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Balancing on a narrow ridge: biomechanics and control

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The balance of standing humans is usually explained by the inverted pendulum model. The subject invokes a horizontal ground-reaction force in this model and controls it by changing the location of the centre of pressure under the foot or feet. In experiments I showed that humans are able to stand on a ridge of only a few millimetres wide on one foot for a few minutes. In the present paper I investigate whether the inverted pendulum model is able to explain this achievement. I found that the centre of mass of the subjects sways beyond the surface of support, rendering the inverted pendulum model inadequate. Using inverse simulations of the dynamics of the human body, I found that hip-joint moments of the stance leg are used to vary the horizontal component of the ground-reaction force. This force brings the centre of mass back over the surface of support. The subjects generate moments of force at the hip-joint of the swing leg, at the shoulder-joints and at the neck. These moments work in conjunction with a hip strategy of the stance leg to limit the angular acceleration of the head–arms–trunk complex. The synchrony of the variation in moments suggests that subjects use a motor programme rather than long latency reflexes.

Keywords: balance; control; human; ridge; computer simulation

1. INTRODUCTION

The reigning theory on human balance is that of the inverted pendulum. This theory states that humans tend to fall away from their centre of pressure on the ground as a stick or inverted pendulum would do. Subjects avoid falling by changing the location of the centre of pressure either under the stance foot or by placing their other foot on the ground. This strategy is very similar to the one used when balancing a stick on a finger. The finger makes fast and short-lasting movements in a horizontal plane. These movements are of larger amplitude than those of the centre of mass of the stick. This is a way to give the reaction force from the finger an opposite direction from that of the movement of the centre of mass of the stick. The finger limits the falling motions of the stick in this way. The only difference between this strategy and the one used when standing on one foot is that the subject does not move the foot in a horizontal plane. The subject moves the centre of pressure under the foot by changing moments of force in the ankle. Effectively, this is very similar to a balanced stick, hence the name ‘inverted pendulum’.

MacKinnon & Winter (1993) studied the balance of human subjects in the frontal plane. They studied the balance of the head–arms–trunk complex and the swing leg on top of the hip-joint of the stance leg. They also looked at the balance of the total body less the supporting foot on top of the subtalar joint. For both balance tasks during walking, they produced an inverted pendulum model and found a satisfactory correspondence between model simulations and observations. This study does not include the ground-reaction force nor any changes in it. It

looks at moments generated at the ankle and the hip-joint and shows that the sums of the moments equal the inertial moment of force of the system. D’Alembert’s principle formulates this equality (Huston 1990).

Several authors (Dietz 1992; Massion 1992; Nashner 1985) have claimed that human balance during standing is achieved by what is called the ‘hip strategy’. Shumway-Cook & Woollacott (1995) explain the fast movements around the stance hip-joint during difficult balance tasks as an attempt to keep the centre of mass over the surface of support. Winter (1995) shows in a simple model-study that a hip moment can move the centre of mass without explaining how this works. Among others, Shumway-Cook & Woollacott (1995) propose a hierarchical model in which the ankle strategy is the first one to be active, followed by the hip strategy, followed by the stepping strategy.

Maki & McIlroy (1997) disagree with this hierarchical model and observe an overlap of the strategies. They explain the hip strategy in terms of the generation of shear forces at the feet, which decelerate the centre of mass. They claim, however, that subjects normally do not use the hip strategy. Subjects prefer to step out.

2. MATERIAL AND METHODS

(a) *Subjects*

After being informed about the method, 13 subjects gave their consent for the experiments. The subjects were five men (ranging in age from 19 to 30 years old, body mass 73–77 kg and stature 1.79–1.91 m) and eight women (ranging in age from 20 to 35 years old, body mass 57–80 kg and stature 1.63–1.82 m). The subject specifically mentioned in this paper, H.P., had a stature of 1.89 m and a body mass of 75 kg.

