the COG. Thus if the COG were allowed to move within a few centimetres of the toes, it is possible that a corrective movement of the COP to the extremes of the toes would not be adequate to reverse ω. Here the subject would have to move a limb forward to arrest the forward fall.

Figure 2 is a record of COP versus COG as a subject stood quietly on a force platform with instructions to stand as still as possible. The preceding sequence of events is repeated many times during this data collection period. It should be noted that over an extended period of time the average COP must equal the average COG. Researchers are often estimating the location of

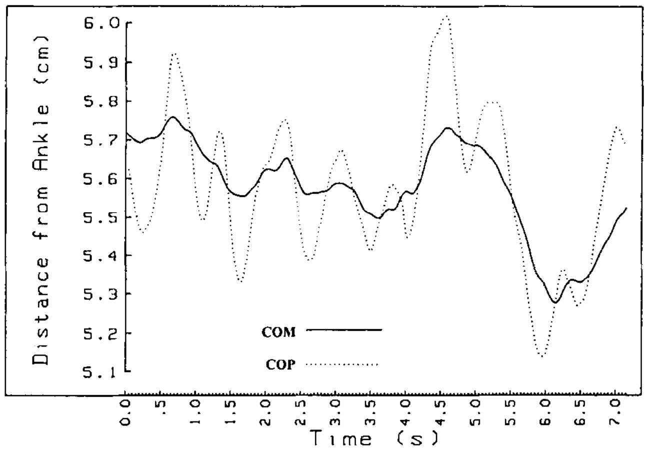


Figure 2. A 7-s record showing simultaneous centre-of-gravity and centre-of-pressure fluctuations for a subject in quiet stance. Centre-of-pressure excursions oscillate either side of the centre of gravity and have a higher frequency and greater amplitude.

the total body COM and tracking its trajectory over the course of time. In an inverted pendulum we can estimate the horizontal linear acceleration ($\ddot{x}$)of the COM from the relationship:

$$\alpha = \frac{\ddot{x}}{d}$$

where: d is the distance from the ankle joints to the total body COM.

We know from Eq 1 that

$$Rp - Wg = I\alpha$$

$$Rp - Wg = I\frac{\ddot{x}}{d}$$

$$But\ R = W,\ \therefore W(p-g) = I\frac{\ddot{x}}{d}$$

$$p - g = \frac{I\ddot{x}}{Wd} = K\ddot{x} \qquad \text{(Eq. 2)}$$

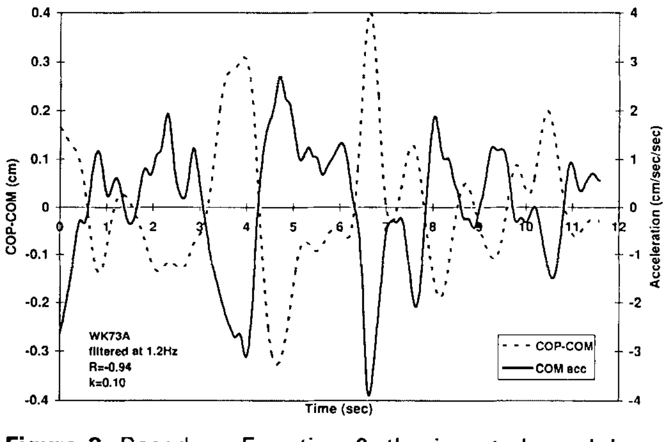


Figure 3. Based on Equation 2, the inverted pendulum model predicts a high correlation between COP-COM and the horizontal acceleration of COM in the A/P direction. In quiet standing the correlation for this subject while standing quietly was 0.94. In large voluntary sways correlations exceed 0.99, giving credence to the validity of the inverted pendulum model in all standing situations.

Thus the difference between the COP and COM is proportional to the horizontal acceleration of the COM. We can think of this difference as being the 'error' signal in the balance control system which is causing the COM's horizontal acceleration. The horizontal acceleration described here is in the A/P direction. The same applies to the M/L acceleration; however, we will see shortly that the biomechanical model of balance changes somewhat. Figure 3 demonstrates the fundamental relationship given by Equation 2. Here we see a 12-s record of a subject standing quietly while COP_{net} and COM were measured. COP-COM was then plotted against the COM acceleration in the A/P direction. As can be seen there is a very high negative correlation between COP-COM and the acceleration. This means that when the COP is ahead of the COM the acceleration is backward, and vice versa when the COP is behind the COM. A similar correlation is evident when the same variables are plotted in the M/L direction.

A more general model of balance during quiet standing

The balance control described in Figure 1 is a special case of a more general approach. Because the side-by-side position has both ankle joint along the same axis we can use a planar (2D) inverted pendulum model. The net result of the neuromuscular control was COP_{net}, which reflects the combined control of both left and right dorsiflexors/plantarflexors. We do not know whether the left and right foot control is symmetrical and in unilateral pathologies (stroke, amputees, etc.) and even in normals there is a dominant limb control. Not until we use two force platforms do we see this independent left and right control at the ankles, and we also see a totally separate control of balance in the M/L direction[9].

During double limb support COP_{net} in either the A/P or M/L directions is calculated as follows:

$$COP_{net} = COP_l \frac{R_{vl}}{R_{vl} + R_{vr}} + COP_r \frac{R_{vr}}{R_{vl} + R_{vr}} \qquad \text{(Eq. 3)}$$

where: COP_l and COP_r are the COP's under left and right feet respectively. R_{vl} and R_{vr} are the vertical reaction forces under left and right foot respectively.

Equation 3 focuses on the fact that COP_{net} in either direction is under the control of four time-varying variables, each of which is under the control of a different set of muscles. In the M/L direction COP_l is controlled by the left ankle invertors/evertors, COP_r is under the control of the right ankle invertors/evertors. R_{vr} and R_{vl} are the loadings under each foot and when expressed as a time-varying fraction of total body weight, $R_{vl} + R_{vr}$, they represent the dynamic load changes under each foot. The biomechanics of these changes has shown that R_{vl} and R_{vr} change completely out of phase and that an increased load on one limb is marked by an equal and opposite unload on the opposite limb. For example the right hip abductors could become more active and increase the right limb load from 49 to 52%; this would result in an instantaneous unloading of the left limb from 52 to 49%. Biomechanically this same change could have been achieved by increase activity in the left hip adductors.

If we wish to partition the changes of COP_{net} into contributions from the ankle muscles or hip muscles the following procedures are followed. If we assume that the hip abd/adductors do not act to load or unload the two limbs we can assume that:

$$\frac{R_{vl}}{R_{vl} + R_{vr}} = \frac{R_{vr}}{R_{vl} + R_{vr}} = 0.5$$

Thus the load/unload mechanism is not operational; therefore the only control will be from the ankle muscles, and this is labelled COP_c where:

$$COP_c = COP_l \times 0.5 + COP_r \times 0.5 \qquad \text{(Eq. 4)}$$

If the two limbs do not load perfectly symmetrically then the percentage load taken by each limb should be set to the average value over the balance trial. For example, $R_{vl}/(R_{vl} + R_{vr})$ could average 0.47, then $R_{vr}/(R_{vl} + R_{vr})$ would be 0.53. We know that COP_{net} is the sum of two separate mechanisms. Thus COP_v can be calculated from:

$$COP_v = COP_{net} - COP_c \qquad \text{(Eq. 5)}$$

where: COP_v is the contribution due to the loading/unloading of each limb.

In side-by-side standing it has been shown that COP_c and COP_v are virtually independent of each other[9]. In the A/P direction COP_l and COP_r act almost perfectly in phase such that COP_{net} lies almost in the middle. Figure

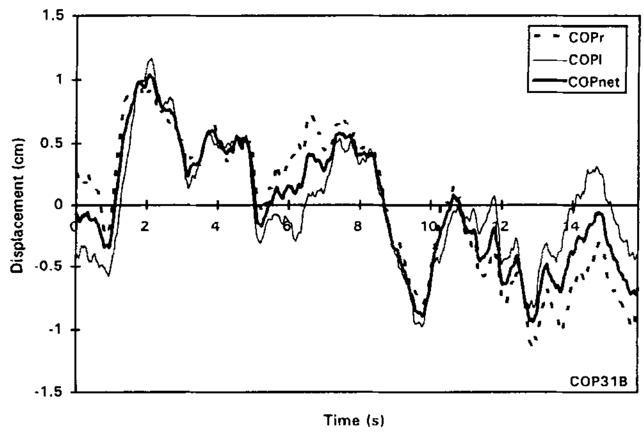


Figure 4. COP_r, COP_l and COP_{net} in A/P direction for a subject standing quietly. Because the weight carried by each limb is about 50% the COP_{net} is approximately the average of COP_r and COP_l. In unilateral pathologies the intact limb will dominate over the affected limb.

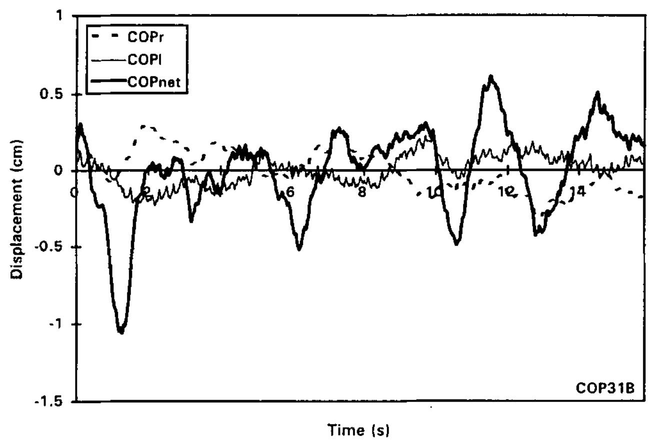


Figure 5. COP_r, COP_l and COP_{net} in M/L direction for the same subject as in Figure 4. COP_l and COP_r oscillate in opposite directions and effectively cancel out and do not appear to be correlated in any way with COP_{net}. This alerts us to the fact that (from Equation 3) the M/L control of COP_{net} must be under the control of the loading/unloading of each limb.

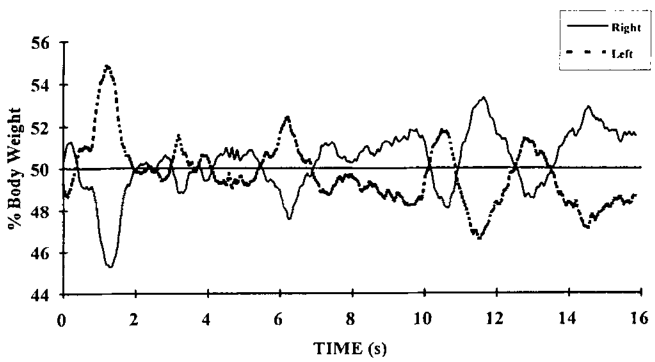


Figure 6. Right and left vertical ground reaction forces for the same subject as in Figures 4 and 5. These forces fluctuate several percent about 50% and are completely out of phase, which indicates that one limb unloads instantaneously as the other loads. Mechanically, this means a tight coupling at each of the joints. Note the shape of these waveforms are the same as COP_{net} in Figure 5, indicating the fact that a load/unload mechanism by the hip abductors/adductors is controlling COP_{net} in the M/L direction.

4 demonstrates this synchronization. However, in the M/L direction there is virtually no collaboration between the ankle muscles. Figure 5 shows that the medial/lateral changes in COP_r and COP_l are almost completely out of phase and that the M/L COP_{net} changes bear no relationship to either COP_r or COP_l. Thus it is evident from Equation 1 that the dominant control in the M/L direction is due to the load/unloading mechanism and not due to the ankle muscle control of COP_r and COP_l. Figure 6 demonstrates the vertical reaction force fluctuations for the same subject whose M/L COP data were presented in Figure 5.

