Several authors have provided evidence that body sway in quiet standing is like the motion of an inverted pendulum pivoted at the ankle joint (Fitzpatrick et al. 1992a,b, 1994a,b, 1996; Fitzpatrick & McCloskey, 1994; Winter et al. 1998; Gatev et al. 1999). Winter et al. (1998) showed that the centre of pressure (COP) and centre of gravity (COG) oscillations for quiet standing fit the equation of motion for an inverted pendulum. Gatev et al. (1999) showed that ankle mechanisms dominate in the sagittal plane with an almost synchronous sway of body parts. Other authors have used more complex models to represent standing (Jacobs, 1997; Lauk et al. 1998; Nicholas et al. 1998; Alonso-Sanchez & Hochberg, 2000), or have disputed the relevance of the ankle strategy and inverted pendulum model in standing (Bloem et al. 2000).

We have investigated the use of the ankle strategy in balancing a large inverted pendulum equivalent in mass and inertia to a human body. What are the advantages of this approach? Standing is a complex activity both mechanically and neurologically. The task of balancing a real inverted pendulum in one plane is much simpler to investigate because there are fewer variables. There is only one joint axis through the ankles and the angular position of the pendulum mass can be precisely measured. Investigation of the strategy used to balance a real pendulum should illuminate the mechanisms used in standing and provide a hypothesis against which standing can be tested.

If the inverted pendulum is to be stabilised, then the change of ankle torque per unit change of angle must on average be greater than the toppling torque per unit angle of the pendulum (the so called ‘gravitational spring’ or ‘load stiffness’). If this were not the case the pendulum would fall to the floor. How are these changes in torque produced? At one extreme the ankles might possess sufficient mechanical stiffness to produce stability in the manner of a tree or tall building. This mechanism would stabilise the pendulum at one angle, effectively converting it into a tall stable object with a narrow base. At the other extreme, the ankles might have zero mechanical stiffness as in the case of a person balancing on stilts. In this case the pendulum would have to be balanced by an impulsive mechanism in the form of intermittent ballistic-like adjustments.

For quiet standing, some authors regard the intrinsic elastic properties of the activated ankle musculature alone as sufficient to achieve stabilisation (like a tall building, above) (Horak & MacPherson, 1996; Winter et al. 1998, 2001). The operation of a sufficiently stiff reflex servo is
also a theoretical solution, but it has been shown that the 
reflex loop has a gain close to unity which is insufficient for 
effective position control based on negative feedback 
(Fitzpatrick et al. 1996). Others regard predictive neural 
modulation of ankle torque to be necessary for quiet 
standing (impulsive mechanism, above) (Morasso & 
Schieppati, 1999). In the present investigation we show that the ankle torque used for balancing the pendulum can 
be apportioned into intrinsic mechanical and neurally 
controlled elements.

It is possible to balance the pendulum (or the body) at any 
reasonable desired position and to move the pendulum or 
body from one position to another. Explanations regarding balance of the human inverted pendulum have 
tended to ignore the problems associated with providing 
stability and control over a range of angles. Many of the 
explanations advanced would produce balance only at a 
single equilibrium point. In suggesting an answer to the 
question of positional control we show that control of the 
pendulum is necessarily associated with repeated, ballistic-
like patterns of ankle torque change vs. angle. We further 
show that the neural modulation associated with this 
positional control scheme increases the operational 
stiffness and provides intermittent, reactive damping. This 
activity 'tops up' the intrinsic mechanical contribution of 
the active ankle musculature which, on its own, is not quite 
sufficient to counteract the 'gravitational spring'.

Finally, our previous research has shown that the mean 
sway size of the pendulum could be systematically reduced 
but this result was not achieved by increasing the change in 
ankle torque per unit angle (Loram et al. 2001). This result 
was contrary to theories that sway is altered by controlling 
operational ankle stiffness or viscosity (Fitzpatrick et al. 
1992a, b; Winter et al. 1998; Carpenter et al. 1999; Gatev 
et al. 1999). In this paper we investigate how the pendulum 
sway size was minimised not by making the 'tall building' 
stiffer, but by refining the performance of the impulsive 
mechanism, by improving the accuracy of intermittent, 
reactively triggered, ballistic patterns of torque.

**METHODS**

Some of the methods used have already been reported fully 
(Loram et al. 2001) so the essential and additional features only 
will be reported here.

**Subjects**

Ten healthy people, of whom six were male, aged between 18 and 
45 years took part in this study. The subjects gave written 
informed consent, and the study was approved by the local human 
ethics committee and conformed to the principles of the 
Declaration of Helsinki.

**Apparatus**

Subjects were strapped round the pelvis to a vertical support that 
effectively eliminated their actual sway (Fig. 1). They stood on two 
footplates with their ankles positioned to be co-axial with the axis 
of rotation of an inverted pendulum. Each subject balanced the 
inverted pendulum, which was free to move forwards and 
backwards in a parasagittal plane while always tending to topple 
backwards. The pendulum had a mass of 61.65 kg with a centre 
supported 0.937 m from the axis of rotation (distance h). The 
topping torque per unit angle of the pendulum was measured to 
be 10.2 ± 0.4 N m deg⁻¹ and the moment of inertia was 
62.6 ± 2 kg m² (means ± S.D.). The same mass and distance (h) 
was used for all subjects.