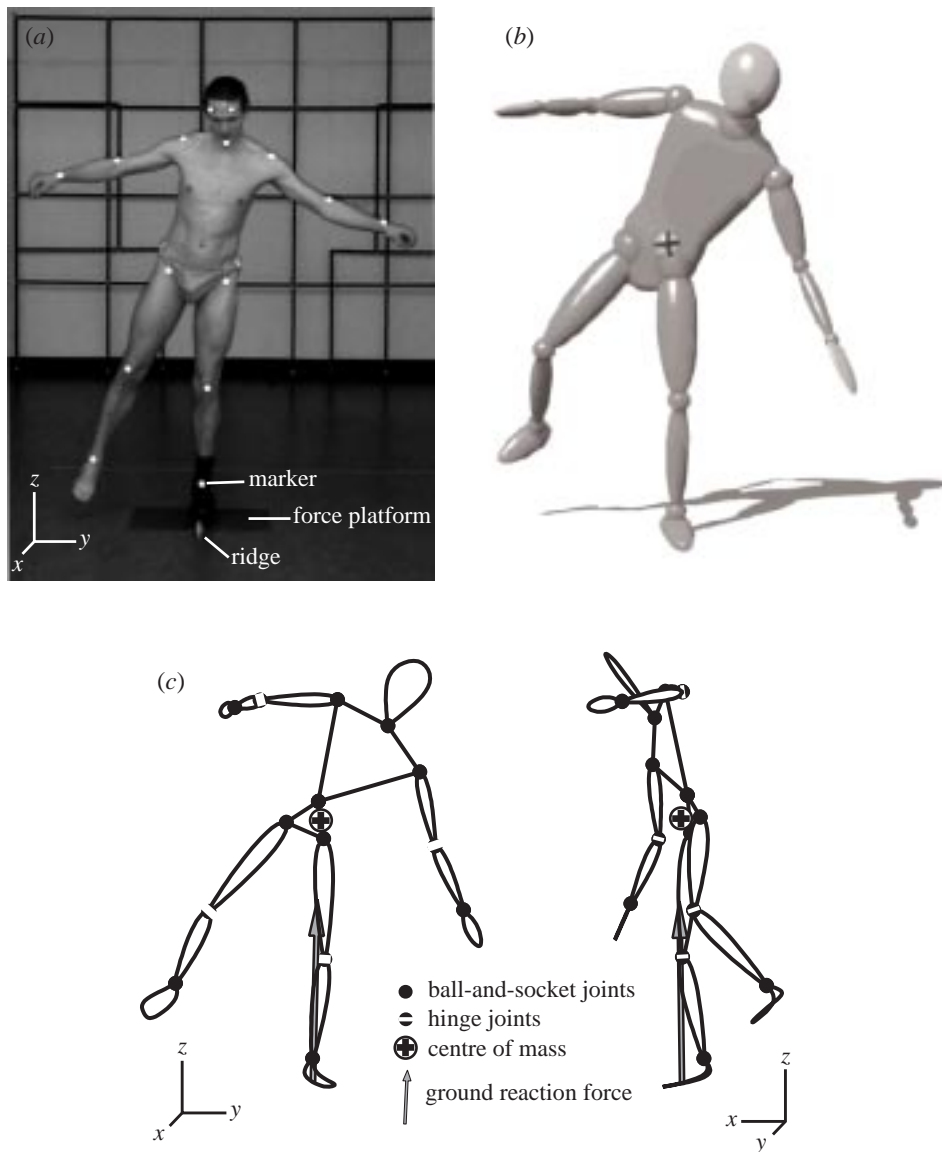


Figure 1. Measurements and human-body model. (a) The placement of 15 markers on a subject standing on a narrow ridge on a force platform. (b) One frame from an animation produced by the human-body model. The cross indicates the centre of mass. (This animation is currently available at www.pubs.royalsoc.ac.uk/publish/phi_bs/rtb1385.htm) (c) Simplified drawings of the human body with an indication of the type of joints used.

(b) Measurements

The movements during balancing of the human subjects were recorded by means of an opto-electronic recording system (ELITE) with two cameras and reflective markers (Ferrigno & Pedotti 1985). The recording rate was 50 samples per second. Each subject was prepared by affixing 15 reflective markers to the bare skin in places indicated in figure 1a. The movement data were low-pass filtered at 4 Hz with a zero phase-shift digital filter. Missing markers (due to occlusion) were handled using the average of a cubic spline and a linear interpolation over time.

The subjects were asked to stand on a ridge with a height of 40 mm and a width of 4 mm with their preferred foot. The instruction was to stand on the ridge with only one foot, while the other foot was not allowed to touch the ground. The ridge was orientated in the dorsoventral direction as though it was a speed skate. The subjects were allowed to wear a shoe, thereby avoiding injury. The ridge was placed on top of a force platform (BERTEC 40 cm \times 60 cm). The six output channels of the platform were digitized at a rate of 50 samples per second. Sampling was performed in synchrony with the movement recordings. Three force components of the ground-reaction

force, the location of the centre of pressure, and the moment of force at the z -axis (see figure 1a,c) were calculated from the samples.

