MECHANISMS FOR HUMAN BALANCING OF AN INVERTED PENDULUM USING THE ANKLE STRATEGY

by

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Abstract

Maintenance of upright, human balance is neurologically and biomechanically a complex process, though the ankle strategy predominates in quiet standing. This investigation seeks insight into the complex problem by studying a reduced, yet related problem of how the ankle mechanisms are used to balance a human proportioned inverted pendulum. A distinguishing feature of the task is that despite one's best efforts to control this unstable load some irreducible sway always remains.

Contrary to published ideas, modulation of effective ankle stiffness was not the way that sway size was altered. Rather, position was controlled by an intermittent, neurally modulated, ballistic-like pattern of torque whose anticipatory accuracy was improved to reduce sway size.

Using a model, and by direct measurement, I found the intrinsic mechanical ankle stiffness will only partially counter the "gravitational spring". Since this stiffness was substantially constant and cannot be neurally modulated, I attribute it to the foot, tendon and aponeurosis rather than the activated calf muscle fibres. Thus triceps-surae muscles maintain balance via a spring-like element which is itself generally too compliant to provide even minimal stability.

I hypothesise that balance is maintained by anticipatory, ballistic-like, biasing of the series-elastic element resulting from intermittent modulation of the triceps-surae.

DEDICATION

Dedicated to my friend Brian Door,

without whose help a PhD would not have been possible,

who provided the inspiration to research the mechanisms of upright balance.

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PREFACE

OVERVIEW OF THESIS

My investigation into human balance has focussed on the use of the ankle mechanisms in controlling a real, human proportioned, inverted pendulum. The stimulus to this project has come from the work of Fitzpatrick et al (1992). These authors built an inverted pendulum for subjects to balance using their ankle musculature. The inverted pendulum was an equivalent body to the subject's own that was balanced without using vestibular and visual feedback. In this work, these authors presented results showing that the reflex ankle stiffness was very large, (sufficient to stabilise the pendulum) and that this reflex stiffness could be substantially altered by intentional set and by the addition of increased sensory feedback.

The apparatus provided a strikingly novel and interesting approach for studying the fundamental principles by which upright balance is maintained. Consequently, a redesigned and improved version of this apparatus was built at the School of Sport and Exercise Sciences at Birmingham University. I was fortunate enough to inherit this apparatus in full working order with all developmental problems solved (Kelly, 1998).

The interpretation of results offered by Fitzpatrick et al (1992) appeared to be based on a misunderstanding of the nature of balancing an unstable load. If one perturbs a load - control system that is stable then the load moves to a new equilibrium position and the system stiffness can be measured. If one perturbs an unstable load - system the situation is different. In fact, their conclusions (i) that stability is achieved by reflex ankle stiffness (ii) that reflex stiffness is greatly altered by intentional set or sensory feedback and (iii) that reflex stiffness was increased to reduce sway, were all thought to be worth further investigation.

My project began with a reinvestigation of the results and conclusions obtained by Fitzpatrick et al (1992). The question addressed was the simple one of whether the effective ankle stiffness is increased in order to decrease sway. Initially I repeated the experiments in the manner described in the original paper. Eventually it became clear that the role of the perturbations was to destabilise the subjects so that the change in ankle torque with ankle angle could be tracked. I realised the artificial perturbations were unnecessary because there were so many natural, spontaneous losses of balance that I could study. The method, results and conclusions of this investigation were published (Loram *et al.*, 2001) and the paper is reprinted as **Chapter 2** of this thesis.

The results published in **Chapter 2** were a rejection of the idea that effective ankle stiffness is altered to modify sway. Balancing the pendulum myself and studying the data leaves a clear impression that the process of balance is somewhat different from the stiffness control paradigm and also somewhat different from the continuous linear systems approach of modern engineering. The process seems to consist of discrete, trial and error, interactions with the unstable load in which a loss of balance is reacted to and in which anticipation also plays a role. An attempt to analyse the process within these terms was published (Loram & Lakie, 2002b) and the manuscript is reprinted as **Chapter 3** of this thesis.

The investigation of **Chapter 3** contains a calculation of the intrinsic, mechanical ankle stiffness based on a simple model relating ankle torque to ankle angle and soleus EMG. A prediction of the intrinsic stiffness was made on the basis of these calculations. It was important to test this prediction by direct measurement and so a novel experimental technique was developed for making these measurements without interfering unduly with the process of quiet standing or pendulum balancing. The development of this technique is described in **Chapter 4**. The method, results and conclusions of the measurements have been submitted for publication (Loram & Lakie, 2002a) and the article is printed in **Chapter 5** of this thesis.

The final chapter, **Chapter 6**, contains eleven conclusions which can substantiated by data gained in the course of this project. It also contains the outline of a hypothetical, ballistic-bias model of the ankle mechanisms used in upright balance. Substantiation of this model is intended to be the subject of further work.

The first chapter, **Chapter 1,** contains a general introduction to the inverted pendulum model in relation to quiet standing. It also contains a review of research relating to quiet standing and to the role of stiffness in maintaining posture.

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CHAPTER 1.

GENERAL INTRODUCTION AND LITERATURE REVIEW.

In normal circumstances people rarely stand still, unsupported, for extended periods of time. Common observation shows that standing people tend to make use of walls, furniture or other supporting surfaces and when these are not available people tend to shift position, take isolated steps to change the position of their feet and walk around a bit. Standing still, unaided, is a skilled activity that many cannot sustain for more than a few minutes before becoming fatigued and uncomfortable (Duarte & Zatsiorsky, 2001). Although prolonged, unconstrained standing has been studied recently (Duarte *et al.*, 2000; Duarte & Zatsiorsky, 1999, 2001) most laboratory studies of standing attempt to simplify and standardise the process so that measurements can be made more easily and interpretative models can be applied more successfully. Quiet standing as it is typically studied requires the subject to maintain their feet in a set position, often side by side, either together or a short distance apart, and this position is maintained for at most a few minutes while data are recorded.

During quiet standing the body is in continuous movement at all its joints (Day *et al.*, 1993; Hodges *et al.*, 2002) and the body centre of mass (CoM) displays a random looking motion with a mean position typically one metre above the ground and a few centimetres in front of the ankle joints (Basmajian & De Luca, 1985). More careful analysis of the CoM motion along each of the antero-posterior and medio-lateral axes shows that it repeatedly sways unstably, unpredictably and unidirectionally for a mean duration of one second. The root mean square distances of these sways are 3 and 2 mm respectively (Collins & Deluca, 1993). The succession of sways accumulates to a motion that is constrained within a horizontal planar area of a few square centimetres (Collins & De Luca, 1995; Collins & Deluca, 1993). Forward collapse of the CoM is prevented by ankle torque generated by the sustained contraction of soleus and the intermittent or sustained contraction of gastrocnemius (Basmajian & De Luca, 1985).

When standing quietly, the predominant movement of the body parts is synchronous sway about the ankle joints particularly in the sagittal plane (Gatev *et al.*, 1999). This mode of balance is described as the ankle strategy (Nashner & Mccollum, 1985). It is only when balance is more threatened, such as when a subject is subjected to large perturbations that control of the centre of mass by movement at the hips becomes dominant and this latter mode is described as the hip strategy (Nashner & Mccollum, 1985). The dominance of the ankle strategy in quiet standing has led many researchers to use the inverted pendulum model to represent the body centre of mass supported above the ankles. While this model may be simple, even simplistic, it does characterise most of the movement of the centre of mass in quiet standing, particularly in the sagittal plane (Winter *et al.*, 1998).

The inverted pendulum model of standing.

The inverted pendulum model represents the body above the ankles as a single lumped mass and the two feet as a single foot as shown in Figure 1 - 'Inverted pendulum model'. The body is thus free to rotate in an antero-posterior sense and in a medio-lateral sense about a single ankle joint. It is assumed that the foot is rigid and fixed in position and that the single axis of rotation does not move. It is also assumed that the distance between the ankle and the centre of mass of the lumped body does not change. These assumptions require that compliance of the foot and movement at the knee, hip, and vertebral joints are inconsequential so that there is no movement of the single axis of rotation and so that the distance from the ankle joint to the body centre of mass does not change. The model also assumes that the position of the CoM is controlled through torque exerted only at the ankle joint, and not from any other joint.

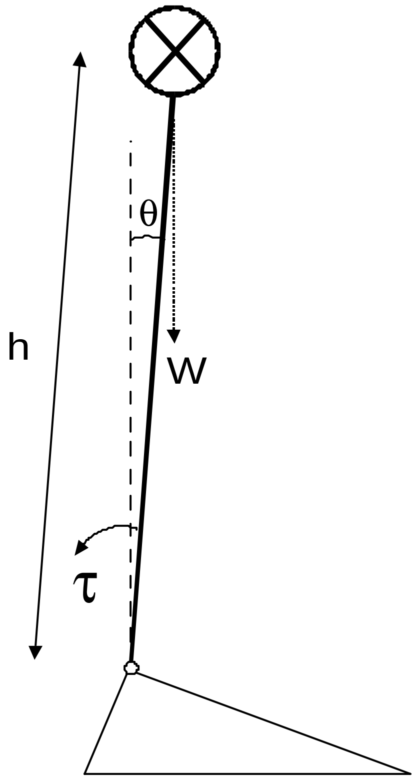

Figure 1. Inverted pendulum model.
The mass of the body above the ankle joints is represented by a single centre of mass **Ā**, which is at a distance ***h*** from the ankle joint of a single foot. The centre of mass is shown at an angle ***q*** to the vertical and a torque ***t*** is applied to the centre of mass by the ankle musculature. Gravity exerts a weight ***W*** acting vertically downwards through the centre of mass.

Let us consider the lumped body above the ankles as having weight W, moment of inertia about the ankle I, and distance h from the ankle. This inverted pendulum is at angle q to the vertical. The gravitational torque on the pendulum is $Wh \sin(q)$ and the pendulum is subjected to an ankle torque t which acts in the opposite sense to gravity. In the ideal case of no frictional or drag forces the equation of motion for that inverted pendulum in a single plane is

$$I\, d^2q/dq^2 = Wh \sin(q) - t \tag{1}$$

which for small angles close to the vertical simplifies to

$$I\, d^2q/dq^2 = Whq - t \tag{2}$$

In the absence of an ankle torque to control it ($\tau=0$), the inverted pendulum is highly unstable. Equation (2) shows that the acceleration of the pendulum is proportional to its angle from the vertical and this leads to an angle that increases somewhat exponentially through time if the pendulum is allowed to fall freely. If the pendulum is released at rest at angle q_0 from the vertical, the analytical solution to the linear equation (2) is $q = q_0 \cosh (\text{Ö}(Wh/I)\, t)$. A typical human proportioned pendulum, falling freely from rest at an angle of 0.001 degrees to beyond the length of the foot, is shown in Figure 2 - 'Free fall of inverted pendulum'. This graph illustrates the fact that for that for a very small torque imbalance the pendulum moves relatively little for several seconds. This is a consequence of the large moment of inertia of the pendulum about the ankle joint. For a large torque imbalance the pendulum is rapidly destabilised despite its large inertia. This tells us that the ankle strategy is viable so long as the torque imbalance is kept very low. If the pendulum becomes more unbalanced or likely to topple out of control then the CoM has to be moved more rapidly. This is more easily achieved by a hip strategy since the moment inertia of the CoM about the hip is very low and large torques can be generated by the powerful hip muscles, (although verticality of the trunk is lost).