The angular position of the pendulum, the ankle torque from each 
leg and the soleus and tibialis anterior surface EMGs from each leg 
were recorded. The angular velocity and angular acceleration of 
the pendulum were calculated by successive differentiation of the 
position signal using Savitzky-Golay filters (Press et al. 1999). 
EMG activity was recorded using bipolar surface electrodes with 
capsulated preamplifiers. These signals containing the entire 
bandwidth were then amplified, passed through an analog full-
wave rectifier and root mean square (r.m.s.) averaging filter with a 
time constant of 100 ms, and then sampled at 25 Hz and stored on 
hard disk. The sampled EMG level corresponding to the ‘noise 
floor’ was 0.04 V.

**Experimental protocol**

Subjects were asked to balance the inverted pendulum under two 
different instructions. Sometimes they were asked to reduce the 
sway of the pendulum to an absolute minimum and to keep it at a 
constant angle. This intention was called 'stand still'. Alternatively, 
they were asked to balance the pendulum while giving the least 
possible attention to the sway of the pendulum. This intention was 
called 'stand easy'. In both cases the subjects were asked to keep 
the pendulum between 0.5 and 5.5 deg from the vertical so as to 
approximate standing sway. Both instructions were carried out 
one using visual feedback, which was provided using an 
osilloscope, and once without using visual feedback. Thus there 
were four separate trial conditions:

1. 'stand still using visual feedback';
2. 'stand still with no visual feedback';
3. 'stand easy using visual feedback';
4. 'stand easy with no visual feedback'.

The order in which the four trials were carried out was 
randomised. The duration of each trial was 200 s.

When subjects were 'standing still' they were encouraged to give 
their full attention to the oscilloscope when it was turned on and 
and to give full attention to what they could register through their legs 
when the oscilloscope trace was blanked. When subjects were 
'standing easy' they were engaged in meaningful conversation to 
take their mind off the task as much as possible.

In this investigation we studied the effect of subject intention on 
pendulum movement and were less concerned with the changes 
produced by altered sensory input. So in subsequent analysis we 
concentrate mainly on the clear changes produced by altered 
intention by pooling conditions (1) and (2) and conditions (3) 
and (4).

**Principles and methods of data analysis**

During balancing, the pendulum sways to and fro in a quasi-
regular fashion. The unidirectional movement between one 
reversal point and the next was categorised as a sway. For any trial, 
the mean sway size was the average magnitude of the sways and the
mean duration was the average duration of the sways. A two-way analysis of variance (ANOVA) was performed to test for differences in the mean values between different conditions. Post hoc, Tukey, pairwise comparisons were made using 95% simultaneous confidence intervals.

Inspection of the torque and EMG time records shows these quantities are clearly related to the motion of the pendulum. Patterns of activity appear to recur but never in exactly the same way twice. Our aim was to look for underlying processes so we decided to average data surrounding spontaneous departures from equilibrium as a means of eliminating inconsequential randomness and identifying recurring patterns.

In each unidirectional sway a spring-like equilibrium occurs at least once. This was represented by a positive gradient of torque vs. angle crossing the line of equilibrium where the line of equilibrium is defined by the gravitational torque acting on the pendulum at each angle. The speed was maximal at these instants. The data surrounding these equilibrium points were averaged to show the mean responses. This method of identifying equilibrium points, sampling around these points and averaging is illustrated in Fig. 2.

When the subject was keeping the pendulum still, there were more small, slow sways and less large, fast sways than in the ‘easy’ condition (Fig. 4 in Loram et al. 2001). In order to compare spring-like line crossings of the same velocity from trials under different conditions these line crossings were binned into sways of different velocity using the velocity at equilibrium to categorise them. The velocity bin boundaries were 0.0, 0.0625, 0.125, 0.25, 0.50, 1.0 and 2.0 deg s\(^{-1}\) and these were chosen to equalise the numbers grouped into each bin for each trial condition as far as possible. For each spring-like line crossing we calculated the acceleration (effectively the torque error) at the end of that sway. For each bin, the average acceleration at the end of the sway was calculated.

We also wanted to analyse the effect of initial torque error on the size and duration of the following sway. Each sway was then binned according to its initial acceleration (torque error). The acceleration bin boundaries were 0.0, 0.25, 0.50, 1.0, 2.0 and 4.0 deg s\(^{-1}\), and again these were chosen to approximately equalise the numbers in each bin for each condition. The mean sway size and sway duration associated with each bin was then calculated. The mean sway size and duration to the first line crossing equilibrium (necessarily positive gradient) were also calculated.

For each binned analysis described above, the dependent parameter’s absolute values from the rising and falling cases were used. For each parameter and for each bin this generated 80 values from 10 subjects, each performing trials under the four conditions described above. A two-way analysis of variance (ANOVA) was performed to test the effect of intention (‘still’ vs. ‘easy’) for each of the parameters for each bin.

Modelling

We wanted to decompose the averaged line crossing torque records into intrinsic elastic and neural components. The model and procedure that we used for this are described in Appendix A.

For the purpose of normalising the soleus EMG signals between subjects, each subject was asked to balance the pendulum still at every half degree between 1 and 5 deg for 20 s. The mean soleus EMG and torque signals at each angle were calculated. The regression line between combined ankle torque and combined soleus EMG gave a neural gain in newton metres per volt (N m V\(^{-1}\)) relative to which the decomposition model neural gain of each subject could be normalised.