(c) Simulations

The coordinate system shown in figure 1 was adopted throughout. Simulations of human-body dynamics were performed by means of a package (HbmTbx, or Human body model Toolbox), written by E. Otten. The package runs on Power Macintosh computers. Toolbox is based on the approach offered by Huston (1990), which in itself is based on that of Kane & Wang (1965). The configuration chosen for the present study consisted of a 15-element human-body model. These elements were (i) one thoracic element; (ii) one pelvic element; (iii) one head element; (iv) two upper-arm elements; (v) two lower-arm elements; (vi) two hand elements; (vii) two upper-leg elements; (viii) two lower-leg elements; and (ix) two foot elements. The model had 40 degrees of freedom, which consisted of three system displacements, three system rotations and 34 degrees (see figure 1c). Ball-and-socket joints have three degrees of freedom each, and hinge joints one degree of freedom. Dimensions and inertias were used as published by Hanavan

(1964). When using the standard location of the joint between the thoracic element and the pelvic element, a mismatch was observed between the displacement of the hip markers and the hip-joints of the model. Therefore, I moved the joint in a caudal direction. In this way, an agreement was obtained between the displacement of the model hip-joints at a given rotation of the pelvic element and the displacement of the hip markers. The dimensions of the model were scaled linearly, with the stature of the subject divided by that of the standard human model of Hanavan. The segment masses were scaled linearly by the mass of the subject divided by that of the same human-body model. The inertias were scaled quadratically with the stature factor and linearly with the mass factor. The 15 markers together provided the information necessary to calculate all degrees of freedom of the model with the exception of the hand and feet angles. I kept these angles at neutral values (see figure 1*c*). The global degrees of freedom of the thoracic element were calculated from the two shoulder markers and the two hip markers. A local coordinate system of the thoracic element was set up by aligning its y -axis along the line connecting the shoulder markers. The y - z -plane was chosen through the shoulder markers and the middle between the hip markers. The z - and x -axes were chosen at right-angles to the y -axis. Here, as throughout the paper, a right-handed coordinate system was used. The local coordinate system of the pelvic element was set up by aligning its y -axis on the line connecting the hip markers. The y - z -plane was chosen through the hip markers and the middle between the shoulder markers. The z - and x -axes were chosen at right-angles to the y -axis. The local coordinate systems of the upper arms and upper legs were set up by aligning their z -axis along the length of the element. The positive segment of the axis pointed proximal. The z - x -plane runs through the three markers of the limb. Again the y -axis and the x -axis were chosen at right-angles to the z -axis. The elbow and knee angles were calculated from the three markers on the limb. The local coordinate system of the head was chosen with its y -axis along the two forehead markers, the z - y -plane through all three markers and the other two axes at right-angles to the y -axis.

The chains of dependency of the elements were chosen from the thoracic element to distal elements. The moments of force were defined as acting from the more proximal element to the more distal element. The same holds for the degrees of freedom of the system.

The simulations included inverse dynamics of the measured movements, resulting in moments at the modelled joints. These simulations also offered positions of the whole body centre of mass over time. For some purposes, simulations of forward dynamics were performed, in which case constraint equations (Huston 1990) were added to simulate floor interactions.

(d) *The centre of mass*

The centre of mass was calculated in two ways. In the first method I calculated the centre of mass from the ground-reaction force by dividing it by the mass of the subject and integrating the result twice over time. Since the initial values of position and velocity of the centre of mass are unknown, an integration error was built up that needed to be removed. This was done by using a high-pass digital filter with zero phase-shift with a cut-off frequency of 0.25 Hz. In this way the average of the position of the centre of mass becomes zero, but the short-lasting variations over time are maintained.

The second method uses the human-body model. Since the degrees of freedom of the model are known from the measure-

ments, the centre of mass follows from the configuration of the model over time. The position of the centre of mass is calculated by taking the algebraic sum of the positions of the centre of mass of all body elements, each multiplied by their mass. This sum is divided by the total mass of the system.

The results of the two methods were compared. Since the results come from different sources and along different mathematical pathways, such a comparison has ramifications for the reliability of the conclusions of this paper.

The accuracy with which the centre of mass can be reconstructed from the movement measurements was established in the following way. I asked one subject to stand as motionless as possible in six different poses on one foot for 10 s. The centres of mass and pressure were reconstructed, averaged (assuming that they would have the same average). These averages were subtracted from each other, generating a two-dimensional vector in the ground plane. The standard deviations of this vector along the x - and y -axes were calculated. It appeared that the position of the centre of mass as calculated has a standard deviation of 4.5 mm in the frontal plane (along the y -axis) and 2.0 mm in the sagittal plane (along the x -axis).