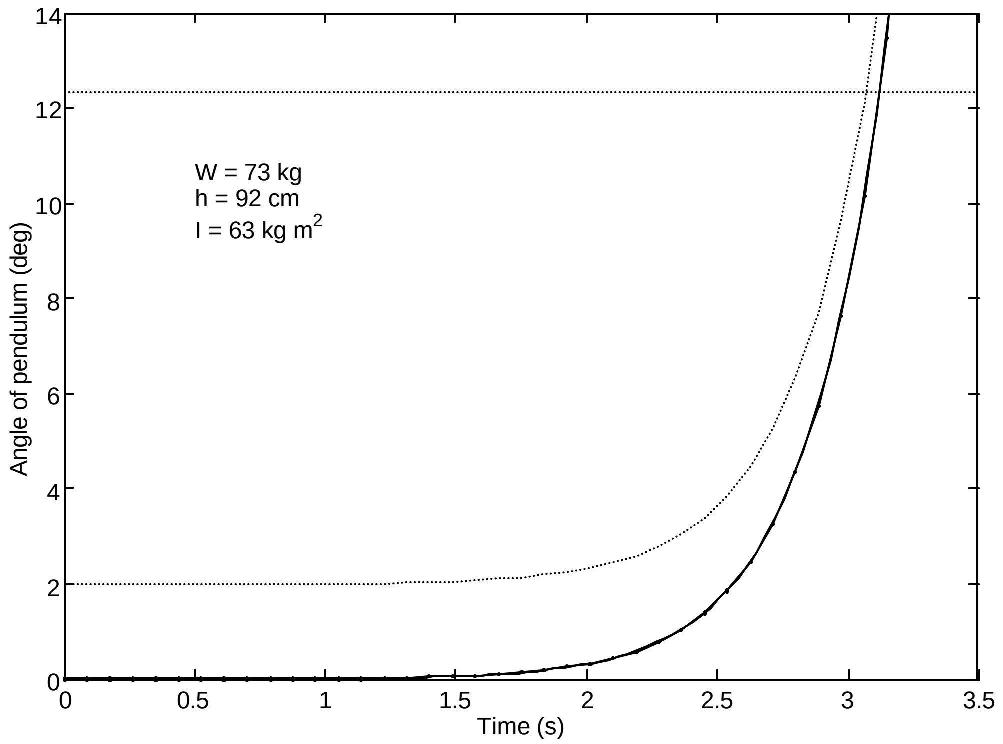

Figure 2. Free fall of inverted pendulum.
The solid line and dashed curves show trajectories of a pendulum released at rest from 0.001 and 2.001 degrees respectively. The weight, W, height, h and moment of inertia, I, of the pendulum are shown. For the solid curves the ankle torque is zero Nm and for the dashed curve the ankle torque is maintained at a constant value that would balance the pendulum at 2 degrees. The straight dashed line represents the angle at which the centre of mass would pass beyond a foot of length 0.2 m in front of the ankle. The dots and dotted line represent numerical solutions to the non-linear equation of the pendulum and the solid line represents the analytical solution to the linear equation of the pendulum.

The ankle torque required to balance the pendulum is proportional to the angle from the vertical and is given by $t = Wh\ sin(q)$ » Whq. If the ankle torque is maintained at a constant level judged appropriate for the angle of the pendulum, then any torque imbalance will be subjected to positive feedback and will cause the pendulum to accelerate exponentially away from its original position. The dashed curve in figure 2 'Free fall of pendulum' shows the pendulum falling from rest at 2.001 degrees when the torque is maintained at a constant level appropriate for an angle of 2 degrees. If balance and positional control of the pendulum is to be maintained then the ankle torque has to be modulated as the pendulum changes angle. For stability, that is to prevent a torque imbalance from being subjected to positive feedback, the ankle torque must change with angle at least as rapidly as the gravitational torque does. What ever mechanism or process is used to control the inverted pendulum, be it passive, reflex, predictive or an engineering control system, the minimum criterion for stability is

$$(Dt/Dq)_{average}\ ^{3}\ Wh.$$

If on average the rate of change of ankle torque with CoM angle is greater than that of gravity, then the pendulum will oscillate in some fashion. In standing this oscillation is irregular and unpredictable. However, along either the antero-posterior axis or the medio-lateral axis the mean duration of sway in one direction is 1 second (Collins & Deluca, 1993) hence the mean time of a sway forwards and sway backwards is 2 seconds giving an average frequency of 0.5 Hz.

To sustain a regular oscillation at a single frequency, f, the ankle torque must change uniformly with CoM angle according to the relationship $dt/dq = Wh + (2pf)^2 I$. This result can be obtained by Laplace transformation of equation (2) and substitution of the steady state frequency (iw) for the Laplace variable (s). Although oscillations in standing are not regular, this relation tells us to expect a mean ankle torque v CoM angle gradient of 200% of the gravitational toppling torque per unit angle (*Wh*) to produce a mean frequency of 0.5 Hz (Morasso *et al.*, 1999). Figure 3 - 'Torque gradient required for oscillation' shows the ankle torque gradient required for 0.5 Hz oscillation is typically 22 Nm deg^{-1} for a human proportioned inverted pendulum, and 11 Nm deg^{-1} (100%) is the minimum required for marginal stability.

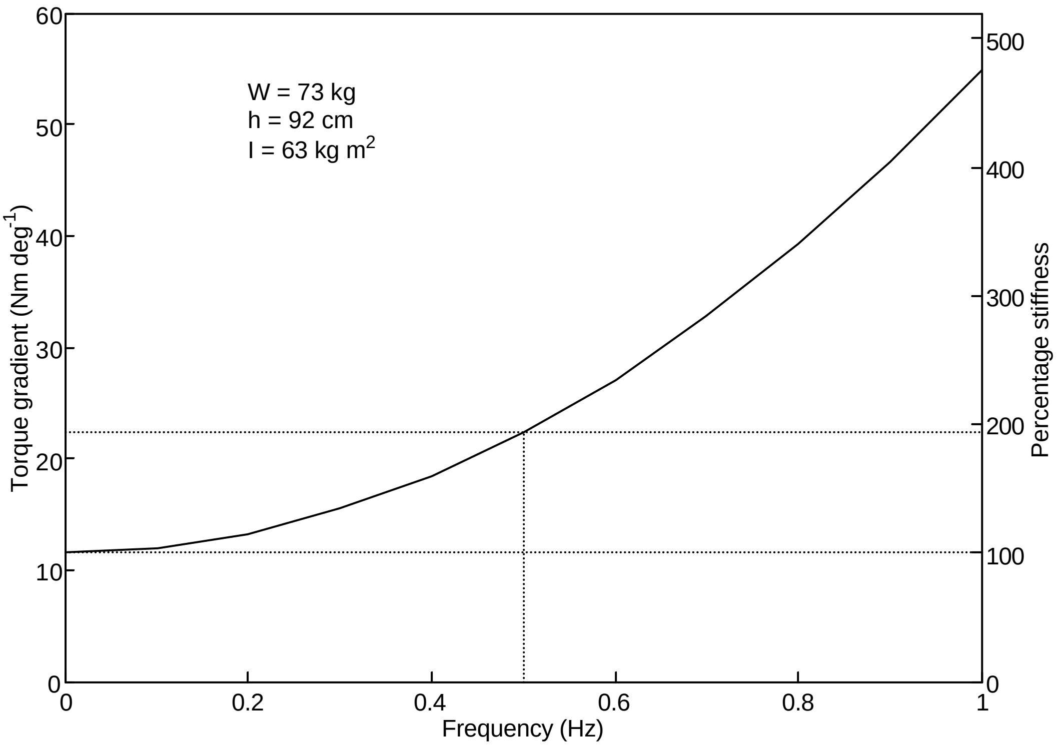

Figure 3. Torque gradient required for oscillation
The solid curve shows the rate of change of torque with angle (stiffness) that is required to maintain oscillation at a regular frequency. The right axis shows the stiffness relative to the gravitational torque per unit angle of the inverted pendulum. The weight, W, height, h and moment of inertia of the pendulum are shown. For quiet standing, sway is irregular, but a typical mean rate of oscillation (0.5 Hz) is indicated.

Review of previous enquiries into standing that use the inverted pendulum model.

An early investigation of quiet standing identified inverted pendulum oscillations of the CoM and the ankle torque at around 0.3 Hz (Smith, 1957). Today, this figure would be regarded as being a little low (Morasso *et al.*, 1999; Morasso & Schieppati, 1999). The active muscular ankle torque that prevented forward collapse of the CoM was measured and found to vary between 18 and 50 Nm over a long period, which would correspond to a large variation in angle of several degrees. The passive ankle torque from both ankles combined was measured to be about 7 Nm and this shows that the passive elements of the ankle joint do have a role in contributing to stability.

Gurfinkel et al have applied the inverted pendulum model to quiet standing and derived opposite conclusions in two of their earlier papers. In 1972 they reasoned that the intrinsic stiffness of the activated plantar flexor muscles cannot stabilise the inverted pendulum because of the asymmetrical shape of the muscle length-tension relationship combined with the changing moment arm of the muscle. They concluded that active modulation of the muscle activity is required, that some feedback system is required and that velocity feedback is necessary to overcome the time delays present in the neuro-muscular system (Gurfinkel & Osevets, 1972). Later in 1974, they used the inverted pendulum model to calculate the minimum ankle stiffness required for stability. For their five subjects they calculated that 12 - 15 Nm deg^{-1} would be required for minimal stability. By comparison with my own measurements (Loram & Lakie, 2002) they overestimated this value on two counts. First they overestimated the value of h (the distance between the CoM and the ankle) by 5%. Second they over estimated the mass of the inverted pendulum by 3% by not subtracting the mass of the feet. They also measured the ankle stiffness by rotating the force platform on which their subjects were standing. These rotations were applied in the sagittal plane at 0.6 deg s^{-1} through 0.1 and 0.2 degrees and are comparable ankle speeds and displacements to those encountered in normal standing. The combined stiffness results for both legs are 13 - 19 Nm/deg which is higher than our own measurements (Loram & Lakie, 2002). Their values were admittedly noisy and the sample was small. Most importantly, they did not separate the intrinsic mechanical changes in torque caused by the perturbation from the neurally modulated changes in torque associated with the change in position of the body CoM and with the reaction to the perturbation. From visual inspection of raw EMG they concluded that there were no stretch reflexes induced even though the movements were greater than the threshold of detectability. In contradiction to their 1972 paper, they concluded that the intrinsic ankle stiffness is great enough to compensate for the small sways of quiet standing (Gurfinkel *et al.*, 1974). In reality, their mean relative stiffness of 112% would only provide minimal stability and would not account for the now known frequency of oscillation of the body CoM. Active modulation of ankle torque would still be required to account for these oscillations.

Given the multiplicity of joints in the human structure and the number of muscles available to control movement at all the joints, there is a very large number of possible ways in which movements can be made to and from upright balance. Using only two organisational hypotheses, it can be predicted that the posture control system produces discrete combinations of a limited repertoire of distinct movement strategies including the ankle strategy, the hip strategy and the suspensory strategy (Nashner & Mccollum, 1985). The suspsensory strategy involves vertical motion of the body through buckling of the legs at the ankle, knee and hip joint and this strategy is rarely observed in standing sway or perturbations to standing sway since the knee tends to be maintained in the extended position (Nashner & Mccollum, 1985). The ankle strategy involves the subject swaying as a flexible inverted pendulum about the ankles with little hip or knee motion. This strategy is used predominantly during small to medium translational or rotational perturbations of the platform that the subject is standing on. The hip strategy involves trunk rotations about the hip and is evoked when subjects are unable to generated sufficient ankle torque such as when standing on a narrow beam which is translated.

A fundamental problem in scientific observation is how to measure a system without altering the characteristics of the system you are studying. Experimenters frequently apply disturbances to standing people and observe the reaction, although that response may not be characteristic of quiet standing, especially if the disturbance is large. One approach is to analyse spontaneous sway and attempt to identify the closed loop feedback characteristics but without taking external disturbances into consideration (Ishida & Miyazaki, 1987). Using the inverted pendulum model a light emitting diode at the hip gave sagittal sway angle as a system input. A force platform was used to give ankle moment as an output. For six subjects, a second or third order model gave a transfer function with the best fit. The transfer function showed derivative characteristics which were seen as an increase of gain with frequency (up to 0.5 Hz) and an increase in phase lead with frequency (up to 0.5 Hz). This conclusion substantiated Gurfinkel and Osevets' conclusion in 1972 that velocity feedback (damping) was necessary given the time delays in the feedback loop. Ishida and Miyazaki also concluded that the visual feedback path has a rather long latency and that the visual system guarantees necessary gain and stability at frequencies lower than 0.2 Hz. In other words, without vision, people would drift more and lose their more absolute sense of position.