RESULTS

The quasi-regular variation of ankle torque with pendulum position showed the same general characteristics with all subjects under all conditions. A representative example is shown in Fig. 3A. The pendulum was not confined to one angle and there was no one position of equilibrium. Rather the pendulum made small sways to and fro of irregular duration, size and speed. Movement of the pendulum was controlled by an ankle torque that always kept close to that required for balance and which attained equilibrium transiently every time the line of equilibrium was crossed. Most of these line crossings were spring-like (torque increases as angle increases) with a positive gradient. Less commonly, negative gradient line crossings could also be seen (X).
When subjects gave their full attention to keeping the pendulum still, as opposed to giving minimal attention to keeping it still, there was a clear reduction in mean sway size (Fig. 3B). A significant reduction in mean sway size occurred both when visual feedback was available (condition 1 vs. 3) and when it was not (condition 2 vs. 4) and generally the effect of intention was significant (two-way ANOVA, effect of intention, \(n = 40, F = 10.0, P = 0.003\)). Interestingly, there were no significant differences in mean sway duration between any of the four trial conditions (Fig. 3C) (two-way ANOVA, \(n = 40, F = 0.61, P = 0.61\)). Combining both results, the mean sway velocity mirrored the mean sway size for all four conditions.

In balancing the pendulum, an underlying process was repeated over and over again. Figure 4 shows the entry into and departure from equilibrium averaged for all occurrences for all 10 subjects. The pendulum falling, positive-gradient cases were selected under 'still' and 'easy' conditions with visual feedback available (condition 1 vs. 3). Looking at the ankle torque vs. angle plots (Fig. 4A) we see the same biphasic 'drop and catch' pattern for both conditions as the pendulum swayed from one reversal point (a) to the next (b). If torque error is the difference between torque applied and that required for equilibrium, then from a there was initially an increase in torque error (the drop), followed by a decrease in torque error leading to equilibrium and the maximum speed of the pendulum (starred). This was followed by an opposite increase then decrease in torque error which decelerated the pendulum to rest at b (the 'catch').

Changes in ankle torque during an individual sway relate to both neural modulation and changes in ankle angle. A notional best-fit line through the changes in torque vs. angle (Fig. 4A) would indicate a generally spring-like characteristic with a gradient approximately twice that of the line of equilibrium. Yet on top of that there are changes in torque which are clearly not spring-like in origin. The decrease in torque immediately after the sway begins at a was not caused mechanically/elastically because the muscle–tendon was being stretched at this point (dorsiflexion). Neither was mechanical viscosity the cause because the speed was increasing. In any case the mechanical and reflex viscosity (~0.02 and ~0.07 N m s deg\(^{-1}\), respectively, per leg operating at 15 N m; Mirbagheri et al. 2000) are too small to produce major changes in torque for these averaged sways, which only reached

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**Figure 2. Illustration of the line crossing averaging process**

A, a 6 s record of angular velocity and angular acceleration against time for a representative subject. Equilibrium times were identified by interpolating between the pairs of acceleration data points that cross zero. From these equilibrium times, those that occurred while the acceleration was passing from positive to negative and while the velocity was positive (i.e. the pendulum is falling) were selected. These equilibrium times are shown as an asterisk. Ankle torque and pendulum position records were sampled at 0.04 s intervals before and after these selected equilibrium times. The four selected equilibrium times in A are shown in B, together with \(\pm 0.48\) s of surrounding data, plotted as ankle torque against pendulum position. The straight dashed lines represent the line of equilibrium defined by the gravitational torque acting on the pendulum. The selected equilibria represent falling (increasing angle), spring-like (positive-gradient) line crossings as indicated by the arrow in C. The four 0.96 s records shown in B are averaged to produce the record shown in C. In a 200 s trial, over 100 examples would be averaged. The rising, positive-gradient line crossings were selected and averaged in an analogous manner.
Figure 3. Pendulum sway

A, a 12 s record from one subject is plotted as combined ankle torque against pendulum position. Data points are at 40 ms intervals. The starting point (diamond) and finishing point (square) are indicated. The line of equilibrium (gravitational torque on the pendulum) is shown as a continuous straight line. For each trial condition, B shows the mean sway size and C shows the mean sway duration. For both panels, values were averaged over 10 subjects for each of the four trial conditions. Error bars show 95% simultaneous confidence intervals for the mean values. As described in Methods, a sway is the angular movement between successive reversal points of the pendulum. Trial conditions were (1) stand still with visual feedback, (2) stand still with no visual feedback, (3) stand easy with visual feedback, and (4) stand easy with no visual feedback.

Figure 4. Effect of intention on averaged spring-like line crossings

Averaged data are shown from 1 s before to 1 s after all positive-gradient, equilibrium line crossings while the pendulum was falling. These data were averaged over all 10 subjects in the ‘still’ (1) and ‘easy’ (3) conditions while visual feedback was available. A, combined torque from both legs vs. pendulum position. B, combined soleus EMG from both legs vs. pendulum position. EMG data were normalised between subjects. Data points are at 40 ms intervals and proceed from reversal points a to b via the arrow. The line of equilibrium (ignoring pendulum friction) is shown as the continuous straight line in A. The asterisk marks the instant of equilibrium and maximum velocity.
maximum speeds of 0.2 and 0.5 deg s\(^{-1}\), respectively. Thus the decrease in torque must have been caused by neural modulation and this is confirmed in Fig. 4B where it can be seen that the soleus EMG was decreasing in the 140 ms preceding the reversal point at \(a\). The nervous system did not anticipate the initiation of the sway as the soleus EMG only rose as the sway began at \(a\), leading to a delayed increase in ankle torque that will catch the falling pendulum.