3. RESULTS

From the recordings of the balancing subjects, the position of the centre of mass was calculated along the y -axis according to the two methods described above as a function of time. The results of one particular recording are shown in figure 2*a*. After *ca.* 7.2 s, the subject stood with one foot on the ridge. The subject lost balance after *ca.* 39 s. The curves show a good resemblance, the standard deviation of the difference being 5.5 mm during balancing. This number is in agreement with the accuracy of the determination of the centre of mass given above.

Figure 2*b* shows the movements of the centre of mass (based on the model simulation) projected on the ground plane together with the measured centre of pressure. As can be seen, the centre of mass projection moves up to 4 cm outside the base of support that is visible from the cloud of centre-of-pressure points.

According to the inverted pendulum model, it is not possible for the centre of mass to move beyond the base of support, unless the other foot is placed, extending the base of support. That implies that this observation requires a new explanation.

Figure 3 shows three frames of the simulation of the inverse dynamics, each separated by only 0.2 s from its neighbour. Although the general posture is comparable, the horizontal component of the ground-reaction force varies between about -100 N and $+100$ N. The inverted pendulum model requires that there is some relation between the direction of the line connecting the centre of pressure and the centre of mass and the direction of the ground-reaction force. This is the reason why a stick can never return from a falling motion unless the base of the stick is moved. Clearly, in this task of balancing on a ridge, there is no such relation. On the contrary, the ground-reaction force and the position of the centre of mass show opposite signs (figure 4). This means that the direction of the ground-reaction force can be manipulated by the subject. This offers the possibility of using the ground-reaction force to pull the centre of mass back above the surface of support.

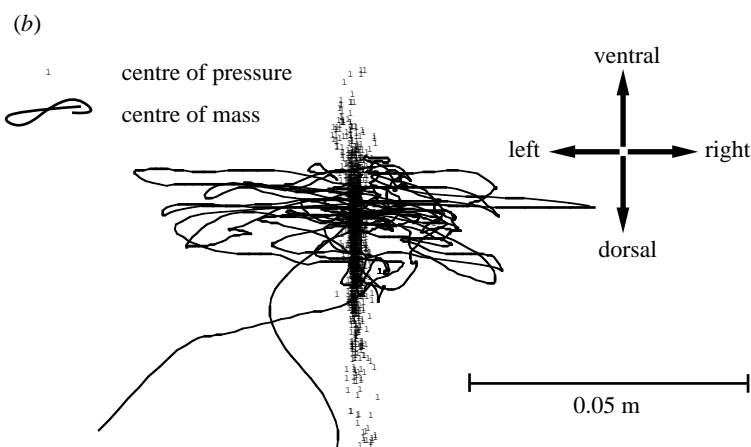
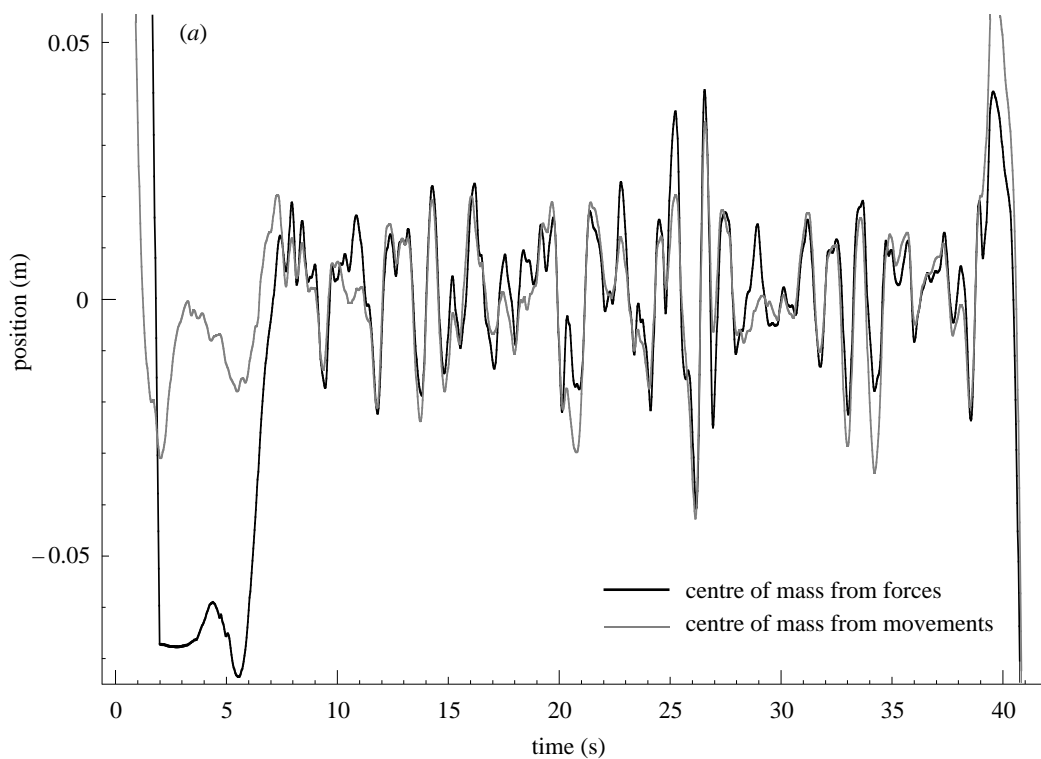