The validity of a third order feedback model for ankle torque was tested the following year using external perturbations (Johansson *et al.*, 1988). Using an inverted pendulum model of sagittal sway, corrective ankle torque was modelled as a PID controller. This is a third order model with stiffness, damping and swiftness parameters. Stiffness and damping have clear mechanical analogies in a hydraulically damped spring. Swiftness represents the time taken to reset a reference position that has drifted. External perturbations were provided by vibration of the calf muscles. Maximum likelihood estimation of parameters in ARMAX models was used to show that the third order model with the swiftness parameter is statistically valid.

Given the use of the inverted pendulum model in interpreting data from standing, it was natural that eventually someone would build a real human proportioned inverted pendulum for subjects to balance using their ankle musculature. The credit for this novel approach goes to Fitzpatrick et al who built the first working apparatus of this type (Fitzpatrick *et al.*, 1992b). These authors used the apparatus in a whole series of experiments into standing lasting several years. The underlying theme of their research was the stabilising role of muscle reflexes in quiet standing. The purpose of the inverted pendulum apparatus was to provide reduced sensory conditions under which reflexes could be studied. Since subjects were strapped to a vertical support, balancing the real inverted pendulum was considered as balancing a load equivalent to their own body in a situation where neither visual nor vestibular information could assist. Fitzpatrick et al used the apparatus for studying reflexes under reduced sensory conditions but did not use it as an end in itself for studying the first principles by which an inverted pendulum load is controlled by the ankle musculature.

In the first paper these authors measured the reflex ankle stiffness during quiet standing and under a variety of reduced sensory conditions. First they showed that the inverted pendulum model of stance successfully predicts the low frequency torque-angle relationship at the ankle when standing and that this relationship determines the minimal ankle stiffness required to stand (Fitzpatrick *et al.*, 1992b). Then they tracked the changes in ankle torque and ankle angle during a 1.5 second period while their subjects were subjected to slow, imperceptibly small, ramp perturbations at the waist. They claimed that the change in torque per change in ankle angle measured the reflex ankle stiffness. The responses were interpreted as being reflex because the perturbations were not detected. These authors concluded that the reflex stiffness was greater than the minimum required for stability and in particular the local sensory reflexes from the legs alone are sufficient to stand. They also concluded that subjects increased their reflex ankle stiffness when they were "standing still" compared with when they were "standing easy".

These authors appear to have assumed that when a perturbation is applied the load will move to a new equilibrium position that is determined by the system stiffness. This is true for a stable load - control system but is not the case here. In this case, the perturbation destabilises the load and I argue there is no system stiffness that determines a new equilibrium position. In Figure 4 - 'Fitzpatrick experiment', position 1 shows the ankle torque and ankle angle of a subject prior to perturbation. The subject is producing an ankle torque appropriate for the pendulum load at that angle. Positions 2 and 3 show the ankle torque and position 1.5

seconds after the perturbation while the subject was instructed to stand still (2) or stand easy (3) respectively. At these positions the subject is producing a torque that balances the pendulum load and the additional load produced by the pull of the perturbing spring at the waist. Subtraction of the perturbation torque from the ankle torque leads us to positions 4 and 5 on the figure. This shows that in response to the perturbation the subject has both accommodated the perturbation and moved to a greater or lesser extent. While they were pulled forward, the subjects were balancing a slowly increasing perturbation load as well as an increasing gravitational load. What the results tell is that subjects can unconsciously accommodate an increasing load at the ankle over a period of 1.5 seconds while moving to a greater or lesser extent. Had a slightly longer period been observed for the "stand still" condition the pendulum may have returned close to the original position or even moved to a smaller angle and the measured values of stiffness would have been completely different.

Moreover, there was little justification for claiming this result was achieved by sensory reflexes. To many readers, reflex stiffness refers to such things as the stretch reflex, or the motor servo (Houk, 1979; Houk & Rymer, 1981) which is an autogenetic, segmental level reflex mechanism. Although one might stretch the understanding to include longer-loop reflexes, the evidence for supraspinal servo loops is thought to be weak (Houk, 1979). Over the 1.5 second period it was likely that the changes in ankle torque resulted from an unconscious but anticipatory and predictive process. While standing quietly or balancing an inverted pendulum, subjects seem able to vary the ankle torque in a manner that is quite unrelated to the ankle angle producing curves on a torque v angle plot that sometimes have infinite and sometimes have negative stiffness. It was implausible that the changes in torque result from a reflex negative feedback mechanism.

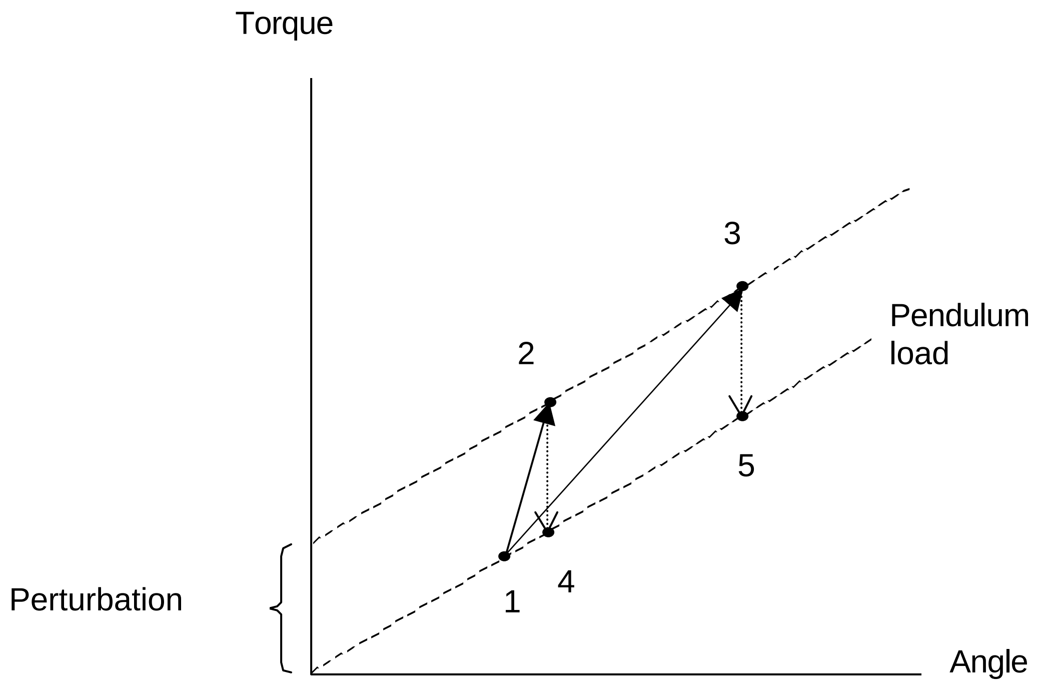

Figure 4. Fitzpatrick experiment.
This figure shows my interpretation of the results from Fitzpatrick et al (1992b). The lower, inclined, straight dashed line shows the gravitational load of the pendulum for every angle. The higher, parallel dashed line shows the gravitational load plus a constant perturbation load which has been added after 1.5 seconds. Position 1 shows the torque and angle of the subject before the perturbation. Positions 2 and 3 show the torque and angle of the subject 1.5 seconds after the ramped increase in perturbation for "stand still" and "stand easy" instructions respectively. Positions 4 and 5 show the respective torque and angle of the subject after subtraction of the torque which is balancing the perturbation.

In a second experiment published that year continuous random perturbations were applied at the waist (Fitzpatrick *et al.*, 1992a). Using spectral analysis, transfer functions were calculated relating the soleus EMG to the angular perturbations (reflex response) and also relating the ankle torque to ankle angle (ankle stiffness). Like Ishida and Miyazaki they found the reflex gain and phase advance both increased with frequency up to 5 Hz. They found that the response and transmission characteristics of the lower limb proprioceptive reflex in freely standing subjects were similar to those in subjects balancing a inverted pendulum load at the ankle, a situation in which vestibular and visual inputs could not contribute. The conclusion was that reflex feedback related to ankle movement contributes significantly to maintaining stance, and that much of the reflex response originates from lower limb mechano-receptors stimulated by ankle rotation. Their results also showed that reflex gain could be increased when subjects were instructed to stand still and this was interpreted as meaning that their was an increased reflex ankle stiffness when standing still. These conclusions should be qualified by the same comments as above. Reflex in this context only means that the responses were unconscious. They could have included anything from short latency stretch reflexes to trans-cortical feedforward predictive mechanisms.

Not all investigators ascribe fully to the inverted pendulum model for quiet standing (Day *et al.*, 1993). The three dimensional position of eight infra-red emitting markers were tracked on thirty five subjects who stood as still as possible on a force platform. The markers were placed at shoulder, hip, knee and ankle level allowing computation of angular movement at the ankle, knee and hip joints in both the sagittal and frontal planes. For normal stance the medial malleoli were separated by 16 cm. These authors found that angular movement at the knee is approximately the same as at the ankle whereas angular movement at the hips is twice that of the ankle or the knee. Their data also shows (c.f. their Figure 6) that the positions of the markers move predominantly in a synchronous manner with markers higher up moving faster and farther than markers lower down. This is true in both the frontal and sagittal planes. They concluded that the synchronous movements of knee, hip and shoulder are consistent with the inverted pendulum model but found that the inverted pendulum model is incomplete due to the greater angular movement at hip joint. As the authors point out, angular movements about the ankle produce greater displacement of the body mass than angular movements about the hip. If the angle at the hip is up to twice the angle at the ankle, then the change in distance *h* between the ankle and the body CoM is totally negligible given typical body dimensions. This can be seen in Figure 5 - 'Effect of hip strategy on distance h'. The main consequence of angular movement at the hip and knee is that the ankle angle is not a totally reliable predictor for the angle of the CoM to the vertical. This can be seen in Figure 6 - 'Effect of hip strategy on angle of CoM'. Thus proprioceptive information from the ankle joint has to be combined to a lesser extent with proprioceptive information from the knee and hip joints if the CNS is to know where the CoM is. Whether or not the model of an inverted pendulum controlled by an ankle strategy is valid depends on whether the movement of the body CoM correlates with the ankle torque. The validity of the model does not depend on whether the movement of the body CoM corresponds with the ankle angle.