Similarly, the decrease in torque after equilibrium (star) and before the reversal point (\(b\)) was also not mechanically elastic or mechanically or reflexly viscous in origin for the same reason as above. This decrease in torque corresponded to the decrease in soleus EMG after equilibrium (star) followed by an inevitable electromechanical delay of approximately 140 ms. Note that this decrease in soleus EMG anticipated the end of the sway, indicating predictive behaviour on the part of the nervous system.

The averaged soleus EMG showed the same pattern in each leg for every trial of every subject. Tibialis anterior EMG was usually close to the noise floor with no sign of modulation. Sometimes, such as when the subject was fatigued or when the pendulum swayed close to the vertical, the tibialis anterior EMG would be modulated either antagonistically with soleus or synergistically with soleus and in these cases the tibialis anterior signal could be comparable in magnitude to that of soleus. However, this modulation was not consistent from trial to trial, from subject to subject or even from leg to leg. The lack of averaged modulation in tibialis anterior compared with soleus is shown in Appendix A (Fig. 8).

One can clearly see that the sway size between \(a\) and \(b\) was less for the ‘still’ condition than the ‘easy’ condition. There was no change in line crossing gradient, which indicates no change in the operational stiffness or viscosity of the ankle mechanisms (Loram \textit{et al.} 2001). Since the dots occurred at 40 ms intervals it was clear that through the line crossing there is a smaller rate of growth of torque error per second in the still condition and this would have minimised the absolute acceleration of the pendulum. It can also be seen that at the reversal point (\(b\)) there was less torque error in the ‘still’ condition than the ‘easy’ condition and this

![Figure 5. Decomposition of ankle torque according to our model](image)

The decomposition of ankle torque into intrinsic elastic and neurally modulated components is shown. The model is described in Appendix A. A–C, illustrative data averaged from all subjects for falling, positive-gradient equilibrium line crossings in condition 3. A, the changes in torque arising from stretching and releasing of the elastic components. B, the variation in torque resulting from neural modulation (continuous line) and the preceding variation in soleus EMG (dotted line). C, the actual variation in torque as well as the modelled variation (dotted line). A–C, the same position, EMG and torque data as Fig. 4 (‘easy’). Data points are at 40 ms intervals and the reversal points \(a\) and \(b\) correspond to those in Fig. 4. The model was applied to averaged, positive-gradient, line crossing data from each trial. Values of parameters for falling and rising line crossings were averaged. For each trial condition, D shows the mean intrinsic mechanical stiffness, E shows the mean neural gain and F shows the mean electromechanical delay between changes in soleus EMG and changes in torque. The neural gain is expressed relative to the isometric neural gain (N m V\(^{-1}\)). Parameter values were averaged over 10 subjects for each of the four trial conditions. Error bars show 95% simultaneous confidence intervals for the mean values.
would have altered the initial acceleration of the subsequent sway. These two observations are relevant to sway minimisation but are complicated by the confounding fact that the ‘still’ and ‘easy’ conditions contain different distributions of sway velocities. In order to contrast like with like, sways of equal velocity should be compared.

For the averaged spring-like line crossing data, changes in ankle torque have been decomposed into changes that result from neural modulation and changes that derive from mechanical stretching and releasing of the elastic structures surrounding the ankle joint (activated muscle fibres, aponeurosis, tendon, foot). For illustration, Fig. 5C shows the actual changes in ankle torque for trial condition 3 averaged over all 10 subjects (cf Fig 4A, ‘easy’). The changes in torque predicted by the model are also shown (percentage variance accounted for, %VAF = 98.5%). The predicted changes in torque resulting from stretching of the activated elastic structures and neural modulation are shown in Fig. 5A and B, respectively. It can be seen that neural modulation makes the greatest contribution to changes in torque and adds operational stiffness to the torque changes at the spring-like line crossing. Neural modulation also adds changes in torque orthogonal to changes in position that cause additional acceleration at a (the ‘drop’) and cause additional braking as b (the ‘catch’) is approached.

The average intrinsic mechanical stiffness of all subjects was −8 ± 4 N m deg⁻¹ (±S.D.) and this value is just less than the gravitational toppling torque per unit angle of the pendulum (10.2 N m deg⁻¹, Fig. 5D). The averaged neural gain of all subjects was −0.5 ± 0.3 (±S.D.), which was expressed relative to the isometric gain in newton metres per volt measured for each subject (Fig. 5E). This shows that small rapid fluctuations in EMG produce relatively less change in torque than the large slow changes in EMG that were required for the isometric calibration. The average electromechanical delay between changes in EMG and changes in torque was 140 ± 40 ms (±S.D.) (Fig. 5F).

When subjects were minimising pendulum movement they minimised the torque error as the pendulum was brought to rest at the end of a ‘catch’. Figure 6A shows that at all velocities, apart from the lowest, the torque error

![Figure 6. Effect of intention on binned sways](image)

A, the effect of intention on sways sampled and grouped according to their velocity at the first positive-gradient line crossing. A, the acceleration at the end of the sway. B and C, the effect of intention on sways sampled and grouped according to their initial acceleration. B, the sway size to the first positive-gradient line crossing equilibrium and to the reversal point at the end of the sway. C, the duration to the first line crossing and to the end of the sway. For all panels the ‘still’ results (crosses on continuous line) and the ‘easy’ results (dots on dashed line) were averaged over falling and rising sways and over with and without visual feedback for all subjects. The abscissa values are the mean binned values. The error bars represent 95% confidence intervals in the mean ordinate values for each bin. Two-way ANOVA for ascending bins in A gives n = 80, F = 1.0, 8.7, 16.1, 9.7, 11.7, 9.1, P = 0.3, 0.004, 0.0001, 0.003, 0.001, 0.003.
when the pendulum was first brought to rest following the line crossing was significantly and substantially less in the ‘still’ condition. The mechanical consequence is that the initial acceleration of ‘the sway after the catch’ will be less in the ‘still’ condition than the ‘easy’ condition.