Figure 2. The location of the centre of mass in one experimental trial. (a) The location as a function of time according to the two methods used. The solid line shows the ground-reaction force components, the dotted line shows the movement recording. (b) The location of the centre of mass projected on the ground plane together with the centre of pressure. Note that the centre of mass does not remain within the base of support.

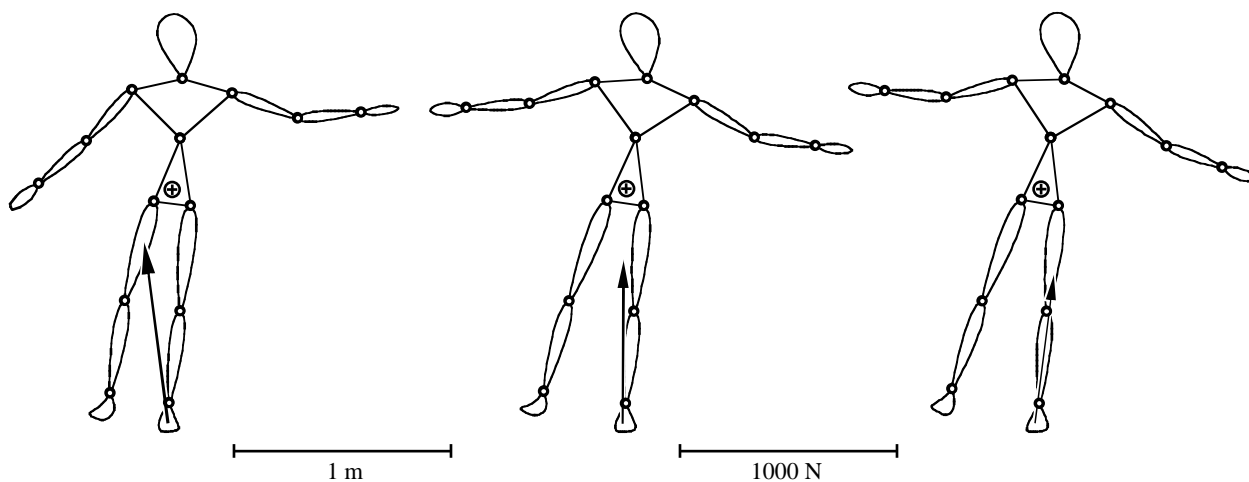


Figure 3. Three frames from the human-body model simulation, each 0.2 s apart. Note the large changes in ground-reaction force direction.