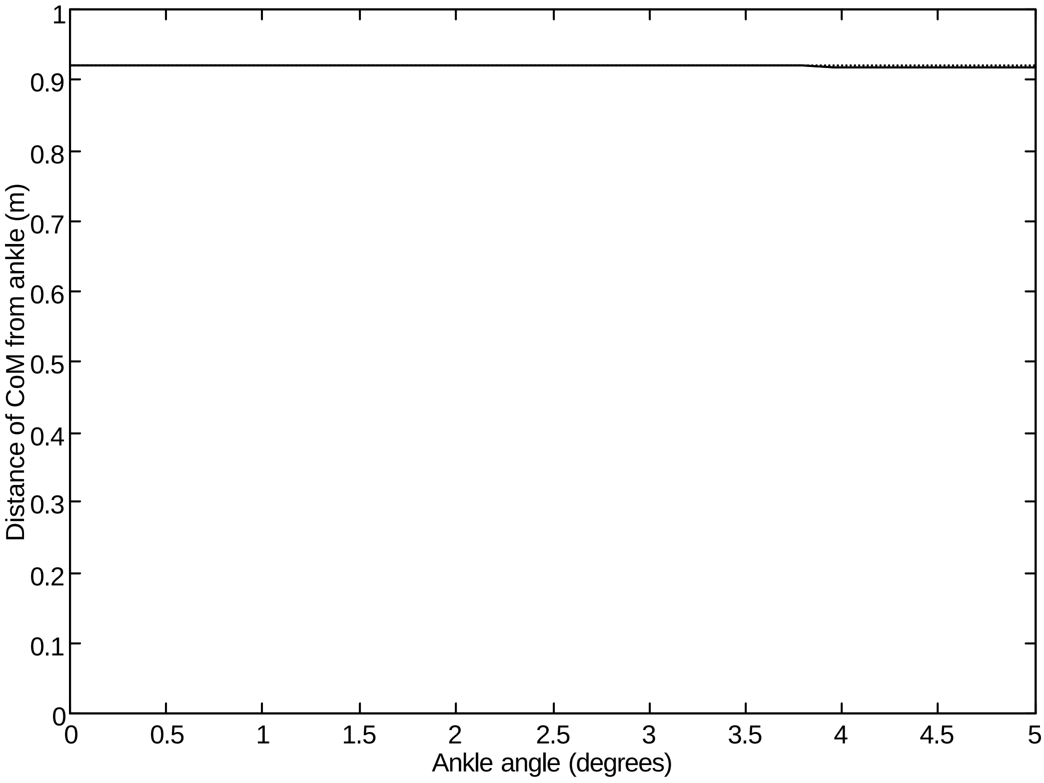

Figure 5. Effect of hip strategy on distance h.
There are two lines almost indistinguishable from each other. The dashed line shows the distance of the centre of mass from the ankle joint (0.92 m) for a variety of ankle angles assuming an inverted pendulum model. The solid line shows the distance of the centre of mass from the ankle joint assuming that the angular change at the hip is twice that at the ankle.

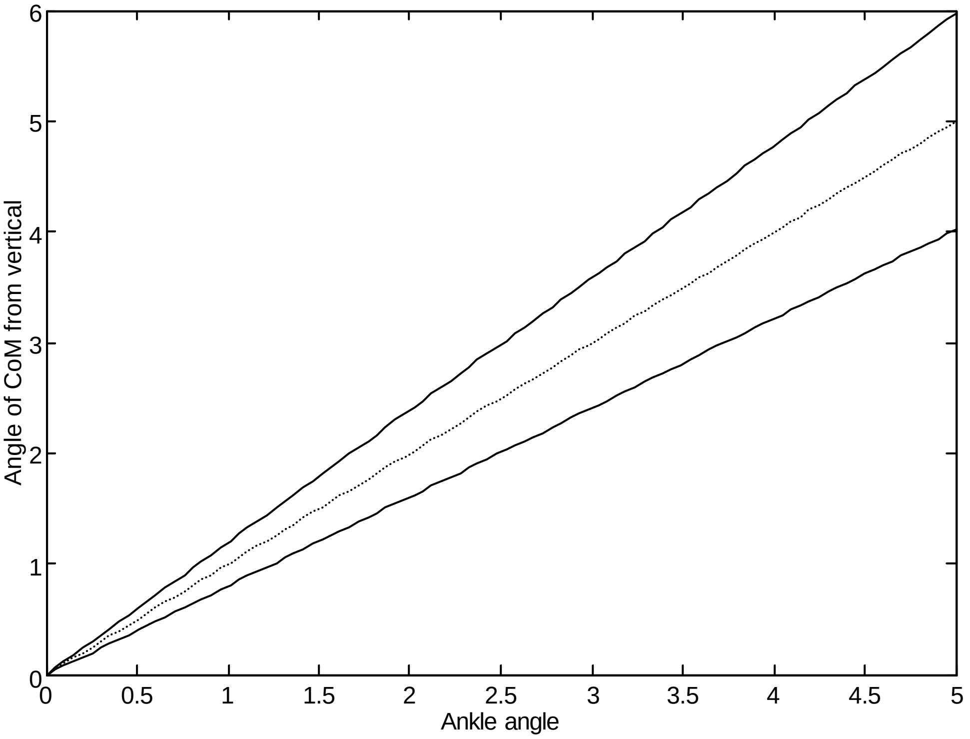

Figure 6. Effect of hip strategy on angle of CoM.
The dashed line shows that for the inverted pendulum model the angle of the centre of mass (CoM) from the vertical is the same as the ankle angle. The solid curves show the angle of the CoM from the vertical when the hip angle is ± twice the ankle angle. This calculation assumes a hip to CoM distance of 0.09 m and an ankle to hip distance of 0.83 m.

The following year, Fitzpatrick et al published the results of three interesting experiments. First they tested subjects for their minimum visual, vestibular and ankle proprioceptive thresholds (Fitzpatrick & Mccloskey, 1994). Using a variety of reduced sensory arrangements thresholds were determined as the point at which a stimulus could be reported. To limit the available sensory input to proprioception from the legs, subjects were held stationary and using their ankle musculature they balanced an inverted pendulum load that was equivalent to their own body. In this situation, perturbations were applied to the 'equivalent body' and these could only be perceived from the resulting ankle movements. Results showed that during standing sway, vestibular thresholds were ten times the visual and proprioceptive thresholds. Minimum perception thresholds were 0.17 degs at 0.06degs/s and less at higher velocities which is interesting because 0.17 degrees is approximately the average size of a sway in quiet standing. The legs provided the most sensitive perception during normal sway. Visual sensitivity was greater at higher velocities and when visual objects were closer. The underlying conclusion was that subjects use the most sensitive information available at the time and other sources of information become redundant.

To identify the vestibular contribution to human standing, responses in leg muscles were evoked by galvanic vestibular stimulation (Fitzpatrick *et al.*, 1994a). Step impulses of current were applied between the mastoid processes of normal subjects and the effects on the soleus and tibialis anterior electromyograms (EMGs), ankle torque, and body sway were studied. Short latency, reciprocal soleus and tibialis responses were observed after 56ms. Larger medium latency opposite response were observed after 105ms giving a biphasic pattern overall. These short- and middle-latency responses were modified during different postural tasks according to the subject's dependence on vestibular reflexes. When the support platform was unstable, the EMG responses to galvanic stimulation were larger than when the subject was standing quietly. There were no vestibular-evoked responses when subjects stood upright with the trunk supported, using the ankles to balance the body-like inverted pendulum load. In conclusion, vestibular responses were greatest when vestibular information was needed such as in unstable standing, but vestibular responses were small or absent when they were not needed such as in quiet standing and balancing the inverted pendulum respectively.

Finally, these authors tested subject's ability to stand using only proprioceptive information from the lower leg. Also, balancing the inverted pendulum load was compared with splinted standing and ordinary standing (Fitzpatrick *et al.*, 1994). Subjects were tested in a variety of reduced sensory arrangements including ankle cuffs to eliminate sensory input from the feet and ankles. The modulation of ankle torque required to support the inverted pendulum in an upright position was similar to that required to support the subject's own body when standing. Compared with quiet standing sway increased by 17% when proprioception from the feet and ankles was excluded. Sway also increased by 17% when eyes were shut and when vestibular information was excluded by balancing the equivalent load which means that proprioceptive signals from leg muscles, the only remaining source of information, is sufficient for upright balance. It was also found that visual input was necessary for maximum stability and that information from the feet and ankles has smaller but significant effect on stability. During standing, vestibular inputs were not responsible for modulating activity in leg muscles to assist stability. When standing splinted at the knee and hip, rms sway was 52% greater than that in quiet standing. Sway of the equivalent body, the inverted pendulum load was 38% greater than sway in quiet standing. This shows that segmentation of the body significantly improves stability in quiet standing and demonstrates that the inverted pendulum model of quiet standing is indeed incomplete.

The combination of hip and ankle strategy in quiet standing was studied in an interesting modelling investigation (Kuo, 1995). An optimal control model was constructed with four segments and three joints at the hip, knee and ankle although the model was simplified to the situation where the knee is fixed. The model optimised cost functions to minimise the neural effort (muscle activation), to control the centre of mass, to provide a stable platform for the head and to maintain vertical, upright stance. Results showed that the choices available to the CNS are limited by the decision to stand upright and by biomechanical considerations such as the ease of achieving certain combinations of joint angular accelerations. When attempting to preserve upright balance with an economy of movement the model chooses slow movements mostly of the ankle strategy for small disturbances and switches to faster responses using the hip strategy for larger disturbances. This is similar to real human standing where the ankle strategy is dominant for slow disturbances such as in quiet standing. However, unlike a computer model it cannot be assumed that human subjects will necessarily stand in an optimal manner.

In 1996 a reinvestigation into the ability of reflexes (unconscious reactions) to stabilise human standing was published by Fitzpatrick et al. Random perturbations given at the waist enabled calculation of a transfer function relating soleus EMG to sway. Random galvanic vestibular stimulation enabled calculation of a transfer function relating ankle torque to soleus EMG. The former transfer function interpreted as reflex action, showed a phase advance that increased with frequency and exceeded 180 degrees at 5 Hz. This phase advance partially compensated for the phase lag of the muscle and load which increased with frequency and reached 360 degrees at 5 Hz.. Combination of the reflex gain and the muscle and load gain enabled these authors to calculate the opened loop gain of muscle, load and reflexes. The result was that the loop gain was approximately unity. This gain is too low for a position negative feedback system based on reflexes. The results suggest that a feedforward process is in operation since feedforward models require a gain of about unity and an inverse controller which compensates for the phase lags in the loop.

A continual exponent of the model of an inverted pendulum controlled by an ankle strategy has recently tested the model by direct measurement (Winter *et al.*, 1998). The assumptions of the model, given above, require that the ankle torque is closely related to the position of the body CoM. The equation to be tested was $(p_x - x) = (-I/Wh)\ d_2xdt^2$. In this equation x is the forward horizontal projection of the centre of mass from the ankle and p_x is the forward horizontal projection of the centre of pressure (CoP). The CoP is the point of application of the upward reaction force of the ground acting on the foot. This reaction force is caused by the plantar-flexing ankle torque which pulls the foot onto the ground. To a close approximation, the CoP is proportional to the ankle torque. W, I and h are respectively the weight of the body, the moment of inertia of the body about the ankles and the distance between the centre of mass and the ankles. The equation shows that the horizontal acceleration of the CoM is proportional to the difference between the horizontal projections of the CoP and CoM. The horizontal projection of the CoP was measured by standing subjects on two force plates. The position of the CoM was measured by tracking the three dimensional position of twenty one infra-red emitting diodes strategically placed on the subject. The CoM position was calculated from a weighted averaged of fourteen segments.

The noise level in the calculated value of $(p_x - x)$ was 0.2 mm. Ten subjects stood with their feet side by side at 100%, 50% and 150% their hip distance apart. In the sagittal plane the mean correlation coefficient between $(p_x - x)$ and the acceleration was 0.9. (See their figure 5). From the coefficient of determination, $R^2 = 0.81$, this means that 81% of the variance in the CoM acceleration fits or can be associated with the inverted pendulum model (Hampton, 1994). The authors did not corroborate the model further by publishing the constant of proportionality and showing that it was consistent with their prediction of I/Wh. In the frontal plane the mean correlation coefficient and coefficient of determination were r = 0.79 and $R^2 = 0.62$ respectively. Therefore in the frontal plane only 62 % of the variance in CoM acceleration fits the inverted pendulum model. In both the sagittal and the frontal plane the inverted pendulum model is an incomplete explanation of quiet standing. Nonetheless, in the sagittal plane the success of the model is impressive given the simplifying assumptions that it requires. In the sagittal plane at least, the implication is that there are no other major forces acting on the CoM other than gravity and the ankle torque. Thus the main contribution of this paper was to quantify exactly how well the inverted pendulum model and ankle strategy applies to quiet standing both in the sagittal and the coronal planes.