To assess the benefit of minimising the initial acceleration of a sway, we needed to sample sways according to their initial acceleration (torque error). Figure 6B shows clearly that on average the size of a complete sway (and the sway to equilibrium) increased with the initial acceleration for both the ‘still’ and ‘easy’ conditions. For the three lowest acceleration bins the difference between the conditions was not significant, though taking all five bins together, the fact that the sway sizes were always less for the ‘still’ condition is significant (n = 5, P = (½)⁵ = 0.03). This figure confirms that in minimising pendulum movement there is a benefit from minimising the initial acceleration of a sway. Figure 6C shows that for all initial accelerations the duration of a complete sway was virtually unchanged at 1 s, and the duration till equilibrium was unchanged at 0.4 s; the intention of a subject made no difference to either of these times. Clearly, in the ‘still’ condition there was an improvement in the efficacy but not the rapidity of the movement minimising process.

The size of a sway was clearly associated with the maximum speed of the pendulum at the spring-like line crossing in the middle of the sway and this relationship was unaffected by the intention of the subject (Fig. 7A). This result is unsurprising given the large inertia of the pendulum. After each spring-like line crossing the pendulum was eventually brought to rest and then there was another sway in the opposite direction. By calculating the size of the subsequent sway in the reverse direction one sees a fascinating result (Fig. 7B). For each velocity bin, the subsequent sway size in the opposite direction was significantly and substantially less in the ‘still’ compared to the ‘easy’ condition. The intention of the subject to minimise movement had great effect by minimising the

Figure 7. Effect of intention on the current sway and subsequent return sway

For every sway the pendulum starts from transient rest, passes a positive-gradient line crossing (a speed maximum) and ultimately comes to a reversal point where it changes direction. For each bin, A shows the mean size of a sway vs. the velocity at the first positive-gradient line crossing. After the reversal point, the pendulum executes a return sway in the opposite direction to the current sway. B (continuous and dashed line) shows the mean size of the subsequent return sway vs. the velocity at the first positive-gradient line crossing of the current sway. The dotted line shows the size of the return sway calculated by interpolation from Fig. 6A and B. The ‘still’ results (crosses on continuous line) and the ‘easy’ results (dots on dashed line) were averaged for all subjects over falling and rising sways including with and without visual feedback conditions. The abscissa values are the mean binned values. The error bars represent 95% confidence intervals in the mean ordinate values for each bin. Two-way ANOVA for ascending bins in B gives n = 80, F = 7.3, 7.3, 11.5, 8.9, 11.2, 11.0, P = 0.008, 0.009, 0.001, 0.004, 0.001, 0.001. C, for a range of current line crossing speeds the continuous line (tot.) shows the total difference in the size of the subsequent return sway caused by the intention of the subject. The dashed (i. a.) and dotted (oth.) lines show respectively the component differences caused by reducing the initial acceleration of the subsequent sway and by other minimisations occurring during the subsequent sway. All three lines were calculated by interpolation from Fig. 6A and B, not Fig. 7B.
initial acceleration and maximum speed of the ‘rebounding’ sway. Figure 7C shows that for all speeds apart from the very slowest, half the minimisation in sway size resulted purely from minimising the initial acceleration of the sway; and half also resulted from the intention of the subject during the whole course of the subsequent sway.

**DISCUSSION**

The aim of this paper is to illuminate (i) how movement of the pendulum is controlled in general terms, (ii) the relative contributions of mechanical/elastic and neural modulation of ankle torque and (iii) how pendulum sway is intentionally minimised.

(i) **Control of pendulum movement**

Control of the pendulum requires a subject to be able to change its position when required as well as to restrict sway when it is at the intended angle. Movement of the pendulum from one rest position at equilibrium to another such position necessarily requires a biphasic pattern of torque which cannot be achieved by spring-like (torque increases as angle increases) torques alone, which would produce stability only at one angle. As far as we know this point has not previously been clearly stated. We call this pattern of torque a ‘drop and catch pattern’ if the pendulum is being lowered and a ‘throw and catch pattern’ if the pendulum is being raised. An example of this pattern is shown in Fig. 9 where the pendulum starts at rest in perfect balance. The pendulum is dropped, reaches maximum speed as equilibrium is regained and is brought to rest, perfectly in balance at the final equilibrium position. During the drop the torque must be decreased below that required for equilibrium to generate movement. During the catch the torque must be increased above that required for equilibrium to decelerate the pendulum (Fig. 9C). Since the angle of the pendulum has increased, the final level of torque is greater than the initial level. In a perfect catch, the pendulum is brought simultaneously to rest and balance. The pendulum would then never move from that position until the torque is changed. In reality, subjects never attain perfect balance and rest simultaneously (Fig. 4A) and therein is the cause of their subsequent sway. (The pattern can also be seen as standing subjects move to a new angle – Fig. 1 (Gurfinkel et al. 1974).)