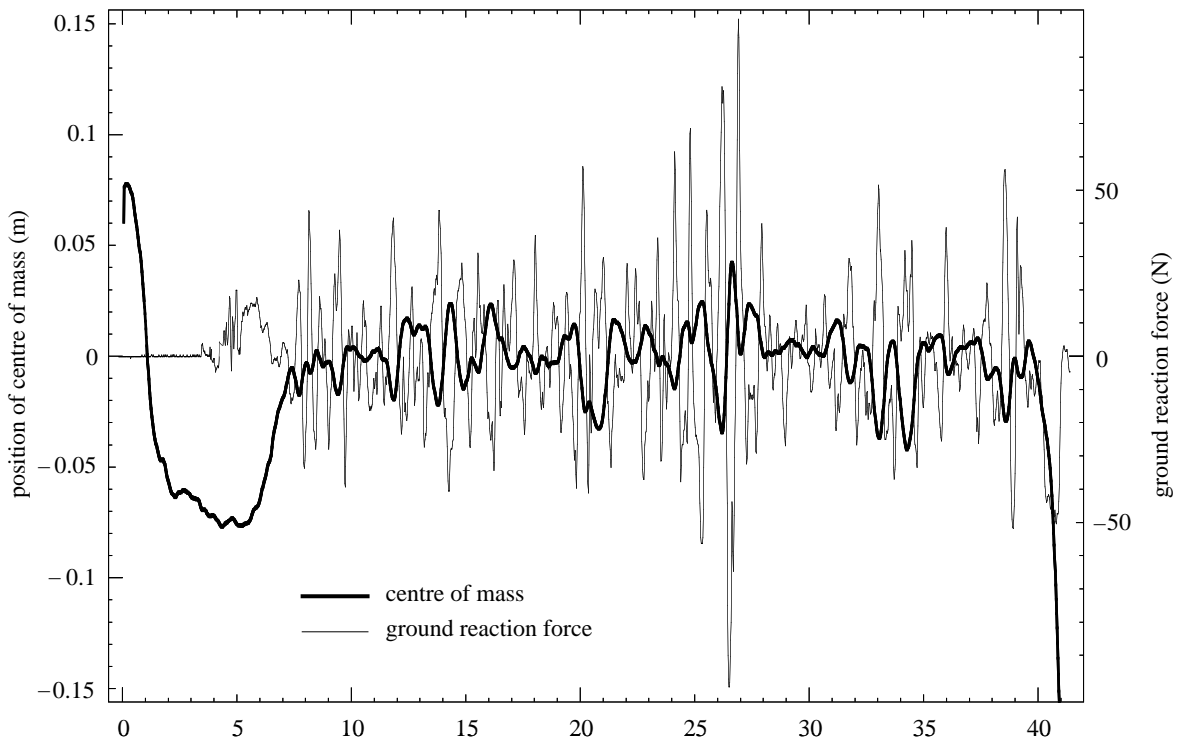


Figure 4. The position of the centre of mass as a function of time compared with the invoked horizontal component of the ground-reaction force. Both are measured in the frontal plane of the subject. There is an anti-phasic relation between the two curves.

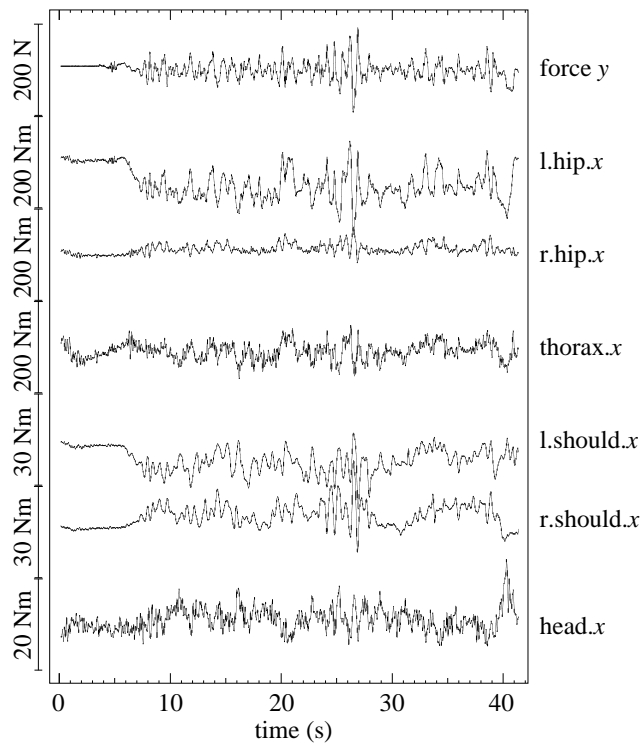


Figure 5. The horizontal component of the ground-reaction force in the frontal plane compared with six main joint moments of force acting in the frontal plane as functions of time. There is a close time relationship between the moments of force and the ground-reaction force.

The profiles of the moments of force acting on six internal degrees of freedom of the human-body simulation are shown in figure 5, together with the ground-reaction force component along the y -axis. Note

the choice in scaling. Three curves show a similar pattern: the ground-force component, the left-hip moment and the thoracic-joint moment. The right-hip moment, the two shoulder-joint moments and the neck moment show an out of phase pattern. All moment profiles have been cross-correlated with the ground-reaction force component in the y -direction, the one acting in the frontal plane. The highest correlation (0.82) was found between the ground-reaction force component and the moment of the hip-joint of the stance leg at the x -axis.