Computer simulations of A/P and M/L control can be conducted using simple biomechanical models as shown in Figure 7. The A/P model has the COM of the lower limbs as shown with ankle and hip joints as possible control sites. The M/L model has similar COM locations with two hip and two ankle joints as potential control sites. Therefore there are four torque motors that can control the parallelogram defined by the two ankles and two hip joints. To cause the COP_{net} to move to the right any one of the following muscle activations could be the cause: left ankle evertors, right ankle invertors, left hip adductors, right hip abductors. The movement of COP_{net} to the right would cause a lateral acceleration of the COM to the left.

Theoretically, the same magnitude of torque acting at any one of these four joints would have the same results. However, due to the biomechanics and anatomy of the ankle and hip joints this theoretical situation never occurs. First, the ankle invertors and evertors cannot act independently. All the evertors (peroneii) and all the invertors (tibialis anterior and posterior, extensor digitorum longus, and hallucis longus) are also plantarflexors and dorsiflexors. The A/P control of balance requires collaboration between right and left plantarflexors and dorsiflexors. The COP under each foot will move back and forth in almost complete synchronization; however, any M/L component by the invertors/evertors will move the COP under both feet in the same medial or lateral direction. For example, if both left and right peroneii are active to cause the COPs to move anteriorly they will also cause the both COPs to move medially. Figure 8 shows these individual trajectories as recorded by separate force plates, and this cancellation of M/L COPs was evident in Figure 5. There is a second reason that the invertors/evertors could not act when large balancing moments were

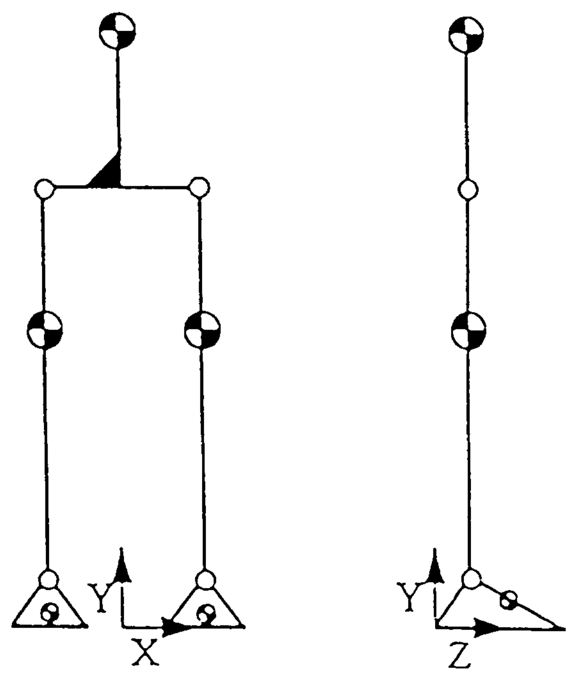


Figure 7. Inverted pendulum biomechanical models for standing. In the frontal plane the pendulum is a parallelogram pivoting about both ankle and both hip joints, and is under the control of four sets of muscles. In the sagittal plane the pendulum pivots about the ankle in quiet stance or about both hip and ankle in perturbed standing.

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

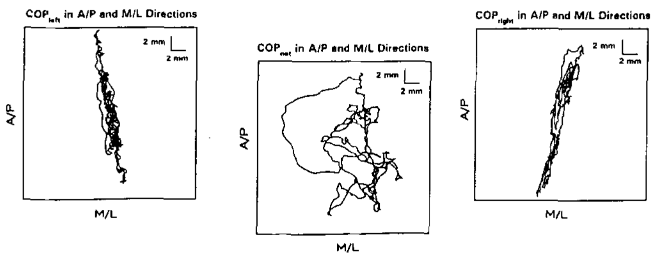


Figure 8. COP trajectories under individual feet and COP_{net} (as would be recorded from a single force platform). COP_l and COP_r move forwards and backwards in synchronization but in M/L direction they move out of phase. Thus any movement of COP_{net} in the M/L direction must not come from COP_l and COP_r but from the load/unload fractions of body weight, $R_{vl}(t)/(R_{vl}(t) + R_{vr}(t))$ and $R_{vr}(t)/(R_{vl}(t) + R_{vr}(t))$. These fractions of body weight are under the control of the hip abductors/adductors.

required. Because of the small width of the foot the maximum moment that could be generated by either invertors or evertors would be about 10 N m. Anything above 10 N m would cause the foot to roll over on its medial or lateral borders. However, the hip abductors/adductors are not so constrained. Each of these muscle groups could generate in excess of 100 N m in emergencies. To summarize this ground load/unload mechanism, Figure 9 is presented. Here we see the out-of-phase vertical reaction forces under the left and right feet along with the M/L COP_{net}. As can be seen the COP_{net} is virtually in phase with the right limb force and out-of-phase with the left. The fluctuations in the vertical reaction forces are due to hip abductor/adductor activity, and a 5% fluctuation in vertical force causes about 1 cm shift in COP_{net}.

Ankle and hip strategies

In the A/P direction both an ankle and a hip strategy have been described[10]. The ankle strategy applies in quiet stance and during small perturbations and predicts that the ankle plantarflexors/dorsiflexors alone act to control the inverted pendulum. In more perturbed situations or when the ankle muscles cannot act a hip strategy would respond to flex the hip, thus moving COM posteriorly, or to extend the hip to move the COM anteriorly. A computer simulation of each of these strategies is presented in Figure 10. Here, a 10-N m ankle moment was applied for 300 ms and the COM of the lower limbs and the upper body were estimated. The total body COM displacement (posterior) was estimated to be 1.56 cm. The same 10 N m was applied as hip flexors to simulate a hip strategy and the posterior displacement of the COM was 2.04 cm. However, a combined ankle and hip strategy was quite possible and with a 10-N m plantarflexor moment plus a 10-N m hip flexor moment the COM displaced 3.53 cm after 300 ms. The experimental protocol described by Horak and Nashner[10] had the subjects standing with both feet side-by-side across a narrow beam such that the plantarflexors could not act. In this protocol a hip strategy was the only option open to the subjects. More experimental work is required to investigate the selection of either or both of these strategies when both feet are firmly planted on the ground. More will be presented on this topic in the section on reactive and proactive control of perturbations to the balance system.

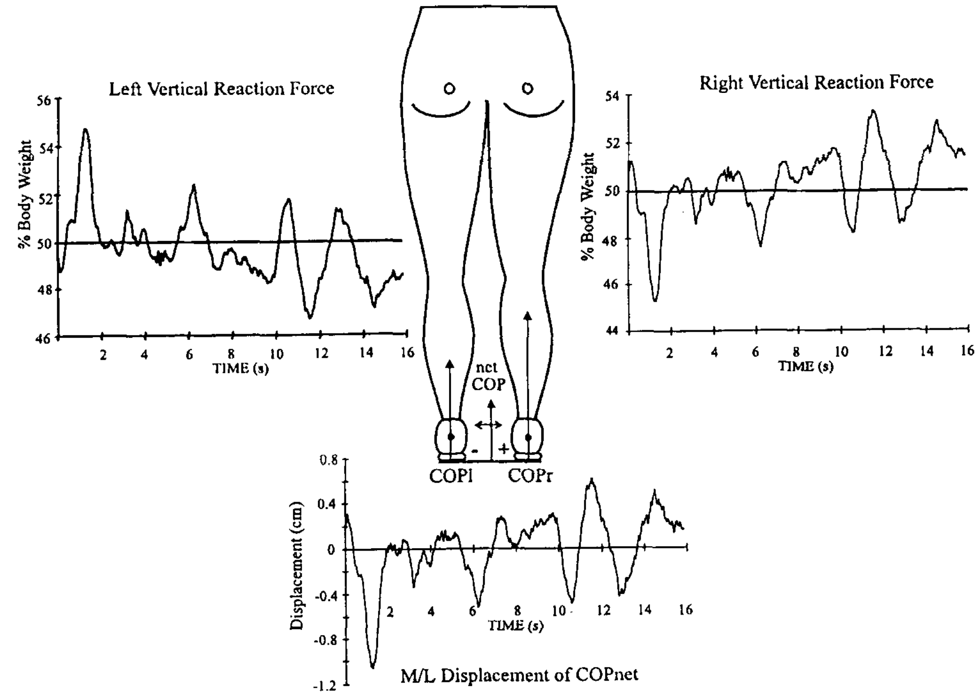


Figure 9. M/L displacement of COP_{net} versus R_{vl} and R_{vr} for the same subject as in Figure 8 showing how the load/unload *fractions control this M/L displacement.*

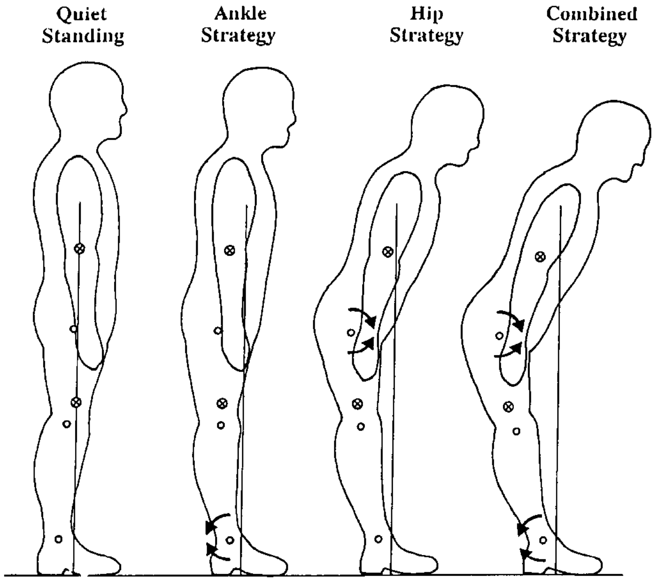


Figure 10. Displacement of the COM of the body as a result of a 10 N m applied for 300 ms at the ankle (ankle strategy), at the hip (hip strategy), and at both the ankle and hip (combined strategy).

Centre of pressure analyses, quiet stance

The vast majority of research in quiet standing has used the COP from a single force platform as the outcome measure. There is considerable confusion regarding the interpretation of this COP signal. A majority of researchers have referred to the COP as 'postural sway' (cf Diener et al.[11]). Others[6,12] have described their COP recordings as COG, while one group[12] equated their COP with independently calculated COG measures from film data. These research reports have ignored much earlier work by Murray et al.[13] who not only showed a distinct difference between the COP and COG trajectories but also suggested that the servo control signal was the difference between COP and COG. A similar critique was made by Geursen et al.[14], recognizing in an inverted pendulum model that the difference between COP-COM should be proportioned to the horizontal acceleration of the COM. However, confusion was added by one group who claimed that the displacement of the COP was caused by a shift in the body's COG![15] Murray et al.[13] and Prieto et al.[16] observed quite correctly that the excursions of the COP were always greater than the COG and that the COP signal oscillated either side of the COG at a higher frequency. Thus in our review of postural studies we will correctly interpret that the COG excursions are smaller than the reported COP fluctuations in both the A/P and M/L directions.

In all studies of normal subjects, when the subjects stood with their eyes closed[6,11,17–19] the COP amplitude was higher in both A/P and M/L directions. In single support not only were the COP amplitudes higher for eyes closed but also the fluctuations in the three ground reaction forces were significantly larger[20]. Other measures of COP displacement were path length and area. Path is the length of the COP displacement trajectory and is therefore independent of direction[11,15,19] and infers that the human control system is independent of the direction of the COP trajectory. Collins and DeLuca[21] have analysed the random path of the COP trajectory using statistical mechanics, but the assumed random walks inferred that the control system was the same for the A/P as the M/L direction. From the general model of balance we saw that for side-by-side standing the M/L neuromuscular control is a hip load/unload mechanism while A/P control was independent and controlled at the ankle. In other standard positions, such as the Romberg position (one foot directly in front of the other) there were differences in the M/L versus A/P fluctuations. In the Romberg position the COP amplitude was significantly greater than in the side-by-side position[6,17]. Goldie et al.[22] in a large-scale reliability study of four different stance positions (side-by-side, step, tandem, one legged) report that the three reaction forces were more discriminating than the COP measures. Kirby et al.[23] reported M/L and A/P changes in COP in 14 different stance positions (variable lateral spacings, variable A/P spacings, and variable internal/external foot rotations). The greatest M/L COP changes were with the feet together in the side-by-side position. Placing one foot 30 cm ahead of the other with the feet 15 cm apart resulted in increased A/P and M/L changes. Internally rotating the feet to 25° and 45° also increased both M/L and A/P COP changes.