Subjects use this process to control movement of the pendulum in a manner quite unlike a feedback servo. Movement of the pendulum proceeds as a series of steps. Each step represents a shift from one momentary rest position of imperfect balance to another such position (Figs 4A and 5C). Each step follows an approximation to the ideal drop and catch or throw and catch pattern (Fig. 9A–C). Generally a positional step in one direction is followed by a step in the reverse direction (alternating steps). Less commonly there occurs a repeated approximation to the ideal pattern (additive steps) (Fig. 9D), for example as shown by the X in Fig. 3, which permits additive positional change while keeping the speed low. Individual biphasic drop and catch patterns have some resemblance to a damped oscillator, but the accumulation of alternating and additive steps results in a low frequency positional drift. This may accord with the knowledge that positional changes in quiet standing follow a third order model comprising an underlying first order decay and a damped spring oscillator all buffeted by noise (Johansson et al. 1988; Kiemal et al. 2001).

(ii) **The mechanical and neural modulation of torque**

What is the relationship between the intrinsic elastic and neural origins of changes in ankle torque? As seen in Fig. 4, changes in ankle torque are clearly related to changes in both position and neural modulation. We have interpreted the changes in torque as a linear combination of a mechanical elastic component and a neural component with a time delay (Fig. 5). The basis for this decomposition is given in Appendix A. On this interpretation, the mean intrinsic mechanical stiffness for all subjects (±8 ± 4 N m deg⁻¹, ± S.D.) is less than the minimum stiffness required for stability (the toppling torque per angle of the pendulum (10.2 N m deg⁻¹). This corresponds with the personal experience that simply applying a suitable level of torque with its associated ankle stiffness does not stabilise the pendulum. For the mean level of ankle torque and mean pendulum sway size, our figures are in broad agreement with published measurements of intrinsic ankle stiffness. Published values quoted per leg at 15 N m (Hof, 1998) and 4–6 N m deg⁻¹ (Hunter & Kearney, 1982; Kearney & Hunter, 1982; Mirbagheri et al. 2000).

This interpretation implies that the mechanical stiffness of the activated muscles in combination with the tendon and the foot is large enough to make a real contribution to stabilising the pendulum and is almost sufficient to cancel the positive feedback effects of gravity acting on the pendulum. The implication is that the nervous system is controlling a nearly balanced pendulum that can be accelerated in any direction by modulating the muscle activity appropriately. Our conclusion is intriguingly similar to the result of Fitzpatrick et al. (1996). They found that the feedback loop gain in standing is close to unity, which is too low for a position-based negative feedback system. They concluded that a feedforward predictor was in operation to stabilise sway.

The electromechanical delay between soleus EMG and ankle torque (140 ± 40 ms, ± S.D.) is less than the advance of 250–300 ms between lateral gastrocnemius EMG and COP and COG (which were maximally correlated in
phase) measured by Gatev et al. (1999) but accords with their finding of predictive control. Our delay corresponds with a value of ~140 ms calculated for soleus from Fig. 2 in Bawa & Stein (1976). The modulation of soleus EMG shown in Fig. 4B is directly contradictory to published theories that the CNS is not involved in regulating balance on a sway by sway basis (Horak & MacPherson, 1996; Winter et al. 1998, 2001). As suggested by Morasso & Schiepatti (1999), neural modulation of ankle torque gives additional operational stiffness and a pattern of torque that can be changed without reference to position to bring about impulsive control of the pendulum.

(iii) Minimisation of pendulum swing

The third question concerns how pendulum sway is minimised intentionally by the subject. When sways are sampled on the basis of their line crossing (maximum) velocity one can see that sways of larger size are associated with a larger maximum speed and that this relationship is not affected by the intention of the subject (Fig. 7A). However, the torque error at the end of the sway is substantially and significantly reduced when the subject is intending to minimise pendulum movement (Fig. 6A). The effect of this minimisation of torque error at the reversal point is to minimise the initial acceleration and consequently the size of the subsequent returning sway (leftward shift of operating point on Fig. 6B). This mechanism alone can account for half the reduction in sway size in the ‘still’ condition compared with the ‘easy’ condition (Fig. 7C). The remaining reduction occurs during the subsequent sway. For a given initial acceleration the sway of the pendulum is minimised (vertical difference on Fig. 6B). In a ballistic-like manner the first mechanism reduces the subsequent sway by setting the starting conditions. The second mechanism operates continuously over the sway. For both mechanisms, the reduction in sway size is not accompanied by a reduction in duration (Fig. 6C).

In minimising pendulum movement, the most important single goal is to bring the pendulum to rest as close to equilibrium as possible using an appropriate pattern of torque. This will reduce the size of the subsequent sway by reducing the initial acceleration. This is in large part the answer to the question of how movement was minimised without increasing operational ankle stiffness or viscosity. This pattern of torque cannot be produced by elastic mechanisms alone. It requires anticipatory skill from the subject who must produce a predictive pattern of neural modulation that estimates the load properties and compensates for the inherent properties of the ankle musculature, the Achilles’ tendon and the elasticity of the foot (Partridge, 1979). In Fig. 4B, the predictive neural modulation of soleus is shown. The anticipatory decrease in soleus EMG occurs just after the maximum velocity indicated by the star. This decrease in EMG results in the subsequent decrease in torque while the muscle and tendon are still being stretched and as the pendulum is brought to transient rest. This predictive process occurs when the subject is paying minimal attention to the task as well as when close attention is given.