In this paper the authors went further. From the frequency spectrum of the acceleration signal they calculated the stiffness and damping of the system that was responsible for the oscillations. Then without adequate justification they ascribed this stiffness to the intrinsic mechanical stiffness generated by contraction of the ankle musculature. They claimed that no neural reaction or feedback control was needed and that the CNS merely set a level of stiffness and allowed the body to oscillate in a severely under-damped, near resonant manner. They claimed that the size of the oscillation was regulated by the level of stiffness set by the CNS. Part of their reasoning was that on average the CoP was in phase with the CoM and therefore showed no evidence of neural reaction. They admitted that energy was being fed into the system to keep the oscillations alive but gave no explanation of the source of sway. A more obvious explanation of the variation in ankle torque that controls the acceleration of the CoM is that it is at least partly caused by active neural modulation. The stiffness control theory of these authors will not be believed unless they can show that the variation in ankle torque derives almost wholly from mechanical stretching of the elastic elements and not from neural modulation.

A rapid repudiation of Winter et al's paper was published the following year (Morasso & Schieppati, 1999). Morasso and Schieppati rightly pointed out that through prediction, the CNS is capable of compensating for sensory time delays so that the CoP is on average in phase with the CoM. Next they used the assumptions of the inverted pendulum model to derive the same proportional relationship between horizontal acceleration of the CoM and the difference between the horizontal projections of the CoP and CoM. From this relationship they argued that is was unnecessary to measure the position of the centre of mass using a multiple camera system because the centre of mass position could be calculated using the equation, and using CoP and CoM acceleration data obtained from a force platform. Thus they missed the point that Winter et al had measured all terms in this equation and had demonstrated how well the inverted pendulum model applies to quiet standing. Finally they reviewed literature giving values of ankle stiffness and concluded that the intrinsic mechanical stiffness is substantially insufficient to stabilise the inverted pendulum and no where near able to account for the observed mean frequency of oscillation. Unfortunately their arguments were weakened by the fact that they presented combined stiffness values based on values for one leg only and compared this with the gravitational torque per unit angle of a person more massive and taller than average. They argued for active mechanisms of stabilisation and pointed out the potentially crucial role of the cells in the skin of the foot capable of detecting shear forces and hence the acceleration of the body centre of mass. In the same year the idea of predictive stabilisation using internal models was argued further and the idea that the cerebellum acts as a Smith's predictor was advocated (Morasso *et al.*, 1999).

Previous experiments (Day *et al.*, 1993; Nashner & Mccollum, 1985) and modelling (Kuo, 1995) have suggested that people stand with a combination of ankle and hip strategy. Using the latest equipment, a recent study investigated the strategy used during quiet standing and aimed in particular to confirm whether a hip strategy is used, or an ankle strategy, or a combination of the two (Gatev *et al.*, 1999). A VICON system of five CCD cameras was used to track the three dimensional movement of reflective markers placed to enable calculation of ankle, knee and hip angles as well as movements of the body through space. The positional noise level of the markers was 0.2 mm. A force platform was used to record the CoP. It is a pity that only seven subjects were studied while standing quietly. Moreover the stance used was a little unstable with only one to two inches between the heels and ten inches between the toes. In the sagittal plane results showed that the CoP movements correlated very highly ($r > 0.95$) with zero phase difference with the body CoM. This demonstrates the use of the ankle strategy because with the hip strategy alone there would be no correlation between the CoP and the body CoM. For example, if a subject was using a pure hip strategy to balance while standing on a knife edge support, there would be no ankle torque and the CoP would always be under the foot at the position of the knife edge. The head position also correlated very highly (r > 0.95) with the body CoM and the knee position correlated highly with the shoulder position (r = 0.8). Results showed greater angular movement at the hip rather than the ankle, however, only ankle motion, not hip or knee, correlated significantly (r = -0.5) with the CoP movements. The high positive, zero phased correlations between positions of the CoP, CoM, head, shoulder and knee suggests almost synchronous sway of the body parts in the sagittal plane. In the sagittal plane there was no significant correlation between hip position and hip angle which would indicate hip strategy. Thus ankle mechanisms were found to dominate in the sagittal plane. However, the low correlation between ankle angle and CoP indicates that ankle angle is not a reliable predictor for body CoM position and is consistent with angular motion occurring at other joints. In the frontal plane, results showed that that ankle (inverter/evertor) and hip (abductor/adductor) mechanisms control the mediolateral sway).

Recently, subspace methods have been used to identify a complex system relating vestibular and proprioceptive stimuli to body movements and ground reaction forces (Johansson *et al.*, 2001). The subject was modelled as a stabilised, segmented inverted pendulum. Using pseudo random binary sequences, electrical stimulation of the vestibular nerve and vibration of both calf muscles provided inputs to the model. Outputs of the model were forces on the force platform and the position of two point markers located at the navel and the sternum. Identification of the multi-input multi-output transfer function produced a statistically valid 30th order model that was able to predict the forces on the force platform and the position of the two markers. The model could predict the outputs with mean variance errors of approximately 10% which represents an impressive achievement.

In response to criticisms of their 1998 paper, Winter et al published further data showing the results of linear regression between ankle torque and sway angle of the CoM for ten second periods of sway (Winter *et al.*, 2001). The mean coefficient of determination ($r^2 = 0.92$) shows a clear linear relationship between ankle torque and CoM angle which supports the inverted pendulum model for quiet standing. However, these authors claimed firstly that the coefficient relating ankle torque to sway angle is a measure of the intrinsic mechanical ankle stiffness and secondly that this stiffness is the same as the one they measured in 1998. The gradient only represents the intrinsic mechanical stiffness if the change in ankle torque produced results solely from mechanical stretching of the muscle. Over the ten second period of sway there is considerable neural modulation of muscle activity relating to sway angle. This is evident from their figure 2 which shows ankle torque changing non-elastically with ankle angle, i.e. ankle torque decreasing as ankle angle increases, and so the measurement does not reflect intrinsic stiffness. Secondly the stiffness measured in 2001 was 858.9 Nm rad^{-1} which is 109% relative to the gravitational spring of 789.4 Nm rad^{-1}; whereas in 1998 the ankle stiffness was effective stiffness 802 Nm rad^{-1} + a gravitational spring stiffness of say 789.4 = 1591.4 Nm rad^{-1} which is 202% relative to the gravitational spring. The two measurements are clearly different and would have different effects on the human inverted pendulum. Winter's conclusion that the intrinsic ankle stiffness is sufficient for stability is again not justified from the data.

The idea of balancing an inverted pendulum as an equivalent body to one's own centre of mass was invented by Fitzpatrick et al as a means of manipulating the sensory information available to a subject (Fitzpatrick *et al.*, 1992b). This experimental arrangement has recently been repeated in a modified form. A computer model of an inverted pendulum has been balanced using the ankle musculature with the subject strapped to a vertical support (Fukuoka *et al.*, 2001). The ankle torque generated by the subject is recorded and the footplate is rotated by a servo-mechanism to mechanically replicate the antero-posterior movement of the virtual inverted pendulum. These authors used perturbations applied via the footplate and a variety of experimental conditions to identify in isolation the somatosensory, the visual and the vestibular feedback responses to perturbation. They found that the somatosensory system was most consistent between subjects and they speculated that this system was thus the most automatic and dominant when a subject balances using the ankle strategy. It appears that the unstable inverted pendulum could be balanced using somatosensory feedback alone but not necessarily using visual feedback alone. For most subjects using visual feedback alone, the visual feedback gain was less than the toppling torque per unit angle of the inverted pendulum and the feedback response showed a phase lag. Using vestibular feedback alone, subjects could balance themselves using the experimental arrangement in which somatosensory and visual information was eliminated or at least suppressed.

A recent study has challenged whether posture in quiet standing is controlled as an inverted pendulum (Hodges *et al.*, 2002). During standing, respiration is associated with small movements of the thorax. The authors studied the extent to which movements of the lower limbs and pelvis compensate for these small movements of the thorax and abdomen that occur during respiration. While eleven subjects stood quietly on a force platform reflective markers were tracked using a high resolution VICON camera system. They showed that small angular movements (~ 0.5 deg) of the trunk and leg were phase locked with rhythmical respiratory movements at a frequency of ~0.23 Hz. This demonstrated the multisegmental mobility of the body during quiet standing and shows that control of the whole cannot be reduced to control of a single joint such as the ankle. However, the CoP movements were not correlated with respiratory movements. This means that effectively respiration is de-coupled from balancing of the CoM. The respiratory movements are compensated for by movements in the lower limb and pelvis and do not result in any movement of the body centre of mass. This confirms that so far as control of the centre of mass is concerned the movements associated with respiration do not invalidate the inverted pendulum approximation.

The role of ankle stiffness in quiet standing.

Whether one considers the upright human as an inverted pendulum or as a multisegmented structure in constant motion at all joints, it is clear that maintenance of upright balance presents a considerable problem of motor control. Motor control can be exercised at a variety of levels. These levels range from the intrinsic mechanical properties of the joints, ligaments and muscles, through the peripheral reflexes, long latency spinal and CNS reflexes up to and including the highest levels of the cerebral cortex which may or may not involve conscious deliberation. The extent to which higher levels of the nervous system are involved on a moment by moment basis in planning and calculation for maintaining balance is unknown. Equally, the extent to which control is delegated to intrinsic mechanical and peripheral reflex mechanisms is still being discovered. An approach to solving this question has been to start at the lowest levels in the hierarchy which are most accessible to investigation. These ideas are attractive because they relieve the higher nervous system of the complexities of control on a moment by moment basis. Hypotheses regarding control at the lowest levels have been developed and these can be reviewed in the context of quiet standing.

As Magnus is reported to have demonstrated, a cadaver placed on its feet will not stand. Rather it collapses to the ground. This demonstrates that entirely passive balancing is not possible for humans. A living nervous system is necessary for upright balance. It has also been observed that the body cannot be supported as a stacking column of bones because the centres of gravity of all the bones and the movement centres of the joints between them cannot be brought into a common line of gravity (Steindler, 1955). In the spine, ligaments in tension and some, usually intermittent, muscular activity is required. Generally, very little muscular activity is required for standing. According to accumulated results from electromyography, at its minimum, only soleus (ankle plantarflexor) and iliopsoas (deep hip flexor) require constant activity (Basmajian & De Luca, 1985). Necessary activity from a few other muscles, e.g. gastrocnemius, can be reduced to being intermittent.

As described above, Gurfinkel et al (1974) investigated whether the intrinsic stiffness of the ankle joint maintained by tonic activity of the triceps surae was sufficient to compensate for the small oscillations encountered in quiet standing. Using what today would be regarded as rough and ready methods he concluded that the intrinsic stiffness was sufficient for marginal stability and also concluded that there were no stretch reflexes in quiet standing.

An interesting demonstration of the entirely passive stability of the ankle joint is described by MacConaill and Basmajian. An osteoligamentous preparation including the foot, the tibia and fibula and the ligaments and capsule of the ankle is stood on a table with the tibia and fibula vertical. It will remain standing even if the table is hammered gently. Only if the table is hammered violently will the tibia and fibular fall and when they do fall they fall forwards (Macconaill & Basmajian, 1977). This shows that the pedicrural skeleton itself is stable rather than unstable. With the rest of the body added, it was argued that the weight transmitted through each leg would tend to flatten the tibial and talar conarticular surfaces by mutual pressure. Thus the ankle joint in standing was predicted to be at least metastable requiring minimal muscular stabilisation.

In an interesting paper Winters et al (1988) articulated the idea that postural position can be maintained by joint impedance. They defined impedance as the automatic capability of a system to resist a load without voluntary intervention. Their definition of impedance incorporates stiffness, viscosity and inertia. This understanding of impedance includes intrinsic mechanical impedance generated by cocontraction at a joint and also impedance resulting from spinal feedback mechanisms. It was suggested that several factors co-operate to prevent positional drift including (i) low gain spinal feedback (ii) viscosity resulting from the muscle's force v. velocity relationship at low speeds and (iii) the mild, passive friction like and stiction like properties a joint complex.