In studies of the elderly there have been reports of increased amplitude of COP[4,18,24–27] and higher frequency content of the COP signal[18,27]. An excellent review of these standing postural measures and other skeletal muscular and neural measures in the elderly has been presented by Vandervoort et al.[28]. The question arises as to whether increases in the amplitude of the COP under these very safe conditions is necessarily an indication of loss of balance[29], especially when some major pathologies report a decrease in COP amplitude[30], and children have larger excursions similar to the elderly[31].

In balance disorders the magnitude of the COP excursions is drastically increased. Cerebellar patients with five common localizations were compared amongst themselves and with normals[11]. Some COP measures showed significant differences. For example, A/P amplitude and 'sway' path for the eyes closed condition was greater for anterior-lobe patients than the other four localizations. Lucy and Hayes[18] also reported the r.m.s. COP in A/P and M/L directions to be significantly higher for 10 cerebellar patients with a variety of diagnoses.

Centre of mass analyses, quiet stance

As has been indicated in the previous section COP measures from a single force platform have dominated all balance studies. Separate COM and COP measures have been calculated a number of different ways by a small number of researchers[12,14,32–39]. These COM and COP measures were compared in a wide variety of movements including gait, initiation and termination of gait, and quiet and perturbed standing. However, with the exception of the quiet standing trials of Spaepen et al.[12,37] and Panzer and Hallett[36] all researchers recognized the fact that the COM and COP were quite different variables and that the difference between the COP and COM was somehow related to the movement of the COM. In fact, several researchers have predicted that the (COP-COM) signal should in an inverted pendulum model be directly related to the horizontal acceleration of the COM[14,35,39,40]. This relationship for an ideal inverted pendulum was predicted in Equation 2.

Because the (COP-COM) signal is directly related to the horizontal acceleration of the COM it can be considered the error signal that the balance control system is sensing. The magnitude and frequency of this error signal is of importance in the interpretation of the balance control system. The 'gain' of the feedback control system would alter both the magnitude and frequency of the error signal. Classical feedback control theory would describe the inverted pendulum model as being an underdamped system. Thus an increased gain would not only increase the amplitude of the error signal but also the frequency of the oscillations.

One research group[41] has questioned the validity of a inverted pendulum model. Unfortunately their rationale was flawed: they compared the full-wave-rectified shear force with a range of anthropometric measures on their subjects and found no correlation. They gave no mathematical reason that these passive anthropometric measures should be correlated with the shear force magnitude.

Control of balance during perturbed standing

Researchers have long recognized the fact that research into quiet standing is very limited in revealing the mechanisms of balance and as a diagnostic tool to pinpoint deficits of the system. With severe challenges to the balance system researchers can isolate defence strategies and also identify the role of individual sensory systems: vision, vestibular, and somatosensory. Perturbations to the systems have been either external or internal. Internal disturbances result from voluntary movement of the body such as raising the arms or bending the trunk. In these situations the response is proactive and the researcher is interested not only in the nature of the disturbance but also the anticipatory response of the CNS to protect against imbalance. External perturbations are applied without the knowledge of the subject and have the goal of testing the reactive response of the three sensory systems. A wide variety of perturbation systems have been developed: everything from platforms that tilt and translate to levers that push and pull. Again, the researchers not only document the kinematics and kinetics of the disturbance but also record the body's responses. Hopefully the protocol allows them to identify different mechanisms and strategies and even isolate the role of each sensory system. To this end a number of interesting research paradigms have been developed to cause conflict between sensory systems.

One basic question that should be addressed by all researchers is whether a paradigm is representative of the kind of disturbances experienced in real life. For example, a fairly common system developed in many laboratories is the moving platform, which can rapidly translate or tilt. How well does such a perturbation at the ground level simulate the conditions that cause falls? The closest equivalent to a translating platform is a moving bus or subway; how many falls occur in such moving vehicles as opposed to being jostled in shopping centres or while running to catch a bus?

Platform perturbations

The pioneering work by Nashner must form a large part of the introduction to body of knowledge relating to perturbations applied by a moving platform. The platform could translate horizontally, the base could rotate the foot and in some cases the platform displaced vertically. The subjects were tested in the sagittal plane only. Classical stretch responses resulted[42]. When the platform moved backwards, the gastrocnemii and hamstrings had the most common response with latencies of 100–120 ms after the onset of platform translation. This response by the posterior muscles was matched by a common response by the anterior muscles (tibialis anterior and rectus femoris) when the platform was translated forward. However, there were small percentages of exceptions. In 18% of the imposed dorsiflexor trials only the gastrocnemii responded and in 22% of the plantarflexor perturbations all four muscles responded. In a few trials only the gastrocnemii and tibialis anterior responded to a backward translation and the gastrocnemii also responded with the normal tibialis anterior/rectus femoris to a forward translation.

Tilting of the platform produced some interesting variations on the induced dorsiflexion and plantarflexions resulting from linear backward and forward translations of the platform. Tilting the platform upwards stretched the gastrocnemii and a combined short latency gastrocnemii/hamstring response resulted[43]. However, the upward tilt of the platform did not shift the body's COM in the anterior direction (as did the backward translation), thus this response caused a backward sway of the COM that required a second response (185–250 ms) by the anterior muscles (tibialis anterior, rectus femoris) to prevent loss of balance. Further analysis of the initial responses[43] revealed that the posterior anterior group of muscles turned on in a specific order. With additional lumbar muscles (erector spinae and abdominals) recorded there was a bottom-up sequence in the latencies. For a backward translation of the platform the gastrocnemii responded first, followed by the hamstrings, then the erector spinae. For a forward platform movement this sequence was tibialis anterior–rectus femoris–abdominals. Thus it appears the CNS recognizes the need to stabilize the joint closest to perturbation first, followed by the knee, hip, and spine.

Because the responses radiated from the ankle towards the body's COM, those responses were described as an 'ankle strategy'.

An alternate strategy, called a hip strategy was identified when the ankle muscles were unable to respond (because the foot was positioned sideways on a narrow beam)[10]. When the platform was displaced in a backward direction the CNS responded with a strong hip flexor (abdominal + rectus femoris) pattern. As was seen in the section on Ankle and hip strategies a hip flexor moment is more effective in shifting the body COM than an ankle strategy. The reverse response by the extensors (hamstrings + erector spinae) was evident when the platform was translated forward.

Higher-level responses to platform displacements in the A/P direction were reported by McIlroy and Maki[44]. Three patterns of response were noted: both left and right foot responded with a symmetrical load and unload but with no stepping; loading of one limb and unloading by the other with no stepping; complete unloading of one limb with a synchronized loading of the other followed by a short step by the unloaded limb. Whether a hip strategy was also involved in any of these patterns was not reported.

Responses to horizontal translations of the platform over 360° were reported by Moore et al.[45]. Polar plots of eight muscle groups documented the amplitude of the response versus the direction of perturbation. A/P perturbations were essentially the same as reported by others. However, perturbation with lateral components involved hip abductors and adductors. They referred to this response as 'movement about the hip' but did not interpret the abd/adductor muscle activity as one of loading one limb and unloading the other as would have been predicted from the section on a General model of balance.

Arm perturbations

A second common site of perturbation is via a handle pull or push on the arms. Cordo and Nashner[46] demonstrated latencies of 30 ms in the biceps muscles followed by 50 ms in the gastrocnemii when unexpected pulls were applied to the arms holding a handle in front of the body. When the body was stabilized by a frame to prevent a forward fall the gastrocnemii (and presumably other non-recorded posterior muscles of the spine and lower limb) remained silent. Frank and Earl[47] reported COP response changes (in the forward direction) due to an unexpected forward pull. Unexpected pushes showed the opposite COP change in the backward direction. Latencies of about 40 ms were seen in those sudden COP changes that directly reflect plantarflexor or dorsiflexor motor responses.

Leg perturbations

A theoretical model of perturbations to quiet standing and experimental validation has been presented by Yang et al.[48]. Her model was a sagittal-plane simulation where forces could be applied at any joint of the lower limb or to the trunk segment. The simulated impulse force was applied for 80 ms and the model was asked to vary the response moments at the ankle, knee, and hip to determine if a safe recovery of balance was achieved. A safe recovery was declared if all three segments (leg, thigh, HAT) returned to within 5° of their initial conditions. A 'solution space' was determined and it demonstrated that various combinations of ankle, hip and knee moments were successful. This 3D 'solution space' demonstrated that many combinations of hip, knee and ankle extensor moments were capable of returning the body to near-normal posture. The solution space was also sensitive to the magnitude of the perturbation force (80 N, 160 N, 320 N). Experimental results on a small number of subjects demonstrated that their results fell within the solution space. EMG responses from several subjects also confirmed different combinations of responses. Some subjects defended primarily with ankle plantarflexors, some with quadriceps, and a small number with hip extensors. These theoretical and experimental results demonstrate the need to look at the total limb (and sometimes the total body) for the net response to a perturbation. As this perturbation was applied to the knee joint, the system's response was somewhat similar to that seen during stance phase of walking: the total limb defence against collapse has been described as a support moment[49]. The support moment was the sum of the ankle, knee and hip moments and was seen to be quite consistent within subjects. The consistency of the support moment results largely from the ability of the CNS to trade-off between the hip and knee extensors[50].

Shoulder perturbations

A single study of perturbations to the trunk has been reported[51]. Impulsive forces of abut 200 N in an anterior direction were applied to the upper trunk. Latency responses of the posterior leg muscles (Sol, MG, MH, LH) were 45–60 ms; these latencies were less than those reported by Nashner[42] for platform perturbations. Also, they did not show the patterning radiating from ankle upwards as reported by Nashner[42]; the medial hamstrings in this study turned on before the soleus and gastrocnemii, indicating a top-down order. The net effect of the muscle responses was evident in the kinetics: 80–100 ms after the perturbation, the response from the hip extensors, knee flexors, and ankle plantarflexors was evident. Also, the small and delayed angular changes at the hip, knee, and ankle supports the conclusion that the neural responses were not stretch reflexes; rather, they were higher-level responses, probably triggered by receptors at or near the site of the perturbation.

Responses to perturbations involving sensory conflict

Sensory conflict occurs fairly often in daily living. The visual system often tricks us when we observe the movement of one vehicle relative to another. When we observe an adjacent bus or train from our own vehicle we often perceive that we are moving rather than the adjacent vehicle. Or in the movement of the deck of a boat we can get conflicting information especially if we are below decks and cannot see the horizon as a reference. Researchers have been very innovative in developing equipment and protocols to achieve conflict among the three sensory systems or even within the same sensory system.

Lackner and Levine[52] applied a vibrator to the ankle muscles of subjects as they stood quietly. The vibrator stimulates the muscle spindles in the same way as if they were being stretched. For example, if the soleus/gastrocnemii were vibrated the CNS senses that these muscles were being stretched (i.e. the subject is falling forward). The CNS 'corrects' by activating these same plantarflexors, which results in a false pull backwards. The reverse would be true if the tibialis anterior were vibrated.

Nashner[43] caused a visual surround to rotate in the same direction as the body swayed when the platform was displaced. For example, if the backward displacement were matched with an angular rotation of the visual surround that matches the induced forward lean the subject responds differently to that described in the section on Balance and posture in human gait. The initial stretch reflex response ($\approx$ 120 ms) of the gastrocnemii and hamstrings was evident but at a much reduced level. The visual system had been temporarily 'tricked' into thinking that the body was still erect. Then, when the conflict was sorted out, the gastrocnemii came in with a second response about 250 ms later.