This correlation does not prove that there is a causal relation between the two. A simulation of the forward dynamics of the system does prove it, however.

The moments of force at the x -axis of the shoulder-joints, the hip-joints, the thoracic joint and the head joint were varied by $+50$ Nm and -50 Nm independently. The ground-reaction forces were calculated. The results are depicted in figure 6. The numbers in the thoracic elements show the value in Newtons of the ground-reaction force in the transverse direction. As can be seen, there is an influence on the direction of the ground-reaction force, but also on its magnitude. The influence on the magnitude is directly related to the acceleration of body segments as a result of the change in the joint moment of force. For instance, the fact that the ground-reaction force becomes smaller when a positive moment is applied to the left hip-joint is a result of the falling motion of everything above that hip-joint. The same holds for the other joints.

Changes in the left ankle do temporarily change the direction of the ground-reaction force, but this cancels itself by buckling of the foot's lower-leg complex. The ankle moves sideways and the initial change in direction of the ground-reaction force disappears. The main function of the ankle moment of force is to balance the ankle above

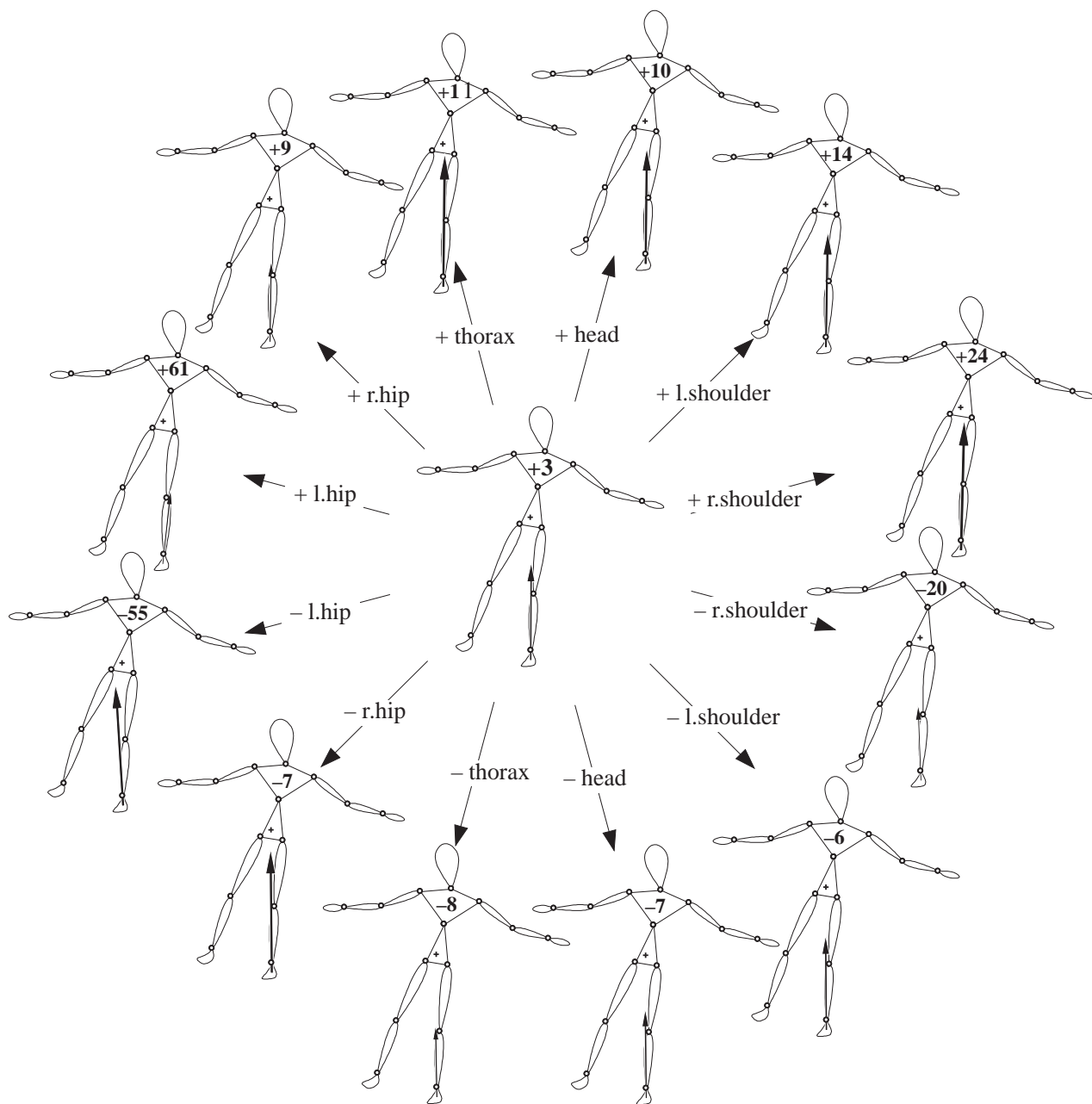


Figure 6. A simulation of the forward dynamics of a balancing movement on a ridge. Changes of +50 Nm or -50 Nm in six main joint moments of force acting in the frontal plane and their effects on the ground-reaction force are shown. The changes in the horizontal component of the ground-reaction force are shown in the thoracic elements in Newtons. The central figure represents one frame from an experiment in which the joint moments of force were derived through simulation of the inverse dynamics.

the centre of pressure. This moment cannot be used for balancing the body as a whole since the base of support is too narrow. Only a small change in moment moves the centre of pressure already to the edge of the ridge. At a further increase of moment, the foot starts rotating.

4. DISCUSSION

(a) *Accuracy considerations*

The ground-reaction forces are measured with a resolution of 0.6 N and calibrated weights showed that the absolute values are within 1%. The centre of pressure is calculated from the measured forces and moments, and remains typically within 1 mm. The error of the centre of mass determination is about 4.5 mm in the frontal plane.

(b) *Biomechanics*

According to figure 6, if a subject wants to have a maximal change in the horizontal ground-reaction force to the right, the moments should all be as negative as possible. A negative moment is counterclockwise in figure 6. The resulting rotations of the segments become as follows: thoracic element, clockwise; arms, counterclockwise; swing leg, counterclockwise; head, counterclockwise. This is not what is done by the subjects. A change in horizontal ground-reaction force to the right is accompanied by clockwise rotations of all segments above the hip-joint of the stance leg.

This paradox can be solved by looking at the synergy of the moments of force. If the moment of force at the hip-joint were the only moment attempting to change the