These authors also summarised the existing established body of stiffness regulation theories relating to slow postural movements. There are three types of theory that stabilise a joint at an equilibrium point (Hasan *et al.*, 1985). There is the (i) equilibrium point stiffness hypothesis (ii) the equilibrium point - invariant or λ-model hypothesis and (iii) the stiffness regulation hypothesis. Each theory is a subset of the equation $M = K (q - q_0)$ where M is the joint torque, K is the stiffness acting at the joint, q is the joint angle and q_0 is the offset of the spring-like element acting at the joint (Hasan, 1986). (i) For the equilibrium point stiffness hypothesis an equilibrium point is defined and maintained by cocontraction (Bizzi *et al.*, 1976; Polit & Bizzi, 1979). The intersection of the static, positive slope, muscular tension-angle relationship and the load line defines the equilibrium point. Because equilibrium is maintained by intrinsic mechanical properties of the muscle, no sensory feedback is required. (ii) The 'equilibrium point - invariant' or λ-model' hypothesis states that the bias of the spring, q_0, rather than the stiffness, K, of the spring is varied (Feldman, 1966, 1986). The CNS sets a given bias which is maintained for a period of time. This bias determines a particular spring like relationship between joint angle and torque which is maintained by both sensory feedback and intrinsic muscle properties. The external load applied determines the angle and torque at which equilibrium is attained. (iii) The stiffness regulation theory states that spinal reflexes operate to ensure consistent, linear spring-like behaviour when a transient external load is applied at a joint (Houk, 1979; Houk & Rymer, 1981). Of itself, the tension-length relationship of muscle is non-linear. The stretch reflex and the golgi tension reflex are said to operate in combination to maintain a constant stiffness that is set by the CNS. That stiffness cannot readily be changed to a new level by higher commands (Crago *et al.*, 1976).

A thorough review of the role of intrinsic muscle impedance in joint control was given in 1990 (Hogan). The main idea is that intrinsic muscle impedance generates force extremely rapidly in response to imposed disturbances. For example, the intrinsic stiffness of muscle is responsible for the earliest response of the lower limb to the disturbances encountered in walking (Grillner, 1972). Modulating joint impedance is a possible way of controlling stability. For example, in the control of an inverted pendulum, coactivation of the agonist and antagonist muscles is a way of increasing impedance because the net stiffness at the joint is the sum of the stiffness of each muscle. Since the net moment at the joint is the difference between the agonist and antagonist contributions, for any joint moment a range of joint stiffnesses can be maintained. Such stiffness does cost extra energy to maintain and is thus an energetically inefficient strategy. Since it is a biological principle that energy is not squandered needlessly one would expect co-activation to be kept to a minimum. It was argued that there must be some compromise between minimising energy and maintaining stability.

The idea that an inverted pendulum may be stabilised by cocontraction was further developed (Shadmehr & Arbib, 1992). As part of their paper these authors argued that the minimum stiffness required for stability of the inverted pendulum could be maintained by cocontraction provided the muscle stiffness grows at least linearly with force under isometric conditions. This requirement was necessary because of the change in moment arm with joint angle. They developed Feldman's model to suggest that the CNS controls the offset or resting length of a non-linear spinal reflex spring in which stiffness grows non-linearly with force. An advantage of this reflex spring is that higher stiffness can be produced with smaller muscle forces and thus with lower metabolic cost.

In the same year Fitzpatrick et al claimed that the visual, vestibular and lower limb sensori-motor reflexes make a large contribution to ankle stiffness in quiet standing (Fitzpatrick *et al.*, 1992b). Moreover, these authors claimed that the local sensory reflexes alone provide greater ankle stiffness than the minimum required for marginal stability and thus are sufficient for standing. They produced results showing that this reflex stiffness can be altered by the intentional set of the subject.

In a review of postural control Horak and MacPherson (1996) drew attention to the role of passive stiffness of connective tissue within and around muscle in counteracting the unbalancing effect of gravity. In particular they emphasised the passive and intrinsic stiffness associated with the tonic activity of the soleus muscle and described its ability to reduce sway in quiet stance. Intrinsic stiffness allows significant development of force virtually instantaneously and long before any peripherally centrally driven signal could dynamically change muscle activation. A consequence of the stabilising effect of the intrinsic and passive stiffness is that the amount of muscular activity needed in quiet standing is rather small.

In 1996 Fitzpatrick et al contradicted their 1992 conclusions and published evidence showing that reflexes are unable to explain the stability of normal human standing because the gain of the reflex negative feedback loop was approximately unity. Thus a feedforward or predictive process was necessary.

In 1998 Winter et al published results advocating the idea that the CNS merely sets an appropriate muscle tone and stiffness in the triceps surae and then allows the intrinsic mechanical stiffness to maintain stability without any need for reaction from the nervous system. If the CNS wanted to decrease the amount of sway then it would increase the amount of ankle stiffness since it was claimed that sway varies inversely with the square root of effective ankle stiffness.

The following year Gatev et al (Gatev *et al.*, 1999) produced evidence showing that gastrocnemius activity is neurally modulated in correlation with sway and 300 ms in advance of CoM sway. These authors then advocated a central predictive modulation of ankle stiffness to control sway. They suggested that ankle torque is predictively increased and decreased in a spring-like manner to control sway and that the effective stiffness is increased to decrease sway and vice versa.

In the same year a determined case was made against the role of intrinsic ankle stiffness in stabilising sway during quiet standing (Morasso *et al.*, 1999; Morasso & Schieppati, 1999). It was claimed from published data that intrinsic ankle stiffness was insufficient to stabilise the human inverted pendulum and that active stabilisation was needed. A Smith predictor model was suggested and it was further suggested that the Smith predictor function might be located in the cerebellum.

The idea that the CNS modulates ankle stiffness to control sway was advocated again by Carpenter et al (1999). When subjects stood on a high platform the mean power frequency of the CoP increased and this was interpreted as meaning the CNS was increasing the ankle stiffness to deal with the perceived increased threat of falling.

The purpose of studying human balancing of an inverted pendulum

Following the lead of Fitzpatrick et al (1992b) a real human proportioned inverted pendulum has also been built in our laboratory. The subjects are strapped to a fixed vertical support and use their ankle musculature to balance the inverted pendulum which has a mass and inertia equivalent to an adult human. Details of this apparatus are shown in Figure 1 of **Chapter** 2.

This apparatus enables study of fundamental principles of motor control at the integrative level but in a setting that is relatively simple with a reduced number of variables. For example there is only one joint axis through the ankles and the pendulum only moves in the parasagittal plane. Thus the musculature used specifically for the task is reduced to that crossing the ankle joint. There is only one moving mass, the pendulum, and the position of that can be precisely and accurately measured. Because one is studying merely the control of the pendulum CoM about an ankle axis a variety of interesting questions can be studied relatively easily. For example, (i) is the pendulum balanced at one angle or a range of angles? The inverted pendulum is usually considered to be balanced at the vertical position or to be balanced at a single equilibrium angle. (ii) Is balance of the pendulum a passive or an active process? I.e. is there any obvious neural modulation that corresponds with balancing on a sway by sway basis? (iii) What is the source of pendulum sway? And how does pendulum sway compare with sway in standing? (iv) Do subjects modulate ankle stiffness in order to control sway? Is there any evidence that effective system stiffness or reflex stiffness or intrinsic mechanical stiffness is increased to decrease sway? (v) To what extent does the intrinsic stiffness of the activated ankle musculature contribute to stabilisation of the inverted pendulum? (vi) Is the position control process continuous or intermittent? (vii) Is control of pendulum position a low level task that is delegated by the CNS to muscle properties or spinal reflexes or cerebella reflexes or is control of the pendulum a high level task that requires long loop, trans-cortical processing. (viii) What are the effects of removing visual feedback so that the subject is balancing the pendulum using proprioceptive feedback and pressure from the soles of the feet only?

Human balancing of an inverted pendulum is also worth studying as a comparison with quiet standing. While standing and balancing the pendulum are different activities, enough researchers have analysed quiet standing using an inverted pendulum model to make comparison between the two tasks very fruitful. Theories about ankle mechanisms used in quiet standing can be tested more easily with this apparatus. Standing is a very complicated activity. Standing is biomechanically complicated because of the large number of joints at which movement occurs. Standing is neurally complicated because of the number of moving parts to be controlled and because of the multiplicity of sensory information to be integrated. It is useful to study something closely related to standing but a lot simpler. With the pendulum balancing task movement is restricted to one degree of freedom about the ankle joint. Internal perturbations due to breathing and heart beat are also eliminated. The amount of sensory information available to the subject can also be reduced. Vestibular information is not available because the subject is strapped and does not move. Likewise, shearing force information from the sole of the foot is not available because the subject's CoM does not accelerate horizontally. Visual information can be reduced by screening the pendulum. Proprioceptive information from the legs is retained and exteroceptive information from pressure on the sole of the foot is available. Tactile information from the areas of the trunk in contact with the support was also available, but is likely to be inconsequential (Fitzpatrick et al. 1992b).

Any insights gained concerning human balancing of the inverted pendulum are interesting of themselves and can also be compared with quiet standing without implying that quiet standing follows the inverted pendulum model. If the same processes are observed in quiet standing as in balancing the pendulum this tends to emphasis the inverted pendulum like nature of quiet standing. If different processes are followed then this is an indication of how the activities are different and this sheds light on the inadequacies of the inverted pendulum model applied to quiet standing.

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CHAPTER 2.

HUMAN BALANCING OF AN INVERTED PENDULUM: IS SWAY SIZE CONTROLLED BY ANKLE IMPEDANCE?

Summary.

1. Using the ankle musculature, subjects balanced a large inverted pendulum. The equilibrium of the pendulum is unstable and quasi-regular sway was observed like that in quiet standing. Two main questions were addressed. Can subjects systematically change sway size in response to instruction and availability of visual feedback? If so, do subjects decrease sway size by increasing ankle impedance or by some alternative mechanism?

2. The position of the pendulum, the torque generated at each ankle and the soleus and tibialis anterior EMG were recorded.

3. Results showed that subjects could significantly reduce the mean sway size of the pendulum by giving full attention to that goal. With visual feedback sway size could be minimised significantly more than without visual feedback. In changing sway size, the frequency of the sways was not changed.

4. Results also revealed that ankle impedance and muscle co-contraction were not significantly changed when the sway size was decreased. As the ankle impedance and

sway frequency do not change when the sway size is decreased, this implies no change in ankle stiffness or viscosity.

5. Increasing ankle impedance, stiffness or viscosity are not the only methods by which sway size could be reduced. A reduction in torque noise or torque inaccuracy via a predictive process which provides active damping could reduce sway size without changing ankle impedance and is plausible given the data. Such a strategy involving motion recognition and generation of an accurate motor response may require higher levels of control than changing ankle impedance by altering reflex or feedforward gain.

Introduction

In this study subjects balanced an artificial inverted pendulum equivalent in mass and inertia to a medium sized woman. Motion was restricted to one joint and one plane and sensory feedback could be limited to the proprioceptive system by eliminating visual feedback and preventing movement of the body. It provided a mechanically simple way of studying the ankle mechanisms employed in balancing which eliminated the multi-joint dynamics of real standing. Normal standing is a complex activity both mechanically and neurologically. Our approach was to reduce and simplify the complex system in order to gain an insight into underlying principles. The inverted pendulum balancing task has previously been compared with normal standing (Fitzpatrick et al. 1992a, b, 1994a, b, 1996; Fitzpatrick & McCloskey, 1994). These authors concluded that the task was broadly equivalent to real standing.