Adaptations to false responses are seen to be evident after repeat perturbations. For example, Nashner[43] reports that an upward tilt of the platform produced an ankle stretch response which gave the CNS the false impression that the body was falling forward. A second response to the same perturbation was reduced and by the fourth tilt of the platform the gastrocnemii did not respond. The gastrocnemii have now adapted not to respond when they were stretched. However, when these tilt perturbations were immediately followed by a backward translation of the platform the gastrocnemii remained silent. Then on the 2nd, 3rd, and 4th translations the gastrocnemii increased to its appropriate level of response. Thus the adaptations to the original inappropriate responses had to be reversed because they were inappropriate for the second perturbation.

Conflicting visual input can result in some serious and potentially dangerous responses. Lee and Young[53] describe a series of experiments with adults and children standing inside a moveable room. The room is on wheels and can be translated towards or away from the subjects. The response is to move in the same direction as the walls, and it can vary from a small sway, a corrective step or, in the case of young children, a fall. The same room when tilted sideways as the subject was walking caused the subject to sway laterally in the same direction as the room.

Redundancy of the sensory systems

Because we have three separate sensory systems that control balance it appears that a certain degree of redundancy exists which can be put to use when one or more of the systems fails or is temporarily lost. A classical experiment[54] was conducted with normal subjects, plus one group with somatosensory loss and another with vestibular loss. Six different combinations of perturbations were applied. (i) normal vision, fixed support – all three groups had low sway; (ii) absent vision, fixed support – all three groups had low sway; (iii) visual surround swayed forward, fixed support – all three groups had low sway; (iv) normal vision, platform tilted forward – all three groups had low sway; (v) absent vision, platform tilted forward – vestibular loss group had large sway; (vi) visual surround swayed forward, platform tilted forward – vestibular loss group had large sway. The normal control group always had at least one redundant system available to substitute for the lost input. For example, in conditions (v) and (vi) vision and proprioception was lost or perturbed but the subject's vestibular responses were intact. The same applies for the somatosensory loss patients; they were perturbed by a tilting platform in both (v) and (vi) but they were not aware of the ankle perturbation but as with the normal group they still had their vestibular system intact. In condition (iv) the vestibular loss group maintained their balance with their visual system, and in condition (iii) they maintained balance with their somatosensory system.

Proactive balance responses: internal perturbations

The most common perturbations that occur daily during standing are internal. They are voluntarily initiated and arise from different orientations of the body because of turning, reaching, bending. The goals of research of these responses is twofold: how does the perturbation alter balance of the body, and how does the CNS anticipate with motor programmes in advance of the in balancing event.

Cordo and Nashner[46] had subjects pull a handle against a load. They recorded from the focal muscles (biceps) and from lower limb balance muscles (gastrocnemii + hamstrings). Based on latencies referenced to the activation of the focal muscles it was reported that the gastrocnemii and hamstrings were activated about 60 ms earlier. Thus the CNS stabilized the postural system in advance of the anticipated arm perturbation. Similar results for voluntary pushes and pulls by the arms[47] were evident from the anticipatory COP changes recorded with force platform. Bouisset and Zattara[55] recorded reaction forces and moments from a force platform while subjects raised both arms to the horizontal or raised a single arm to the horizontal. An accelerometer mounted on the wrists indicated the start of the focal movement. In all cases the reaction force and moment changes preceded the onset of the arm movement. For the bilateral arm movement the responses were confined to the sagittal plane (forward and upward forces). However, for unilateral arm raises the responses indicated a stabilization of the body around the vertical axis plus a lateral stabilization in the frontal plane.

Arm movement perturbations have been limited until recently to arm raises, and the response was claimed to be a compensation for the forward displacement of the COM of the arms[55,56]. No generalizations were tested to demonstrate, for example, that arm lowering from above to ahead had a similar compensatory response. A test of all four possible combinations of arm movement was conducted by Eng et al.[57]: FH – arm flexion to horizontal, EH – arm extension to horizontal, FV – arm flexion to vertical, and EV – arm extension to vertical. The response of the postural muscles were either anterior (ankle dorsiflexors, knee extensors, hip flexors) or posterior (ankle plantarflexors, knee flexors, hip extensors) depending on the polarity of the shoulder moment. A flexor shoulder moment (FH or FV) resulted in a posterior balance response while an extensor shoulder moment (EH or EV) resulted in an anterior balance response. These results were contrary to what would be predicted from the arm COM displacement; a posterior response would have been predicted for FH and EH, and an anterior response for FV and EV. The explanation is that the perturbation is a shoulder moment which must be countered in order to keep the trunk erect. A flexor moment acting on the upper arm generates an equal and opposite extensor moment on the upper trunk, which if not countered would cause the trunk to rotate forward. Only a hip extensor moment acting at the distal end of the trunk will stabilize the trunk, which in turn must be countered by a knee flexor moment (to keep the thigh vertical), and also an ankle plantarflexor moment (to keep the leg vertical).

Stepping, initiation and termination of gait

The task in standing is to keep the body's COM safely within the base of support. However, when we wish to move the body over the ground just one step or more the criterion of balance are drastically altered. The goal is now to move the body outside the base of support and yet prevent falling. In steady-state walking or running the COM is always outside the base of support (except during the short double support period in walking). The state of balance has been described as dynamic balance, which means that the swing limb has a trajectory which will achieve balance conditions during the next stance phase. The next section starts to address those situations as they develop during single steps or during gait initiation, and how stable balance is re-established during termination of gait.

Gait initiation

The pioneering work on gait initiation was reported by Herman et al.[58], where force platform, joint kinematics, and EMG data described the components of initiation. The vertical reaction forces from each force platform showed the sequence to loading and unloading of each limb.

Initially the swing limb loaded above its resting 50% BW while the BW while the stance limb unloaded by the same amount. Then there was a rapid unload of the swing limb with an equal loading of the stance limb. This sequence of events is the same as we see in current reports except that there was no comment or explanation in this earlier work. The kinematics were limited to joint angles so it was not possible to interpret this research in terms of the COM relative to the base of support. Thus the mode of unbalancing the body was limited to what was reported in the EMG of the TA, SOL, and MG. Prior to initiation the gastrocnemii and soleus were active to hold the COM in some equilibrium position anterior of the ankle joint. At initiation these muscles turned off to allow the inverted pendulum to accelerate forward, followed by a drastic increase in the tibialis anterior activity to pull the inverted pendulum forward. Then, when the body had sufficient forward lean and velocity, the stance limb plantarflexors became active again to achieve a forceful push-off.

Mann et al.[59] reported force platform measures of the initiation of gait of 10 subjects. Their COP recordings demonstrated an initial posterior and lateral shift towards the swing limb, then a rapid shift towards the stance limb as the swing limb unloaded. The shear forces in the A/P and M/L direction were reported; however, they equated these forces with the moment of the COM; in fact they stated that the 'center of pressure represents the projection of the center of mass . . . '. Although their data were perfectly correct their interpretation of the biomechanics of the inverted pendulum was seriously in error. They attributed the initial COP movement towards the swing limb as being due to peroneii activity of that limb. This is not correct because peroneii activity would cause the COP to move forward and medially. In spite of distinct swing limb hip abductor activity during the initial release phase, they did not comment on the role of those muscles to load one limb and unload the other. Their final conclusion admits that they were not completely clear 'why the initial transfer of some weight to the limb that is to be the swing limb'. Their data were correct but their interpretation was flawed.

Brenière et al.[60–62] using force platform data to advantage were able to predict what the COM was doing during this transition from standing to stepping. They recognized that the shear forces divided by body mass were equal to the A/P and M/L accelerations of the COM. A second integral of those accelerations yielded the trajectory of the COM which could be interpreted relative to the COP_{net} in either the A/P or the M/L directions. During the very initial phase of initiation the COP was seen to move posteriorly as the COM accelerated forward. In the M/L direction the COP was seen to shift initially towards the swing limb, then rapidly across to the stance limb. Because they did not record from two force plates they were not able to observe that this critical shifting of the COP resulted from the loading of the swing limb and unloading of the stance limb and therefore was responsible for the initial acceleration of the COM towards the stance limb. Figure 11 shows this sequence of events. X_G and Y_G are the A/P and M/L trajectories of the COM respectively, X_P and Y_P are the associated COP trajectories. $\ddot{X}_G$ and $\ddot{Y}_G$ are the A/P and M/L accelerations of the COM respectively, t_O is the time of onset of initiation, T_{HO} is the time of heel off of swing foot, t_V is the time of peak horizontal velocity at the end of the first step.

An EMG study combined with kinematic (cinematography) measures (Crenna and Frigo[63]) extended that reported by Herman et al.[58]. On initiation the stance limb soleus was seen to deactivate and to be followed by tibialis anterior activity. Then as the body falls further forward large soleus activity creates the first active push-off. They examined the amplitudes of the deactivation of the soleus and activation of the tibialis anterior and correlated them with the posterior shift of the COP, and concluded that the tibialis anterior was the muscle more responsible for the shift. However, it is impossible to relate EMG amplitude changes in antagonist pairs without modelling the EMG to moment relationship of each muscle, including their twitch characteristics. Also several other dorsiflexors and plantarflexors are involved in the COP shift that was recorded.

Jian et al.[40] reported on the combined COM and COP trajectories during both initiation and termination of gait. Five repeat trials of four subjects were reported. Three force platforms and four videocameras permitted complete 3D kinematic and kinetic analyses. An assessment of the COM versus COP trajectories (Figure 12) showed the COP to move posteriorly and towards the swing limb during the release phase. Inverted pendulum theory would predict that this shift would cause the COM to accelerate in the opposite direction, and this is precisely what was observed; the COM trajectory was forward and towards the stance limb. The kinetics causing this COP shift were very distinct. The posterior movement of the COP resulted from a momentary decrease in plantarflexor activity (aided partially by an increase in dorsiflexor activity). The lateral movement of the COP towards the swing limb was due to an increased loading of that limb by its hip abductors. The release phase was followed by a rapid unload phase. Here the stance limb loads as the swing limb unloads, accomplished by decreased abductor activity of the swing hip and increased abductor activity of the stance hip. This causes a rapid COP shift across to the stance limb, which by the time of toe-off of the swing limb is carrying the total weight of the body. The body is falling forward at this time and as the stance limb plantarflexors increase activity, causing the COP to move forward under the foot. The line joining the COP to the COM is the acceleration vector and after the unload phase this vector is directed forwards and away from the stance foot and forwards towards the future position of the swing foot. By the time the first heel contact is achieved the forward velocity of the COM is about 85% of final steady state. As will be seen later the trajec-

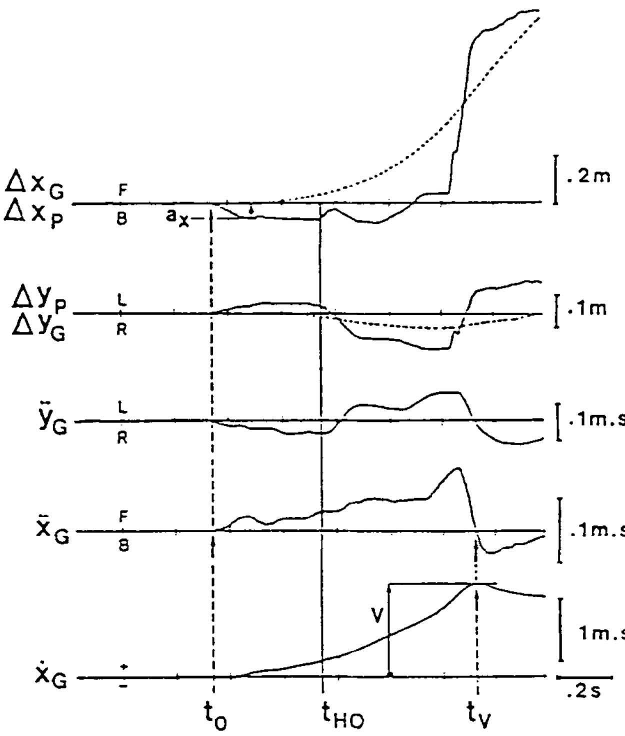


Figure 11. COP records and COM estimates in the A/P and M/L directions during initiation of walking as determined from force plate data alone. Reproduced with permission from *J Mot Behav* 1987; 19: 62–76.