direction of the ground-reaction force, it would indeed have the effect shown in figure 6. The pelvic element, however, would undergo an angular acceleration of -34 rad s^{-2} . The subject can only keep up such an acceleration for a short time before the deceleration would have to start. If not, the end of the feasible range of motion around the hip-joint would be reached. This range is about 0.2 rad in the counterclockwise direction and 0.4 rad in the clockwise direction given the choice of the stance leg. When the shoulder-joints, the head joint and the swing-leg hip-joint help to keep the pelvic element from accelerating, the angular acceleration becomes only 3.2 rad s^{-2} at the same moment applied at the hip-joint. The subject could maintain that for about three times longer, leading to a maintainable excursion of the centre of mass of ten times further beyond the base of support. That is the solution of the paradox: the shoulder-joints, head joint and swing-leg joint all work in the wrong direction for their effect on the ground-reaction force, but they work in the right direction to keep the pelvic element from accelerating. The subjects kept the whole system of head, arms, trunk and swing leg more or less fixed, so that it could turn as a large segment. This explains the phase relations shown in figure 5.

Since the hip-joint of the stance leg is the most effective one, this joint is doing most of the work, while the other joints take care of the stabilization.

(c) Control

In terms of control it is of interest to look at the time-relation between the moments of force and the ground-reaction force. A cross-correlation analysis showed that all moments of force shown in figure 5 had fluctuations that were within one sample (20 ms) from the ground-reaction force fluctuations. It can be inferred that the activation of the muscles producing the moments of force must have been nearly synchronous. This suggests that a motor programme is used, rather than long-latency reflexes to handle the variations in balance. There are various delays in the system. Mechanical waves of motion over the body may have a delay. Neural pathways show some delay. In addition, there is a considerable electro-mechanical delay of the muscles generating moments of force in the various joints. These delays are variable, depending on the location in the body. If the moments of force were to depend on long-latency reflexes, these variable delays would produce noticeable differences in the onset of moments of force peaks. This is best illustrated when peaks of various muscle groups are compared as a

result of external perturbations of a platform on which subjects stand (Horak & Nashner 1986). These peaks may have a difference in time as long as 50 ms. This difference may be due to variation in neural delays, but may also be caused by delays in mechanical interaction. The perturbations are applied at the feet, and are transported with some delay in the direction of the head. Muscle mechanoreceptors respond to local muscle stretch and so long-latency reflexes may show time-differences. In the present observations, these differences are not present. This implies that self-generated perturbations as in a balance task are handled differently from external perturbations.

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