In standing, the body has been considered as an inverted pendulum (Gurfinkel & Osevets, 1972). Using a multi-target movement tracking system, Winter *et al.* (1998) recently validated the inverted pendulum model for sagittal sway. While corroborating earlier observations that greater angular changes occur at the hip joint than the ankle (Day et al. 1993), Gatev *et al.* (1999), using a multi-camera system, have demonstrated that ankle mechanisms dominate in the sagittal plane with an almost synchronous sway of the body parts. Authors who work with more sophisticated models of standing still regard the inverted pendulum model as capable of capturing the gross dynamics of posture control (Nicholas et al. 1998).

It is evident that the extent of postural sway is not fixed. Some subjects sway more than others and in any one subject size of sway can be altered by changing sensory input (Paulus et al. 1984) or by volition (Fitzpatrick et al. 1992a, b). The role of joint stiffness in the maintenance of posture has been debated for some time (Grillner, 1972; Horak & MacPherson, 1996). There have been frequent suggestions that sway size is reduced by increasing ankle stiffness (Fitzpatrick et al. 1992b; Winter et al. 1998; Carpenter et al. 1999; Gatev et al. 1999).

Fitzpatrick *et al.* (1992b) applied slow, imperceptible perturbations to five subjects while standing or when balancing an inverted pendulum. When subjects were instructed to stand still, or had more kinds of sensory feedback available, sway in response to perturbations was reduced and ankle stiffness was increased. The increased stiffness was attributed to increased reflex gain.

Winter *et al.* (1998) measured the centre of pressure and centre of mass oscillations of ten quietly standing subjects. They argued that the relationship of these two parameters could be explained by a non reactive, simple elastic model. With this model, sway size would be predicted to be inversely proportional to the square root of stiffness (K_e). Values for anterior/posterior (A/P) oscillations are not given, but in the medial/lateral (M/L) plane sway size was observed to be proportional to $K_e^{-0.55}$ which was close to their theory.

Gatev *et al.* (1999) demonstrated feedforward modulation of gastrocnemius activity with seven standing subjects. He hypothesised a central, predictive control of ankle stiffness working to restrict sway size with the activated gastrocnemius muscle working in a spring-like manner. Whereas Gatev and Winter found no effect of vision on quiet standing, Fitzpatrick (1992a,b) and Carpenter *et al.* (1999) found conflicting evidence that vision does increase ankle stiffness.

Some experimenters have used relatively large and abrupt disturbances such as moving the platform on which the subject stands. Such experiments provide insight into the mechanisms which respond to a gross loss of balance (Horak et al. 1989; Bloem et al. 2000). Such events are rare in the usual experience of standing and do not illustrate the patterns of muscular activity that are repeated over and over in quiet standing. Fitzpatrick *et al.* (1992b) used small, slow perturbations to measure ankle stiffness while subjects maintained their own balance or balanced an inverted pendulum. These disturbances which were not consciously detected by the subject were comparable in size to normal sway and thus portrayed the standing process more closely.

We employed an alternative approach. In normal balancing there are frequent minor departures from equilibrium. By examining the response to these spontaneous departures from equilibrium the need for external perturbations is obviated. By detecting the losses of balance as they occur naturally, and by averaging many examples of the responses, the impedance of the ankles can be determined. (The impedance is to be preferred to stiffness as the resistance to motion at the ankles involves elastic and viscous components and is frequency dependent).

The aim of this investigation is to address two main questions. Can subjects systematically reduce sway size through their own volition or by use of visual feedback? If so, do subjects reduce sway size by increasing ankle impedance?

Methods.

Subjects.

Ten healthy people, including 6 males, aged between 18 and 45 consented to be subjects in these simple non-invasive experiments for which approval had been given by the local human ethics committee.

Apparatus

Subjects were strapped round the pelvis to a vertical support that effectively eliminated their actual sway (Fig. 1). The subject stood on two footplates with their ankles positioned to be co-axial with the axis of rotation of an inverted pendulum. The footplates were exactly horizontal when the backward lean of the pendulum was 3 degrees, thus approximating typical forward lean in standing. The pendulum had a mass of 61.65 kg with a centre supported 0.937 m from the axis of rotation (distance 'h'). The subject balanced the inverted pendulum which was free to move forwards and backwards in a parasagittal plane while always tending to topple backwards. The same mass and distance 'h' was used for all subjects. The constant static and dynamic properties of the pendulum presented each subject with an identical task. This allowed results from all subjects to be pooled. The toppling torque of the pendulum was measured to be 10.2 ± 0.4 Nm deg^{-1}. Using a spring of known stiffness and by recording the damped oscillations of the pendulum, the moment of inertia was determined to be 62.6 ±2 kg m^2 (which included the contribution of the rod and other rotating parts), the viscous damping was 0.061 ± 0.02 Nm s/deg, and friction was 0.045 ±0.1 Nm.

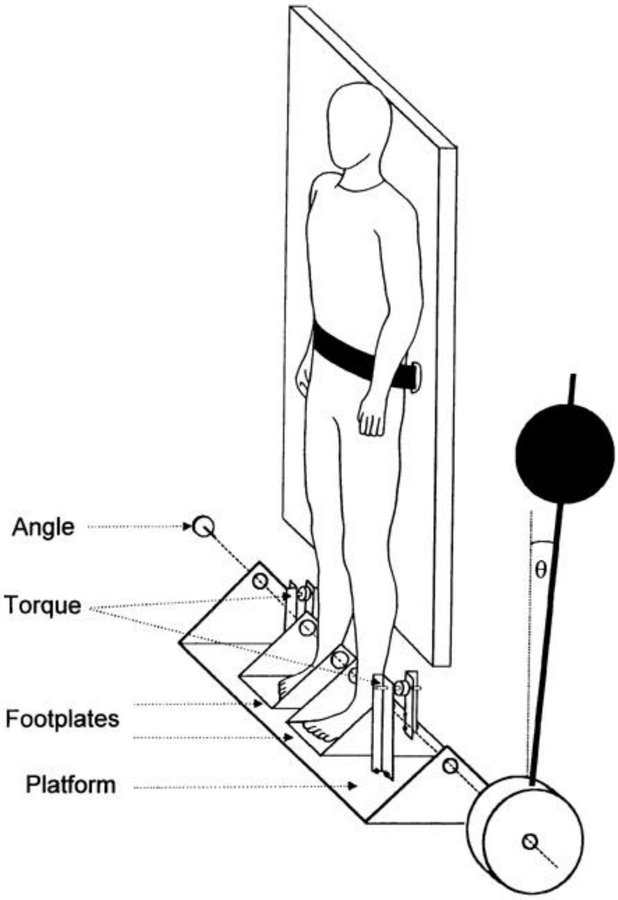

Figure 1. Inverted Pendulum Apparatus.
Subjects balanced a backward leaning, real inverted pendulum of mass and inertia equivalent to a medium sized woman. The subjects were unable to sway since they were strapped round the pelvis to a fixed vertical support. The axis of rotation of the pendulum, platform and footplates was co-linear with the subject's ankles. Force exerted by the subject's ankle musculature onto each footplate was transmitted by horizontally mounted load cells. These measured the torque that each leg applied to the pendulum via the rigidly attached platform. The footplates and the platform were independently mounted on precision ball races. A precision potentiometer measured sway of the pendulum. Absolute angle of the pendulum (θ) and angular velocity were measured by an electronic inclinometer and a solid state gyroscope (not shown).

The relative angular position of the pendulum was measured using a Hall effect precision potentiometer (with an effective range of 15 degrees), (CP 2UTX, Midori Precision Company Ltd, Japan) and fixed gain amplifier. Absolute angular position was measured using an electronic inclinometer (Cline R1, Cline Labs Inc. USA), of resolution 0.001 deg, attached to the base of the pendulum. The angular velocity of the pendulum was measured using a piezoelectric vibrating gyroscope (range ± 90 degrees s^{-1}), (ENV - 05 A+C, Murata Co Ltd, Japan) in conjunction with an instrumentation amplifier. The subject exerted torque on the pendulum via each footplate. The left and right torque signals were recorded using horizontally mounted miniature load cells (Sensotec model 31, Sensotec Inc, USA) followed by a 2 channel bridge amplifier and low pass filter (Sensotec UBP, Sensotec Inc, USA). The load cells were mounted in compression in the horizontal plane. One end of the transducer was rigidly bolted and the other made contact with a polished surface. This method of mounting effectively decoupled the load cells from off-axis loads and prevented the slight deflection of the structure caused by the subject's weight from producing a signal which would be falsely registered as a torque (Kelly, 1998). Electromyographic (EMG) activity from the right and left tibialis anterior and soleus muscles was recorded using home constructed bipolar surface electrodes with encapsulated preamplifiers (Johnson et al. 1977). These signals containing the entire bandwidth were then amplified and passed through an analogue full-wave rectifier and r.m.s. averaging filter with a time constant of 100 ms. Data from all sensors were recorded by computer, sampled at 25Hz via an analogue to digital converter (CED 1401, Cambridge Electronic Design, UK). The resolution of the recorded data was limited by input noise levels of less than 0.002 deg, 0.02 deg s^{-1}, 0.03 Nm, 60 μV for relative angular position, angular velocity, right or left torque and surface EMG respectively.

Experimental protocol.

The inverted pendulum apparatus has been designed to study the effect of limiting the kinds of sensory inputs influencing ankle mechanisms used to control upright balance (Fitzpatrick et al. 1992b). Since the subjects themselves were prevented from swaying, vestibular feedback was not available to them. The pendulum mass and rod were screened from view though an oscilloscope was available providing the option of visual feedback regarding the position of the pendulum. The oscilloscope was one metre away from the subject and had a gain of one centimetre deflection per degree change in angular position. When visual feedback was not used the oscilloscope was turned off though subjects still had their eyes open. Proprioceptive information from the legs was available to all subjects. Tactile information from the areas of the trunk in contact with the support was also available, but is likely to be inconsequential (Fitzpatrick et al. 1992b).

In four separate trials, subjects were asked to balance the inverted pendulum under differing instructions and visual conditions. The four trial conditions were:

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

The order in which the four conditions were carried out was randomised. The duration of each trial was 200s. In all cases the subjects were asked to keep the pendulum between 0.5 and 5.5 degrees from the vertical so as to approximate standing sway.

It was explained that "Stand still" meant to reduce the sway of the pendulum to an absolute minimum and to keep the pendulum at the same angle. Subjects were told that "Stand easy" meant to balance the pendulum while giving the least possible attention to the sway of the pendulum. When subjects were "standing still" they were encouraged to give their full attention to the oscilloscope when that was turned on 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.

All subjects were given a preliminary experience of balancing the pendulum at different angles ranging from 1 degree to 5 degrees using visual feedback from the oscilloscope. They also practised balancing the pendulum without the use of visual feedback. The subjects then reported the angle at which they preferred to balance the pendulum. This was around 3 to 4 degrees for all subjects. For each trial, recording started with the pendulum at the preferred angle of the subject.

Principles and methods of data analysis.

During balancing, the pendulum sways to and fro in a quasi-regular fashion. We identified the times at which the pendulum reversed direction by interpolating between the data points when the velocity changes sign. The unidirectional movement between one turning point and the next was categorised as a sway. For any trial, the mean sway size was the average magnitude of the sways. Mean sway frequency was calculated as the total number of sways, (positive and negative) divided by two and divided by the trial duration.

Gravity exerts a torque on the inverted pendulum given by $T_g = K_{tt} \sin\theta \approx K_{tt}\,\theta$ where K_{tt} is the gravitational toppling torque per unit angle. At any angle θ from the vertical, this formula defines the ankle torque that is required to balance exactly the pendulum. (Dynamic torques due to frictional and viscous damping of the pendulum are very small). On a plot of torque v angle this formula defines a line of unstable equilibrium (which has also been called the load stiffness of the pendulum (Fitzpatrick et al. 1992b)). To keep the sway size between certain limits, the ankle torque must be repeatedly alternated above and below this line. Line crossings represent repeated events around which data can be averaged and from which ankle impedance at equilibrium can be measured.