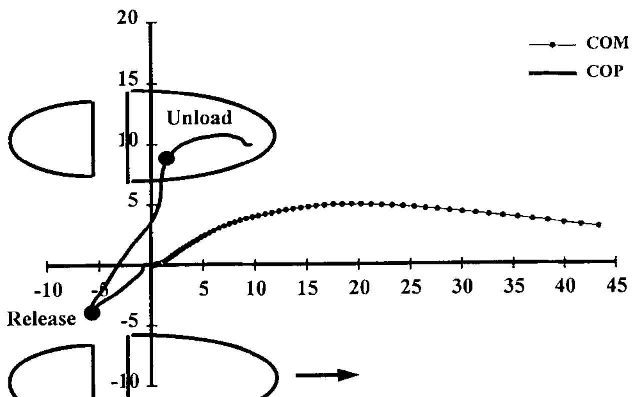


Figure 12. COM and COP trajectories during initiation. During the release phase the COP moves posteriorly and towards the swing limb, thus accelerating the COM forward and towards the stance limb. Posterior COP displacement results from a deactivation of the plantarflexors and, in some cases, an activation of the dorsiflexors. The lateral displacement of the COP results from a momentary loading of the swing limb (right) by the hip abductors. Unloading is achieved by a rapid activation of the stance limb (left) hip abductors and deactivation of the right hip abductors. After unloading of the right limb the COP under the stance moves forwards under the control of the plantarflexors. During this single support time the COM now accelerates forward and away from the stance limb.

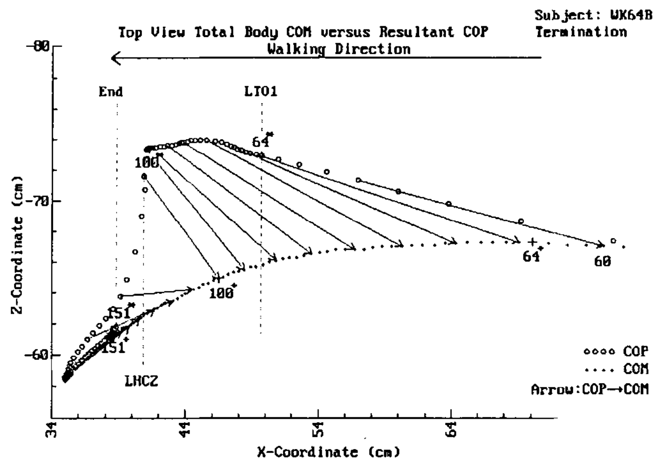


Figure 13. COM and COP trajectories during termination of gait. See text for details. Reproduced with permission from *Gait and Posture.* 1993; 1: 9–22.

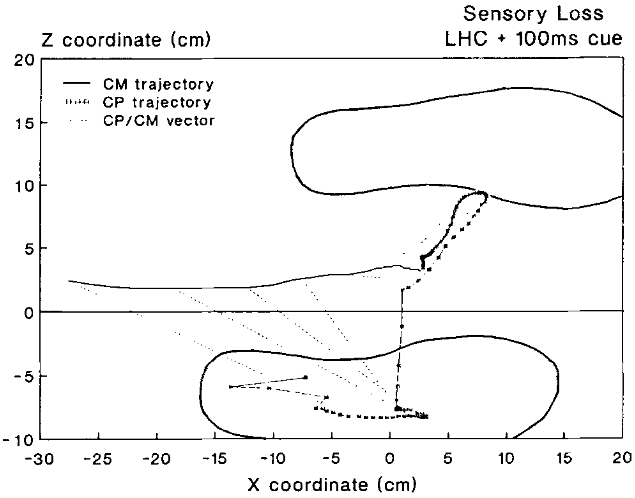


Figure 14. COM and COP trajectories during termination of a subject with peripheral sensory loss. Note the overshoot of the COP in the lateral direction, indicating that the loss of somatosensory feedback from the legs and feet has perturbed the motor control system to the extent that it has lost much of its precision in its estimate of the COM trajectory.

tory of the COM is now almost completely as that seen in steady-state gait.

Gait termination

Little research has been reported on termination of gait. Jian et al.[40] in the latter half of their paper reported on the termination of gait of the same subjects and using identical measures as with the initiation trials. The COP and COM trajectories during termination were virtually mirror images of that reported for the initiation trials. Figure 13 is reproduced from Jian et al.40 to show these trajectories.

The command to stop was triggered by the subject's weight bearing on the first of the three force plates with his or her left foot. The subjects were required to stop during the next two steps, when the right and left foot bore weight on the second and third force plates. Slowing down was not evident for most of the stance phase of the left foot; the only evidence of a decreased forward velocity was related to a much reduced push-off of the left foot. After heel contact of the right foot the COP moved rapidly forward due to increased plantarflexor activity. A mechanical power analysis revealed a significant loss of mechanical energy through absorption by those plantarflexors. The rapid forward movement of the COP resulted in a posteriorly directed deceleration vector. During the loading of the final foot (left) the COP moved rapidly across towards a point midway between the feet. Here the critical nature of the M/L COP shift is seen. The COP must move to a position ahead of the future and final position of the COM. In this way the COP-COM vector will be directed towards the COM to bring it to a stop between the two feet. Then the COP moves posteriorly towards the COM and finally meets it as the subject comes to a quiet standing position. In Figure 13 the meeting of COP with COM was quite efficient; trials from subjects with peripheral sensory loss revealed 'overshoots' by the CNS. Figure 14 is representative of one of those cases.

Termination of gait is somewhat more difficult to achieve than initiation because of the need of the CNS to predict the future position of the COM. There is no doubt that even in quiet standing the CNS somehow calculates the total body COM; we would speculate that the somatosensory system is primarily involved. With this capability it is not hard to extrapolate to the capability to calculate the velocity and acceleration of the COM in order to predict the future position of the COM. Early (unreported) information indicates that the elderly do not have as smooth a termination as do young adults; they tend to overshoot and undershoot the final COM position before they come to a rest. In balance disorder patients it would be valuable to document these combined trajectories with a view to describing the nature of degeneration of their balance control systems.

Balance and posture in human gait

Walking versus *standing*

Human-walking as bipeds provides a particularly challenging balance task to the CNS and this is grossly different from the balance task during standing. Studies of balance and posture during quiet or perturbed standing have identified the dominance of the ankle muscles (plantarflexors/dorsiflexors) in the A/P direction and hip abd/adductor muscles in the M/L direction. This is not surprising when we consider the task of balance is to keep the body's COG safely within the base of support. However, when we walk the task suddenly changes, as was indicated in Figures 12 and 13, which showed the trajectory of the COG during the initiation and termination of gait. In order to accelerate our COG in a forward direction we must voluntarily initiate the start of a forward fall to accelerate the COG ahead of the base of support. The reverse is true during termination of gait where the COG must return within the base of support. Once initiation is achieved the COG is seen to

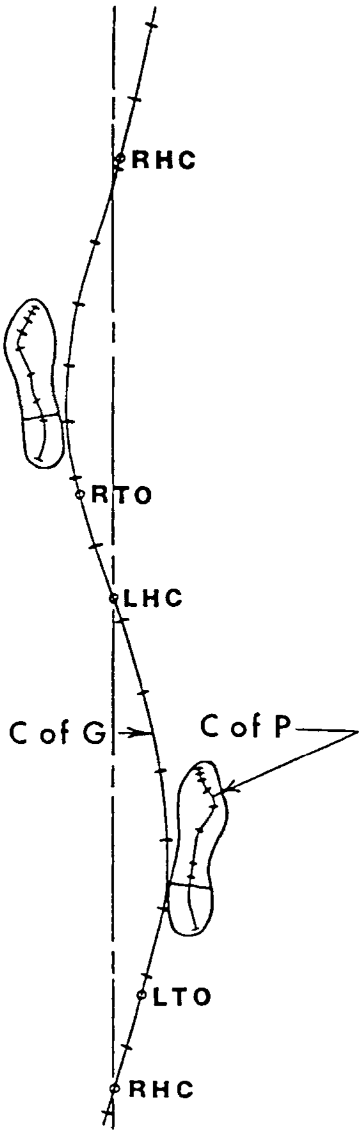


Figure 15. Total body centre of gravity and the centre of pressure under the support feet during level walking. The COG moves forward along the medial border of each support foot and during single support it is accelerated away from the support foot towards the future position of the swing foot. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn, University of Waterloo Press, 1991.

move forward along the medial border of the foot as depicted in Figure 15. Thus the ankle muscles are no longer important because the balance task has changed[64]; any activity of the stance ankle muscles cannot avert a fall, they can merely fine tune the anteri-oposterior or mediolateral acceleration of the body's COG. Only by safe placement of the swing foot do we avert a fall once every step. Restabilization can take place during the two short double-support periods, but during this time the support base is not very firm (one foot is accepting weight on the small area of the heel while the other is pushing off on the forepart of the foot).

The second challenging fact is that the distribution of body mass is such that two-thirds of its mass in the head, arms, and trunk (HAT) is located two-thirds of body height above the ground. Such an inverted pendulum is inherently unstable when we consider the forward momentum of HAT and the trajectory seen in Figure 15.

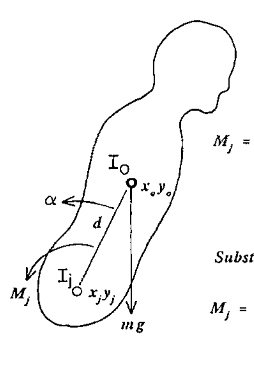
Finally, in the presence of all these challenging control problems the CNS manages to keep the large inertial load of HAT erect (within ± 1.5°) and with severely attenuated head accelerations[65,66]. The moment of inertia of HAT about the hip joint for a typical adult (7.0 kg m^2) is only 1/8th that about the ankle joint (55 kg m^2). Thus eight times the moment of force would be required of the ankle muscles compared with the hip muscles; fortunately the CNS does not even attempt such a control during steady-state walking. Dynamic balance of HAT in both the plane of progression and the frontal plane must be analysed using an inverted pendulum model of HAT, which is now discussed.

Inverted pendulum model

It is important that biomechanical model of large segments such as HAT are based on balance on a joint moving in space. Figure 16 shows such a model for HAT in the plane of progression as the HAT segment rotates about the hip joint. The dynamic equilibrium equations that describe the moments acting on the segment are usually derived as acting about the COM of the segment. In final form it is desired to see this equation with the moments as they act about the pivot point, in this case the hip joint.
The final equilibrium is:

$$M_j = mg\left(x_o - x_j\right) + m\ddot{y}_j\left(x_o - x_j\right) - m\ddot{x}_j\left(y_o - y_i\right) + I_j\alpha \qquad \text{(Eq. 6)}$$

The first term, *Mj*, is the net muscle moment acting at the joint in the plane of interest (either the plane of progression or the frontal plane). The second term is the gravitational moment due to the fact that the COG does not lie directly over the hip joint centre. The next two terms combined are the couples created by the fact that



About C of M

$$M_j = m\ddot{x}_o\left(y_j - y_o\right) + m\left(\ddot{y}_o + g\right)\left(x_o - x_j\right) + I_o\alpha$$

$$M_j = \underbrace{mg\left(x_o - x_j\right)}_{\textit{gravitational}} + \underbrace{m\ddot{y}_o\left(x_o - x_j\right) - m\ddot{x}_o\left(y_o - y_j\right)}_{\textit{couple due to C of M acceleration}} + \underbrace{I_o\alpha}_{\textit{inertial}}$$

Substituting: $I_j = I_o + md^2$ *and rearranging*

$$M_j = \underbrace{mg\left(x_o - x_j\right)}_{\textit{gravitational}} + \underbrace{m\ddot{y}_j\left(x_o - x_j\right) - m\ddot{x}_j\left(y_o - y_j\right)}_{\textit{couple due to joint acceleration}} + \underbrace{I_j\alpha}_{\textit{inertial}}$$

Figure 16. HAT segment as an inverted pendulum on the supporting hip joint. The dynamic equilibrium equation (normally calculated about the centre of mass of H.A.T.) is modified so that the gravitational, acceleration and inertial components acting about the hip joint can be identified.

the joint centre is accelerating and the final term is the associated moment that causes the segment to undergo an angular acceleration, α. By partitioning these terms in this manner we can readily see how large are the disturbing moments (gravitational and accelerational) and how well the CNS recognizes and counters these disturbances with the muscle moment Mj.