Predictive modulation may be large, in which case the onset of the next reversal can be easily predicted. However, in the case of fine balance at low speeds, the timing of the next reversal is probably unpredictable. A small difference in torque pattern could cause the pendulum to either reverse direction or continue in the same direction with reduced speed. Examination of the neural modulation of soleus (Fig. 4B) shows that as the pendulum sways from rest there is no anticipatory rise of the EMG before the sway begins. This indicates that whereas the end of the sway is anticipated, the beginning of the sway is not. The increase in torque error at the beginning of a sway (Fig. 5B) is a consequence of lack of anticipation of a sway. Thus drops (and throws) are on average accidental, the neurally modulated catches are reactively triggered and the completion of the catch and setting up of the next sway is predictive with a certain error.

In conclusion, (i) we have identified a biphasic, ballistic-like pattern of torque (‘throw and catch’) that is repeatedly used to control the position of the pendulum. (ii) The biphasic torque pattern necessarily requires neural modulation of ankle torque since mechanical, elastic forces alone cannot generate it. The activity of the soleus muscle is modulated in such a manner that it will produce this pattern of torque. At the same time, the intrinsic mechanical stiffness of the activated triceps surae in combination with the Achilles tendon and foot looks capable of nearly compensating for the gravitational torque without providing positional control. (iii) Sway is minimised by optimising the accuracy of the ballistic-like pattern of torque thus bringing the pendulum to rest as close to balance as possible. In this manner the size but not the duration of the subsequent sway is reduced. Predictive torque accuracy is the key factor in reducing pendulum movement.

If sway in quiet standing approximates the simple inverted pendulum model controlled by the ankle musculature as proposed by Winter et al. (1998) and Gatev et al. (1999), then the same balancing process should be evident as when controlling a real inverted pendulum (cf. Fig. 2 in Winter et al. 2001). So, we hypothesise that if centre of pressure and centre of mass data are examined, then the three conclusions identified in this paper will be observed. Otherwise, the model for quiet standing of a simple, inverted pendulum controlled by an ankle strategy should be discarded.
APPENDIX A

Decomposition of torque into mechanical/elastic and neural components (Fig. 5)

We assume that the time varying ankle torque, \( T(t) \), is a function only of ankle angle, \( \theta \), and soleus EMG, \( E \), with an electromechanical delay, \( \tau \), between changes in EMG and torque:

\[
T(t) = T(\theta(t), E(t - \tau)).
\]

We have eliminated velocity as a variable because the mechanical and reflex viscosities provide little effective torque at the slow speeds of pendulum sway (Mirbagheri et al. 2000). The averaged tibialis anterior EMG showed no consistent pattern and was therefore on average not responsible for producing changes in torque (Fig. 8A). Gastrocnemius EMG was not recorded for this data set though in the course of a different experiment, recordings on 12 subjects revealed a similar averaged pattern to the soleus EMG and a tibialis anterior signal that barely changed from the background noise level (Fig. 8B).

We assume that changes in ankle angle and soleus EMG are independent of each other. We also assume that for small changes in angle and EMG, the torque function can be expanded as a first order Taylor series about a constant, unspecified operating torque, angle and activation level (Hogan, 1990):

\[
\frac{dT}{dt} = \left( \frac{\partial T}{\partial \theta} \right|_{\theta} \frac{d\theta}{dt} + \frac{\partial T}{\partial E} \frac{dE}{dt} \right). 
\]

The coefficients of the linear terms are intrinsic angular stiffness:

\[
K = \frac{\partial T}{\partial \theta} \bigg|_{\theta}
\]

(change in torque per unit angle at constant EMG), and isometric neural gain:

\[
c = \frac{\partial T}{\partial E} \bigg|_{\theta}
\]

(change in torque per unit EMG at constant angle).

Hence:

\[
\frac{dT}{dt} = K \frac{d\theta}{dt} + c \frac{dE}{dt}.
\]

We use this equation to predict the rate of change of torque \( (dT/dt) \) from the pendulum velocity \( (d\theta/dt) \) and the rate of change of soleus EMG \( (dE/dt) \) with an electro-

Figure 8. Comparison of soleus, tibialis anterior and gastrocnemius EMG

A and B, averaged data from 1 s before to 1 s after all positive-gradient, equilibrium line crossings while the pendulum was falling. Data points are at 40 ms intervals and proceed from reversal points a to b via the arrow. The asterisk marks the instant of equilibrium and maximum velocity. A, soleus (S) and tibialis anterior (TA) EMG from both legs vs. pendulum position. These data were averaged over all 10 subjects and over all 40 trials including ‘still’ and ‘easy’ conditions both with and without visual feedback. B (for one leg only), soleus (S), tibialis anterior (TA), gastrocnemius medialis (GM) and gastrocnemius lateralis (GL) EMG vs. pendulum position as well as ankle torque vs. pendulum position. This data set is different from the one reported in the body of this paper although the methods used were identical. Twelve subjects were asked to balance the pendulum for 200 s.
mechanical delay between changes in EMG and changes in torque. We interpret the angular stiffness $K$ as reflecting the intrinsic, mechanical stiffness of the ankle including muscle fibre, aponeurosis, tendon and foot (Gurfinkel et al. 1994). We appreciate that these components have non-linear stiffness and that the muscle torque does change by $\pm 3\text{--}10\%$ of the mean level but we apply our method to estimate an average value for the changes in torque to which the model is applied. Equally, an average value will be estimated for the neural gain while in reality this variable is likely to be non-linear and history dependent. The time delay between decreasing soleus EMG and torque is likely to be longer than the time delay between increasing soleus EMG and torque.