Kinematics of HAT

The trunk fluctuates about $\pm 1°$ over the stride with a trial variability of about $\pm 1.5°$ [13,65–69]. In spite of these small angular changes of HAT the linear accelerations in the GRS are not insignificant. The horizontal and vertical accelerations of the pelvis, thorax and head are quite revealing in showing the role of spinal column and spinal musculature in gait. Figure 17a,b plots the vertical and horizontal accelerations of the pelvis, thorax, and head ensemble-averaged over nine repeat walking trials[66]. It is evident from figure 17a that there is a very minor attenuation of these vertical accelerations indicating a lack of shock absorption by the spinal column. It may be that these low-level accelerations are well within safe bounds and therefore do not need to be attenuated. However, in the horizontal A/P duration the accelerations (Figure 17b) are seen to be severely attenuated as we progress upwards from the pelvis to the head. These reduced accelerations result from damping of the pelvic acceleration by the spinal column to make the head more stable, presumably for a more stable visual platform. The damping could be completely passive or be assisted by active control of the spinal muscles. The latter appears to be true, as is seen in the EMG profiles of the muscles of the trunk and neck[70]. In fact the phasic profiles of the paraspinal muscles at the C_7 level are 60 ms ahead of the profiles seen at the L_4 level, indicating that the balance control of the vertebrae is a 'top-down' anticipatory control. Thus the CNS stabilizes the most peripheral segment first then stabilizes each vertebral level downwards from the cervical to the thoracic and then lumbar levels.

Dynamic balance of HAT

In the plane of progression the HAT segment appears as in Figure 18 during weight acceptance and push-off. Equation 6 is the equilibrium equation for HAT, showing all the terms. Because HAT is virtually erect the gravitational term can be set to zero. Because the horizontal acceleration, $\ddot{x}_j$, is very large and also acts a

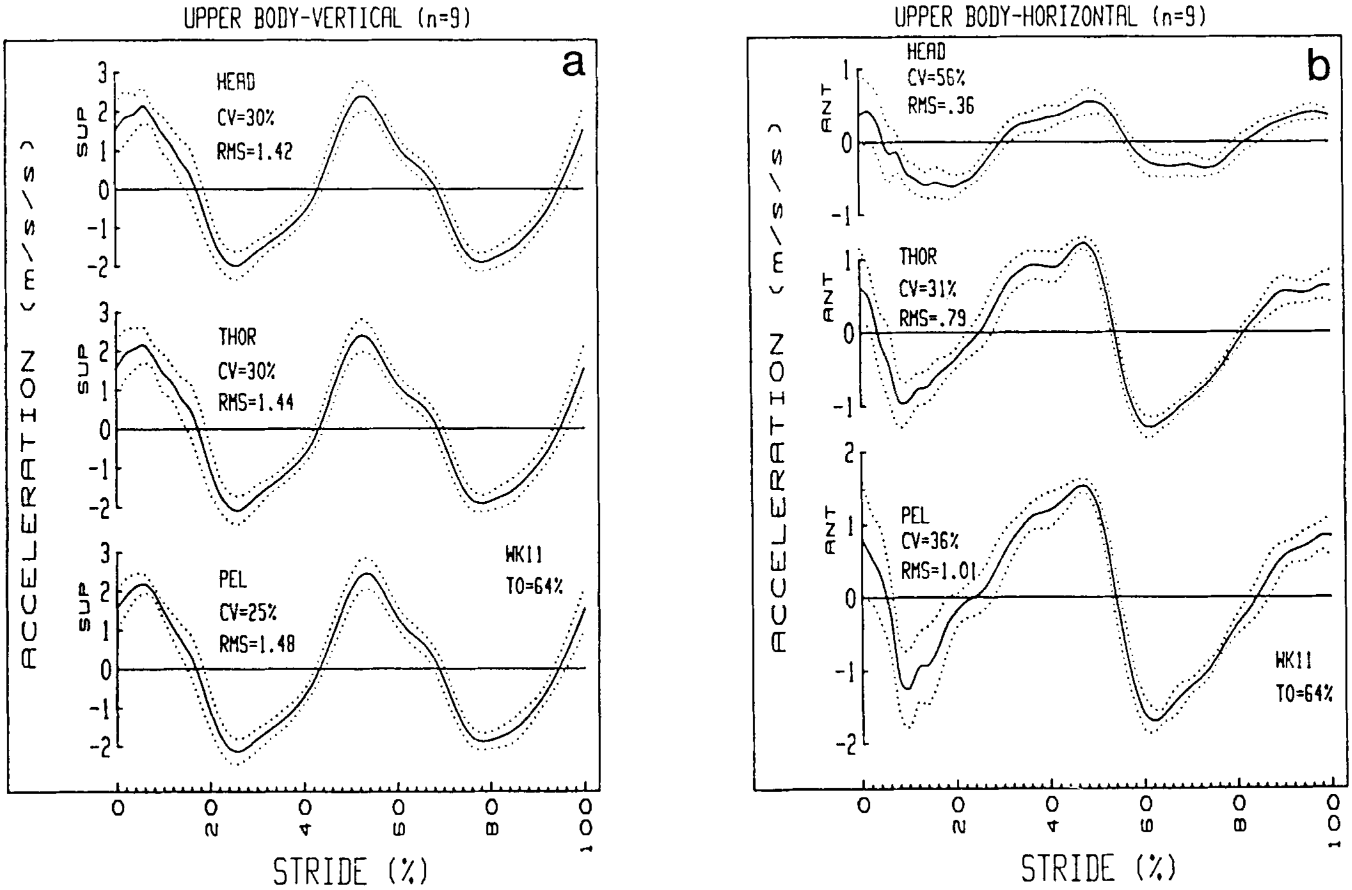


Figure 17a. Vertical acceleration of the head, thorax, and pelvis over one stride. Average of nine strides are shown. There is a minor attenuation of the head acceleration compared to the pelvis, indicating minor shock absorption by the vertebrae. **b** Horizontal acceleration of the head, thorax, and pelvis over one stride for the same subject. Head acceleration is considerably attenuated relative to the pelvis, indicating considerable active or passive attenuation by the vertebrae. Low head acceleration results in a stable platform for the visual system. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

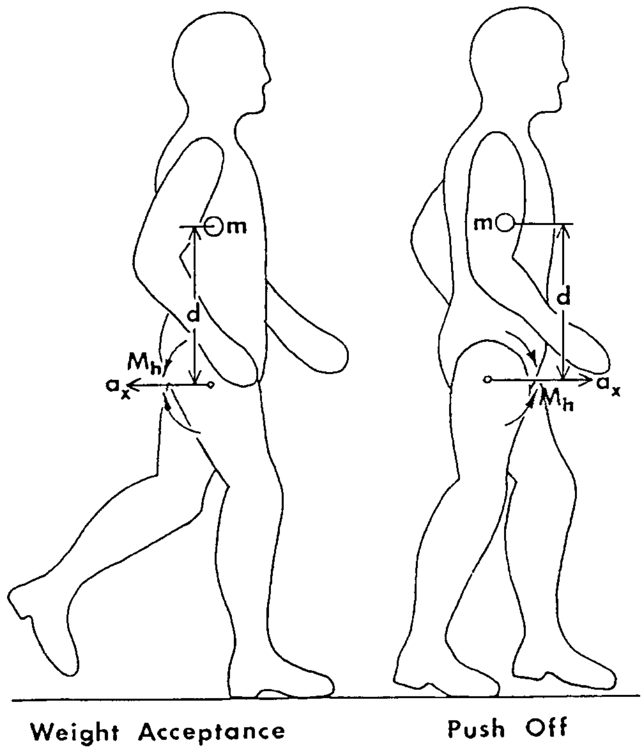


Figure 18. Dynamics of the balance of HAT during stance. With the trunk vertical the gravitational term of Eq. 6 is zero. However, the major perturbation is the posterior acceleration during weight acceptance and the anterior acceleration during push off; the resultant destabilizing couple must be countered by hip extensor and flexor moments to keep HAT erect. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

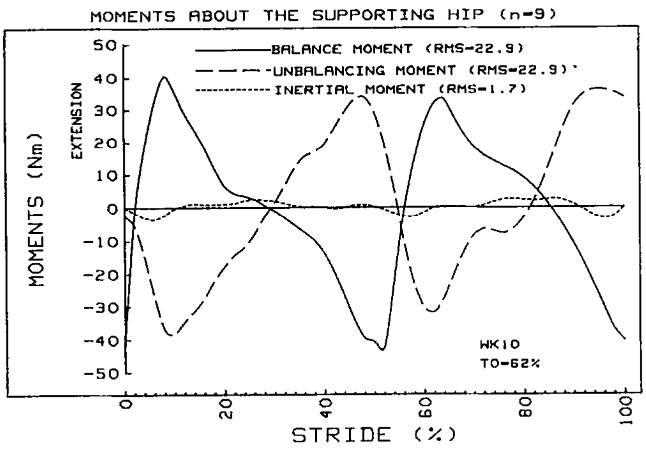


Figure 19. Unbalancing moment over the stride period due to the destabilizing hip accelerations. The balancing moment is the active hip extensor/flexor muscle patterns which are seen to virtually cancel the unbalancing moment. The resultant inertial moment, Iα, is quite small, which explains why the HAT remains erect over the gait cycle. During double-support periods (0–10%, 50–60%) the unbalancing and balancing moments were calculated assuming a linear transfer from one support hip to the new support hip. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

large distance from the COM of HAT the resultant acceleration couple dominates. It is referred to as an unbalancing moment that acts strongly to cause HAT flexion during weight acceptance and then cause extension during push-off. The response of the CNS is to generate an almost equal and opposite hip moment, as seen in Figure 19 for one of the five subjects analysed. This curve represents the ensemble average of nine repeat walking trials. The balance moment was set equal to the stance limb hip moment during single support; during double support a linear weighting was used to transfer responsibility from the push-off limb to the weight-accepting limb. The sum of the unbalancing and balance moments virtually cancel each other, to result in a very small inertial moment, $I_j\alpha$. The resultant low angular acceleration of HAT now explains the very small changes in trunk angle over the gait cycle.

The major conclusion, based on results from 9 or 10 repeat trials of five subjects[66], was that the hip flexors/extensors in the plane of progression have the single role of dynamic balance of HAT. Such a balance synergy has been inferred previously[64]. From a neurological perspective it remains to be seen how such an instantaneous control is achieved. The accelerations seen at the head by the vestibular system (Figure 17b) are an attenuated version of the unbalancing hip acceleration and thereby have the potential as a feedback control signal. However, the balancing hip moment is virtually in phase with the perturbing acceleration; thus the vestibular acceleration is not only the wrong polarity but is not sufficiently in advance to account for the neurological latencies and the electromechanical lag in the build-up and decay of muscle tension.

Postural responses

The previous analyses were based on an erect HAT segment over the walking stride. If a subject voluntarily adopts a forward or backward postural lean, what is the response of our hip motor patterns to this perturbation? Figure 20 shows the response of the right limb motor patterns to a forward lean of 22° (FOR) and backward lean of 8° (BACK) compared with the nine repeat erect walking trials[66]. It is evident that the forward lean resulted in an enhanced response of the posterior muscles at all three joints. The hip was biased with a larger extension moment; the knee was biased in the flexor direction; the ankle was more plantarflexor. The backward lean trial showed exactly the opposite response at all three joints. The left limb showed exactly the same changes during its support period, and when the hip moments were combined to create the hip balance moment (Figure 21) the postural bias component is even more pronounced. The upper body's response at the level of L_{4-5} is also seen in this figure. It is important to note in all these results that the net muscle moment was the sum of the balance and postural components.

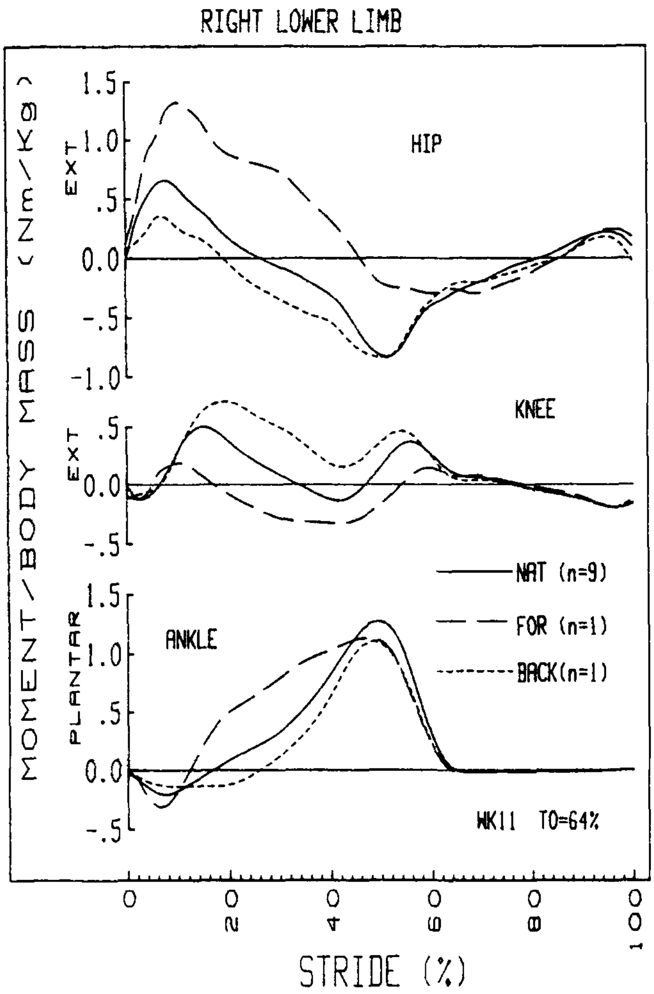


Figure 20. Hip, knee and ankle moments for one subject walking with a forward lean (FOR) and backward lean (BACK) compared with the average of nine strides while walking erect. The postural response during the support period is the same as in quiet standing: for a forward lean all posterior muscles increase their activity, for a backward lean the anterior muscles increase their activity (or at the ankle, the plantarflexors are less active). Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

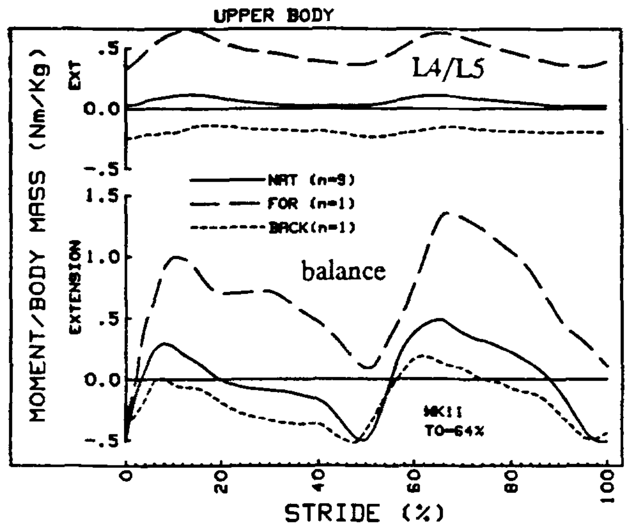


Figure 21. Balance moment and lumbar movement (L_{4-5}) response to a forward and backward lean for the same subject as in Figure 20. The postural response is a bias towards posterior muscles for a forward lean and increased anterior muscle activity for a backward lean. Lower trace is the balance moment, upper trace is the L_{4-5} moment. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

Support synergy

The control of vertical collapse of the body against gravity is one of the three subtasks in gait. It is a sagittal plane task and involves the control of the hip, knee, and ankle angles during stance. In order to collapse vertically the knee must flex but the ankle and hip must also undergo flexion. Thus the extensor muscles at each of those joints are the main line of defence against collapse and this response has been quantified as the total limb synergy called support moment[49]. The support moment, *Ms*, was seen to be quite consistent in repeat intrasubject analyses in spite of considerable variability in the individual joint moments whose summation made up *Ms* [50]. Especially at the hip and knee we saw high variability in the moment patterns. Now, as a result of the balance control synergy of the hip flexors/extensors, we can see the logic behind variability measures. The high hip moment variability on a trial-to-trial or a day-to-day basis results from the changing demands of dynamic balance. Thus the high hip moment variability is desirable in the dynamic balance of HAT but is undesirable as far as support against collapse. Thus a second synergy, the support synergy comes into play, where the stride-to-stride variations in the hip moment are largely cancelled by opposite changes in the knee and ankle moments. There are several ways to describe and quantify the trade-offs between these joints, and these will now be discussed.

Figure 22 shows the nature of these stride-to-stride differences[50]. Two of the nine trials for this subject were selected along with the mean of the nine strides (solid line). WM22D (i.e. subject WM22, trial D) is shown by the long dashed line; WM22J by the short dashed line. During trial J the subject had a dominant hip extensor pattern and a knee flexor pattern, whereas trial D was biased in the reverse direction: hip flexor and knee extensor. Thus the dominant muscle pattern for trial J was hamstrings (posterior muscles) in contrast to that of trial D, which was primarily rectus femoris (anterior muscles). The net extensor pattern acting on the thigh for both of the strides was about the same, but in one case it was being accomplished by anterior muscles and in the other by posterior muscles. Thus this trade-off between anterior and posterior did not influence the net extensor (support pattern).

A more quantitative way to see the trade-offs between hip and knee and between knee and ankle is through a covariance measure. Figure 23 documents the degree of covariance between the joint moment patterns for the subject presented in Figure 22, along with the degree of covariance from a second subject who was assessed over 10 repeat walking trials recorded minutes

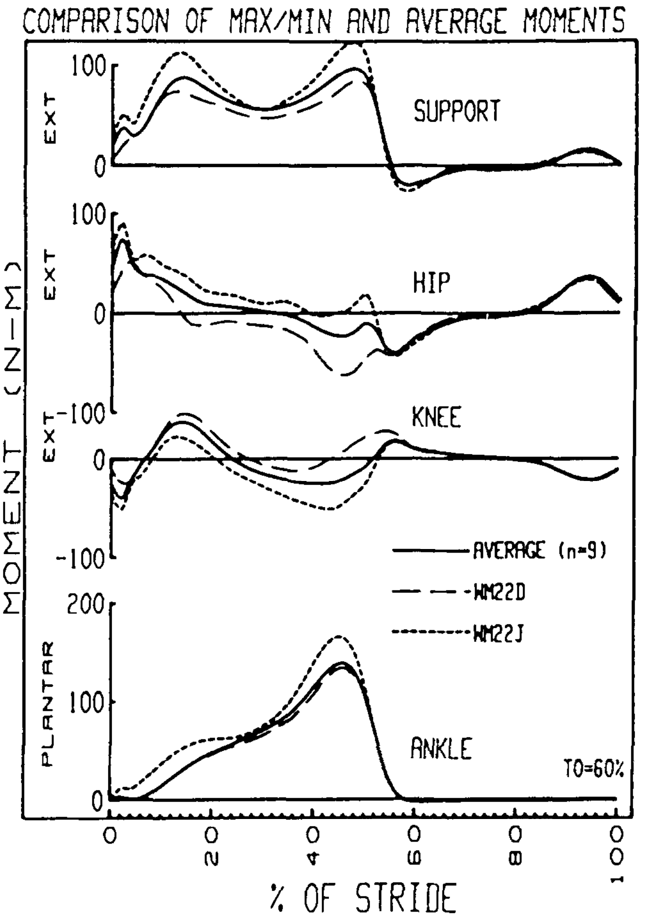


Figure 22. Stride-to-stride differences in the hip, knee, and ankle moments for the same subject walking at her natural cadence on different days. See text for discussion of the day-to-day trade-offs. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

apart rather than days apart. The calculation of these mean variances and covariances is based on the following equations, with all units in (N m)^2:

$$\sigma_{hk}^2 = \sigma_h^2 + \sigma_k^2 - \sigma_{h+k}^2 \qquad \text{(Eq. 7)}$$

where: σ_h^2 and σ_k^2 are the mean variances over stance at the hip and knee.

σ_{h+k}^2 is the mean variance of the sum of hip + knee moment profiles.

σ_{hk}^2 is the covariance between hip and knee moment patterns.

The term σ_{hk}^2 can be expressed as a percentage of the maximum possible covariance, which would be 100% if $\sigma_{h+k}^2 = 0$, and would mean that the variability in the knee moment was completely out of phase with that at the hip. Thus the percent covariance, COV, is given by:

$$COV = \frac{\sigma_{hk}^2}{\sigma_h^2 + \sigma_k^2} x\ 100\% \qquad \text{(Eq. 8)}$$

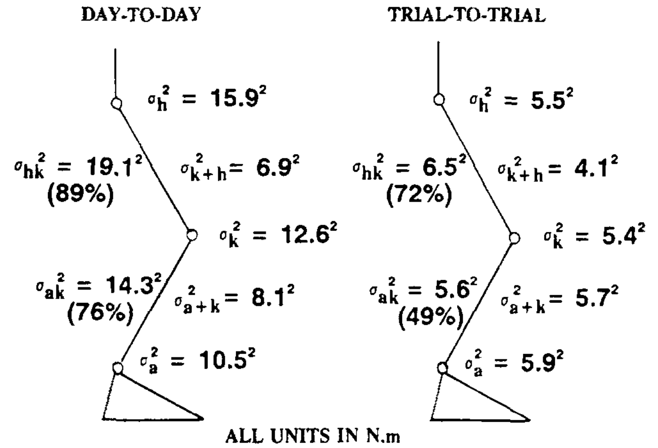


Figure 23. Covariance (COV) calculations of the moment patterns between the hip and knee and between the knee and ankle for nine separate trials on separate days and 10 separate trials minutes apart. See text for discussion of the variances and COV measures as evidence of joint-to-joint trade-off and the explanation for a consistent support moment pattern. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

In the first set of repeat trials the COV between the hip and knee was 89%. A similar calculation can be made at the knee and ankle, yielding a COV of 75%. These high covariances are extremely strong evidence of tight coupling between the motor patterns at these adjacent joints. This result is not surprising when we consider the opposite and cancelling functions of the hamstrings and rectus femoris muscles at the hip and knee, and of the gastrocnemius muscle at the knee and ankle. Therefore this tight coupling is partially due to the anatomy of these biarticulate muscles. Anatomically the potential for force generation can be considered to be approximately proportional to the physiological cross-sectional area of each muscle. However, the single joint muscles constitute 2/3 of the physiological cross-sectional area of all the muscles crossing the hip and knee joints[71]. Thus, the magnitude of COV is well in excess of that possible anatomically, and therefore must be part of a neural control pattern.

The lower COVs from the second subject's trials (73% between the hip and knee and 49% between the knee and ankle) appear to be due to the lower variability (adaptability) seen over these repeat trials that took place minutes apart. Thus it appears that humans adopt patterns that have larger differences on a day-to-day basis than they do over shorter periods of time. These day-to-day and minute-to-minute alterations are largely very deterministic and reflect the plasticity of the motor control system.