This model was applied to the averaged spring-like line crossing changes in torque, angle and soleus EMG. Data from 0.5 s before and after the line crossing were used. The rates of change of torque, angle and soleus EMG with time were calculated using a Savitzky-Golay filter (Press et al. 1999). The coefficients $K$, $c$ and $\tau$ were optimised using a least squares simplex search method. For all 40 trials, for both the falling and rising cases, there was only one minimum in the parameter space defined by the ranges $0 < K < 30 \text{N m deg}^{-1}$, $0 < c < 800 \text{N m V}^{-1}$, $0 < \tau < 0.4 \text{s}$. This was established by starting searches from a variety of locations including the eight vertices of the above three-parameter space.

Two methods were used to estimate confidence intervals for the coefficients. First there was the comparison of results from multiple data sets. The 80 values of $K$, $c$ and $\tau$ from the 40 trials varied with respective standard deviations of $\pm 4 \text{N m deg}^{-1}$, $\pm 0.3$ (relative to the isometric value in N m V$^{-1}$) and $\pm 40 \text{ms}$. Second, using a combined Gauss-Newton/Levenberg-Marquardt search method, the Jacobian matrix at the least squares minimum was estimated and 95% confidence intervals in the coefficients were calculated from this matrix and the residuals (Bates & Watts, 1988). From all 80 results, the mean 95% confidence intervals were, $\pm 3 \text{N m deg}^{-1}$, $\pm 0.09$ and $\pm 20 \text{ms}$ for coefficients $K$, $c$ and $\tau$, respectively.

We wished to assess the goodness of fit between the measured rate of change of torque ($y$) and the rate of change of torque calculated from the model ($Y$). The percentage variance in $y$ accounted for ($\%\text{VAF}$) by the model was calculated by:

$$\%\text{VAF} = 100 \times 1 - \frac{\sum_{1}^{N} (y - Y)^2}{\sum_{1}^{N} y^2},$$

where $N$ is the number of points.

For all 80 results, the mean $\%\text{VAF}$ was 97%. Since this decomposition rests on assumptions, the results need to be

Figure 9. An ideal, perfect drop and catch pattern
A–C, a perfect biphasic drop and catch pattern that will move the pendulum from rest and equilibrium at one position to rest and equilibrium at a new position. $A$, pendulum position vs. time. $B$, torque vs. time. $C$, torque vs. position. $D$, a double drop and catch pattern. Points proceed at 40 ms intervals in the direction of the arrow. The dashed line is the line of equilibrium. The asterisk indicates the positive-gradient line crossing equilibrium which is the instant of maximum speed. The derivation is given in Appendix B.
tested by direct measurement of the intrinsic, mechanical stiffness of the ankle joint during standing and balancing the pendulum.

APPENDIX B

Derivation for a perfect drop and catch pattern (Fig. 9)

The equation of motion for the pendulum is:
\[
I \frac{d^2 \theta}{dt^2} = mg h \theta + T,
\]
where \( I \) is the moment of inertia, \( \theta \) is the pendulum angle, \( m \) is the mass, \( g \) is the gravitational field strength, \( h \) is the distance of the mass from the axis of rotation, \( T \) is the torque applied to the pendulum by the subject and \( t \) is time.

Consider the pendulum to be at equilibrium at maximum speed, i.e. at: \( t = t_e, \theta = \theta_0, \nu = v_0, T = -mg h \theta_0 \).

For a perfect catch the subject has to bring the pendulum to rest at equilibrium, so for the final conditions at the end of the sway: \( t = t_e, \theta = \theta_0, \nu = 0, T = -mg h \theta_0 \).

We will consider that the subject applies a torque that balances the gravitational torque on the pendulum and an impulsive torque, \( T_i \), that will bring the pendulum to rest:
\[
T = -mg h \theta + T_i,
\]
where
\[
\int_{t=0}^{t=t_e} T_i dt = -I v_0.
\]

Any function for \( T_i \) that satisfies the impulsive requirement will suffice, but for simplicity and realism we will choose \( T_i = \text{Asin}(\pi t/t_e) \). Solution of the integral yields the value of the constant \( \lambda = -I(v_0 \pi /2t_e) \). Substitution of \( T \) into the pendulum equation of motion (A1) gives:
\[
\frac{d^2 \theta}{dt^2} = -\frac{v_0 \pi}{2t_e} \sin \frac{\pi t}{t_e}.
\]

Double integration between \( t = 0 \) and \( t = t_e \) gives:
\[
\theta = \theta_0 + \frac{v_0}{2} \left( t + \frac{t_e \pi}{\pi - \sin \frac{\pi t}{t_e}} \right),
\]
and substitution into the equation for subject torque (A2) gives:
\[
-T = mg h \theta_0 + mg h \frac{v_0}{2} \left[ t + \left( \frac{t_e}{\pi} + \frac{I}{mg h t_e} \right) \sin \frac{\pi t}{t_e} \right],
\]
where \(-T\) is plotted in Fig. 9.

The temporal symmetry of the problem allows the full ‘drop and catch’ to be calculated using the range and the double ‘drop and catch’ (Fig. 9D) to be calculated using \(-t_e \leq t \leq 3t_e\).

REFERENCES


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