Frontal plane balance

In the frontal plane we have two inverted pendulum systems[35]. As shown in Figure 24a, when the body is in single support, HAT pivots about the hip joint and in Figure 24b the total body pivots about the subtalar joint. Because HAT maintains a nearly vertical position

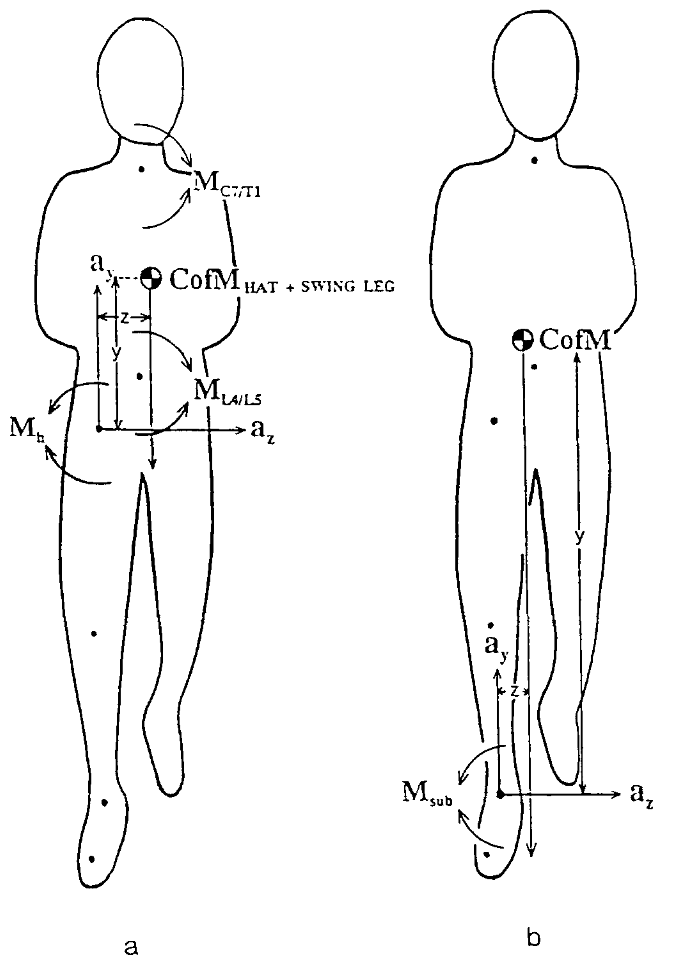


Figure 24. **a** Inverted pendulum model of balance in the frontal plane about the hip joint as the HAT + swing limb are supported on the stance hip. The dynamic equilibrium equation (Eq. 6) is used to separate the components of balance (see Figure 25). **b** Inverted pendulum model of balance of the total body in the frontal plane about the supporting subtalar joint. The dynamic equilibrium equation (Eq. 6) separates the components of balance (see Figure 26). Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

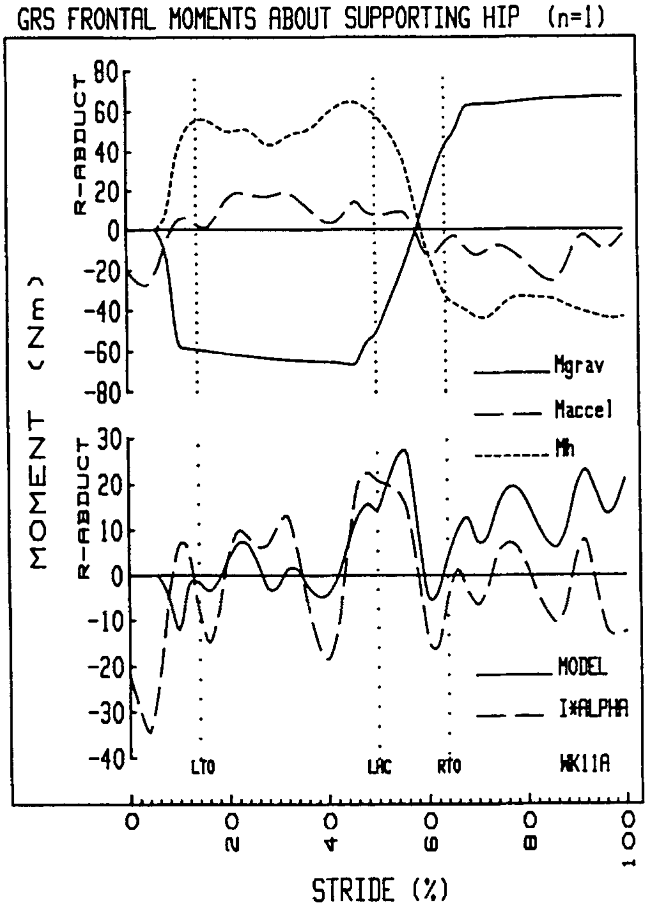


Figure 25. Dynamic equilibrium components for the total body balance in the frontal plane about the subtalar joint. See text for discussion. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

(within ±2°) we can assume that the total body system can be considered a nearly rigid inverted pendulum. Because the COM of HAT and the total body are always medial of the pivot joint we must consider all acceleration terms in Equation 6, including gravitational.

In the GRS the frontal plane moments about the support hip are presented in the upper half of Figure 25 for one of the eight walking trials for one representative subject. The variability in eight moment patterns was quite low. The gravitational moment during single support is that calculated about the support hip; during double support a linear weighting is assumed in the transfer between the unloading and weight-accepting hips. A similar transfer was assumed between the right and left hip muscle moments and the acceleration couple. The sum of these three moments are what cause the HAT segment to undergo a M/L angular acceleration; this summation appears in the lower half of this Figure (solid line). The angular inertial moment, $I_h\alpha$, appears as a dashed line, and the magnitude of these two curves is quite low, about 10 N m. The residual difference between these curves is quite small, which is a validation that the inverted pendulum model assumed in the dynamic equilibrium equation was correct. The gravitational load is about 60 N m during single support and this is countered by the hip abductors which respond with about 50 N m, while the hip acceleration (which is medial during single support) also assists with about 10 N m. These findings support the fact that the CNS knows about the medial acceleration and takes it into account, thus reducing the necessary abductor activity.

The inverted pendulum model shown in Figure 24b yielded three moments that acted about the subtalar joint: gravitational, joint acceleration, and subtalar muscles (invertor/evertors); these are presented in the upper half of Figure 26. It is evident that two of the moments dominate. The gravitational moment (solid line) exceeds 40 N m during single support and is essentially decided by the distance the foot is placed lateral of the body's COG. The second major moment is due to the M/L acceleration (long dashed line) of the subtalar joint. The third and almost insignificant contribution is

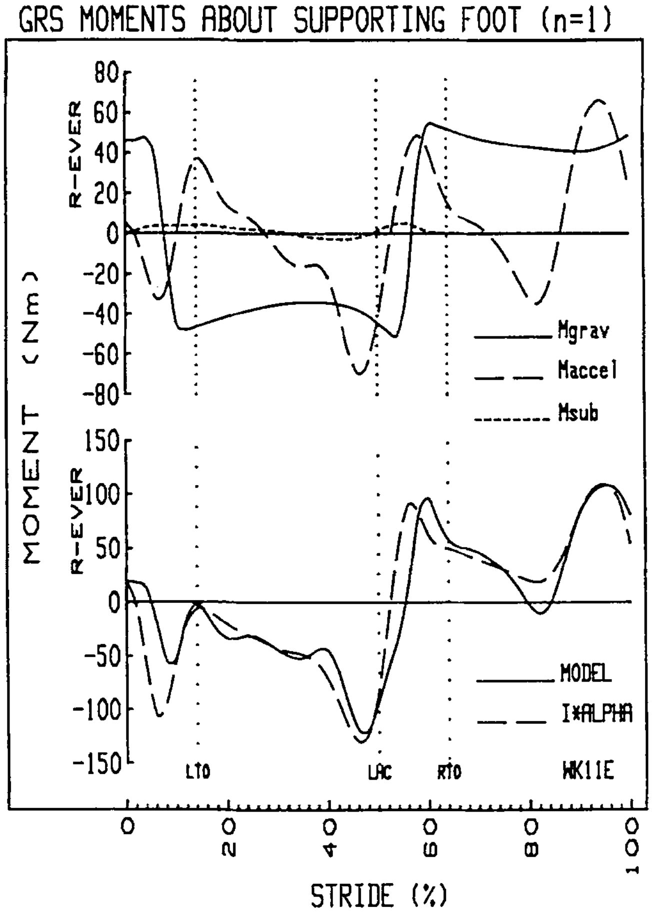


Figure 26. Dynamic equilibrium components for the total body balance in the frontal plane about the subtalar joint. See text for discussion. Reproduced with permission from *The Biomechanics and Motor Control of Human Gait*, 2nd edn. University of Waterloo Press, 1991.

the subtalar moment (short dashed line). As was done previously, the sum of these three moments should equal the inertial moment, $I_s\alpha$; the lower traces of Figure 26 show a good match of this dynamic equilibrium model (solid line) with the independently calculated $I_s\alpha$.

Summary

The common denominator in the assessment of human balance and posture is the inverted pendulum model. If we focus on appropriate versions of the model we can use it to identify the gravitational and acceleration perturbations and pinpoint the motor mechanisms that can defend against any perturbation.

We saw that in quiet standing an ankle strategy applies only in the A/P direction and that a separate hip load/unload strategy by the hip abd/adductors is the totally dominant defence in the M/L direction when standing with feet side by side. In other standing positions (tandem, or intermediate) the two mechanisms still work separately, but their roles reverse. In the tandem position M/L balance is an ankle mechanism (invertors/evertors) while in the A/P direction a hip load/unloading mechanism dominates.

During initiation and termination of gait these two separate mechanisms control the trajectory of the COP to ensure the desired acceleration and deceleration of the COM. During initiation the initial acceleration of the COM forward towards the stance limb is achieved by a posterior and lateral movement of the COP towards the swing limb. After this release phase there is a sudden loading of the stance limb which shifts the COP to the stance limb. The COM is now accelerated forward and laterally towards the future position of the swinging foot. Also M/L shifts of the COP were controlled by the hip abductors/adductors and all A/P shifts were under the control of the ankle plantar/dorsiflexors. During termination the trajectory of both COM and COP reverse. As the final weight-bearing on the stance foot takes place the COM is passing forward along the medial border of that foot. Hyperactivity of that foot's plantarflexors takes the COP forward and when the final foot begins to bear weight the COP moves rapidly across and suddenly stops at a position ahead of the future position of the COM. Then the plantarflexors of both feet release and allow the COP to move posteriorly and approach the COM and meet it as quiet stance is achieved. The inverted pendulum model permitted us to understand the separate roles of the two mechanisms during these critical unbalancing and rebalancing periods.

During walking the inverted pendulum model explained the dynamics of the balance of HAT in both the A/P and M/L directions. Here the model includes the couple due to the acceleration of the weight-bearing hip as well as gravitational perturbations. The exclusive control of A/P balance and posture are the hip extensors and flexors, while in the M/L direction the dominant control is with the hip abductors with very minor adductor involvement. At the ankle the inverted pendulum model sees the COM passing forward along the medial border to the weight-bearing foot. The model predicts that during single support the body is falling forward and being accelerated medially towards the future position of the swing foot. The model predicts an insignificant role of the ankle invertors/evertors in the M/L control. Rather, the future position of the swing foot is the critical variable or more specifically the lateral displacement from the COM at the start of single support. The position is actually under the control of the hip abd/adductors during the previous early swing phase.

The critical importance of the hip abductors/adductors in balance during all phases of standing and walking is now evident. This separate mechanism is important from a neural control perspective and clinically it focuses major attention on therapy and potential problems with some surgical procedures. On the other hand the minuscule role of the ankle invertors/evertors is important to note. Except for the tandem standing position these muscles have negligible involvement in balance control.

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