Using Savitzky-Golay filters, the position data were double differentiated to produce a record of acceleration (Press et al. 1999). From Newton's second law of motion the angular acceleration is zero at equilibrium. Equilibrium points represent moments at which the subject perfectly balances the static and dynamic torques exerted on them via the pendulum. We identified those equilibrium points when zero acceleration was crossed by interpolating between the data points when the acceleration changes sign.

In each unidirectional sway a spring like equilibrium occurs at least once. This is represented by a positive gradient of torque v angle crossing the line of equilibrium. These equilibria were identified by an acceleration changing from positive to negative while the pendulum was falling, or negative to positive while the pendulum was rising. The data surrounding these equilibrium points were averaged to show the mean responses. The impedance at these averaged equilibrium points was calculated as the regression value for Δtorque/Δangle encompassing one data point (40 ms) either side of the equilibrium. This method of identifying equilibrium points, sampling around these points and averaging is illustrated in Figure 2.

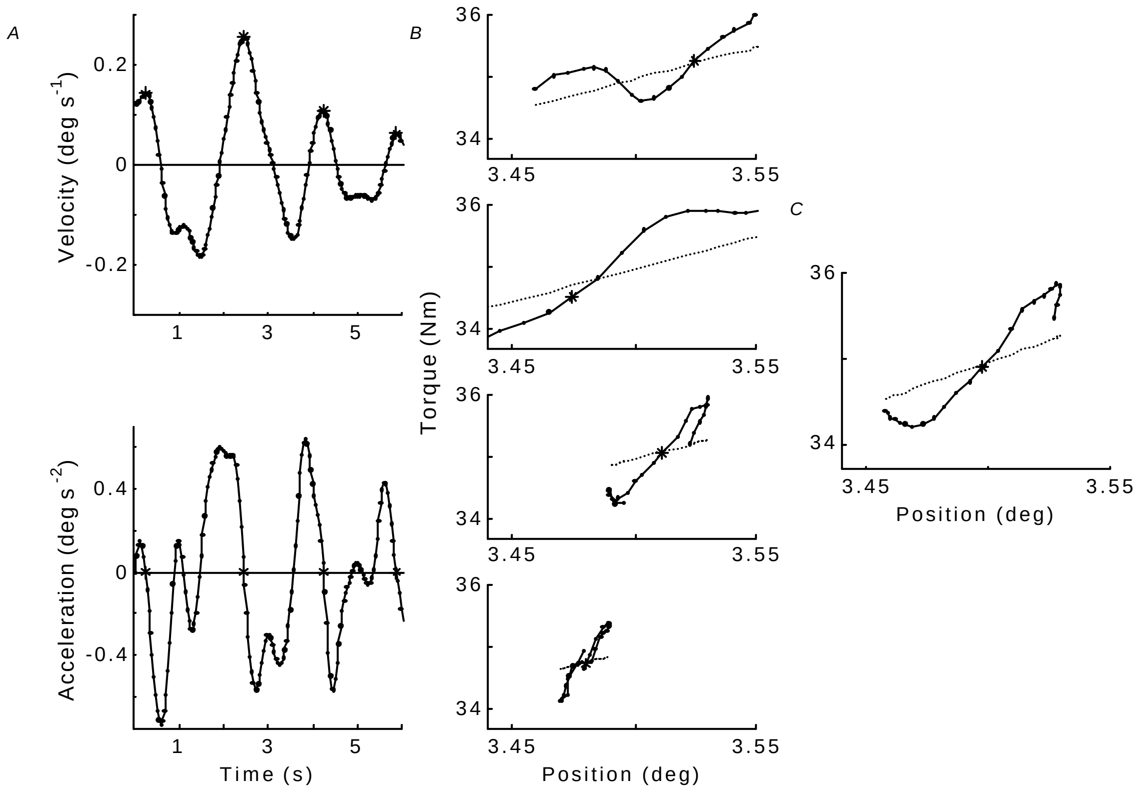

Figure 2. Illustration of the line-crossing averaging process.
Panel ***A*** shows a six second record of angular velocity and angular acceleration against time for a representative subject. Equilibrium times are identified by interpolating between the pairs of acceleration data points that cross zero. From these equilibrium times are selected those that occur while the acceleration is passing from positive to negative and while the velocity is positive (i.e. the pendulum is falling). These equilibrium times are shown as *. Ankle torque and pendulum position records are sampled at 0.04 s intervals for up to five seconds before and after these selected equilibrium times. The four selected equilibrium times in panel ***A*** are shown in panel ***B***, together with ± 0.48 s of surrounding data, plotted as ankle torque against pendulum position. The straight dashed lines represent the line of equilibrium, $T_g=K_{tt}$ *sinq*, (load stiffness). The selected equilibria represent falling (increasing angle), spring-like (positive gradient) line-crossings with an ankle impedance (Δtorque/Δangle) greater than the load stiffness. The four 0.96 s records shown in ***B*** are averaged to produce the record shown in panel ***C***. The rising, positive-gradient line-crossings are selected and averaged in an analogous manner.

This line-crossing impedance is a biased measure in that the ankle impedance of high frequency sways will be greater than that for low frequency sways. So, to investigate the effect of frequency, we have grouped the line crossing equilibria into frequency bins and then separately averaged the line-crossings for each frequency category. The appropriate frequency bin for each line-crossing was determined from the duration between the nearest reversal points surrounding the line-crossing, using the formula *frequency = 1/(2 * inter-reversal duration).*

The position and velocity records possess small, relatively high frequency variations of the same magnitude and frequency as the noise that is recorded when a subject is not standing on the footplate to balance the pendulum. Given the large inertia of the pendulum, these variations are taken to be a noise product of the measuring and recording process and were eliminated by smoothing as part of the differentiation process. A Savitzky-Golay Filter algorithm was used which assumes that the noise is normally distributed and independent of the slowly changing variable and that a moving polynomial can be fitted to the data (Gander & Hrebicek, 1997; Press et al. 1999). This algorithm was effectively used as a low pass filter with a bandwidth of 3 Hz and zero phase shift. Data from the velocity and position sensors were cross-checked to corroborate the differentiation and smoothing process.

Modelling.

We wanted to know the effect of ankle stiffness, viscosity and noise on our line-crossing measure of ankle impedance, on sway size and on sway frequency. A second order model of the inverted pendulum was constructed for simulation purposes as described in the appendix. Torque generated at the ankles was modelled as having a stiffness component, a viscous component and a noise component as described by Winter *et al* (1998). Results generated from this model were subjected to the same analysis procedures used for real data.

Results.

All the subjects could balance the pendulum for an adequate period after minimal familiarisation although some found the task easier than others.

The complex variation of ankle torque with pendulum angle for a representative subject over a 12-second period is plotted in Figure 3. As seen from the 'line crossings', there was no one position of equilibrium though the irregular sway was confined to a small range of angles. At all angles the ankle torque stayed close to the line of equilibrium (approximately ± 1Nm). The torque alternated above and below that required for equilibrium. A given sway size of the pendulum can be maintained by the subject exerting an ankle torque close to the equilibrium line for a long time or far from the equilibrium line for a short time. The torque v angle gradient (instantaneous ankle impedance) was usually steeper than the equilibrium line (load stiffness) and appeared to have some consistency. The equilibrium line was sometimes crossed with a negative gradient, which permitted sustained positional drift while maintaining balance.

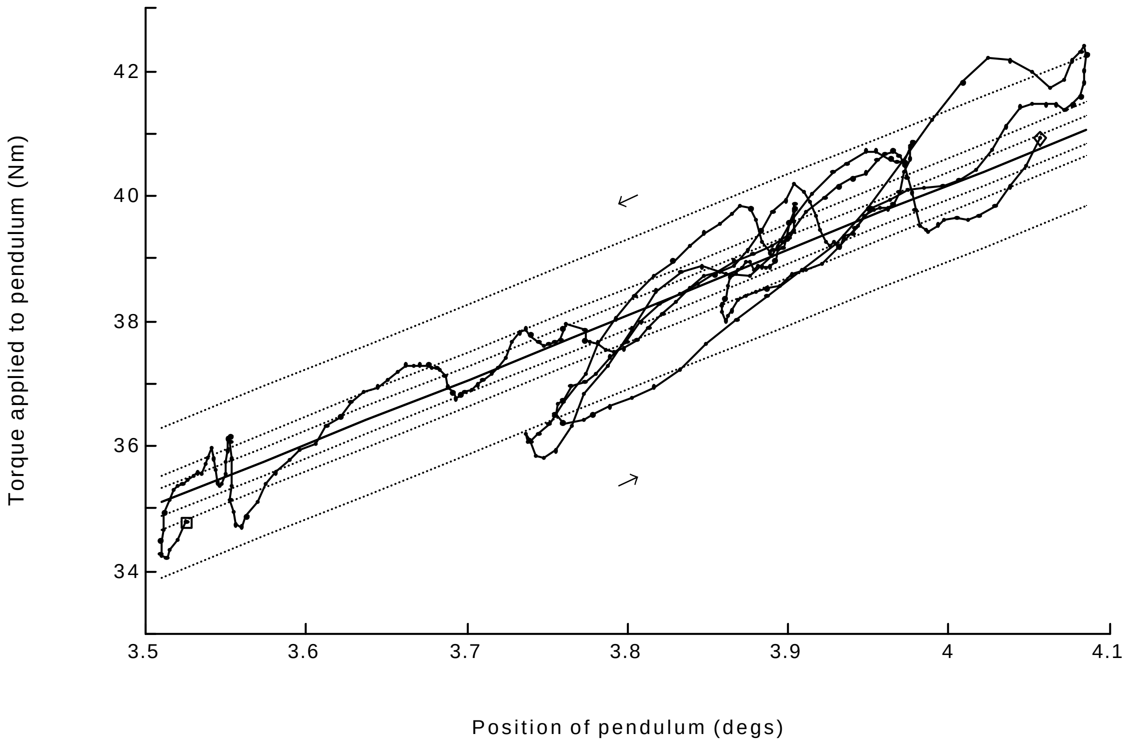

Figure 3. Representative sway of one subject.
A 12 s record from one subject is plotted as combined ankle torque against pendulum position. Data points are at 40 ms intervals. The starting point ◇ and finishing point ☐ are indicated. The line of equilibrium, load stiffness, ($T_g=K_{tt}\ sinq$) is shown as a continuous straight line. Dashed lines parallel to this represent lines of constant torque error. Torque error produces a directly proportional acceleration of the pendulum in the direction indicated by the arrows. The inertia of the pendulum effectively "absorbs" the torque; even with the largest torque error shown it will take 0.6 s for the deflection of the pendulum to reach 0.3 degrees (this trial's average sway size) from rest. The smaller and smallest torque errors are associated with times of 1.0 s and 1.4 s respectively.

Figure 4 shows that the conditions of the trial did make a difference to the sway size. Panel A shows a histogram of sway sizes for a representative subject recorded under each of the four trial conditions. For each condition, most sways were small. There were fewer large sways and more small sways when the subject was attempting to stand still rather than stand easy (1 v 2 and 3 v 4) and sways were slightly smaller when visual feedback was allowed (1 v 3 and 2 v 4).

Figure 4 (panel B) shows that there was a significant difference in sway size between the four trial conditions. (Two way ANOVA, N=2040, F=41.4, P<0.001). The mean trial sway size from the first 51 sways of each trial was averaged over ten subjects for each of the four conditions. Subjects could significantly minimise their sway size when 'standing still' as opposed to 'standing easy' (1 v 2 and 3 v 4, Tukey, 95% simultaneous confidence intervals (SCI)). Subjects could minimise their sway size significantly more with visual feedback (1 v 3, 95% SCI). Visual feedback made less difference to the sway size when subjects were 'standing easy' i.e. giving minimal attention to sway (2 v 4). Interestingly, the trial conditions caused no significant change in the sway frequency (panel C). (Two way ANOVA, N=40, F=0.89, P=0.46). This implies that the amplitude but not the duration of the sways is being changed by the intent or visual conditions of the subject.

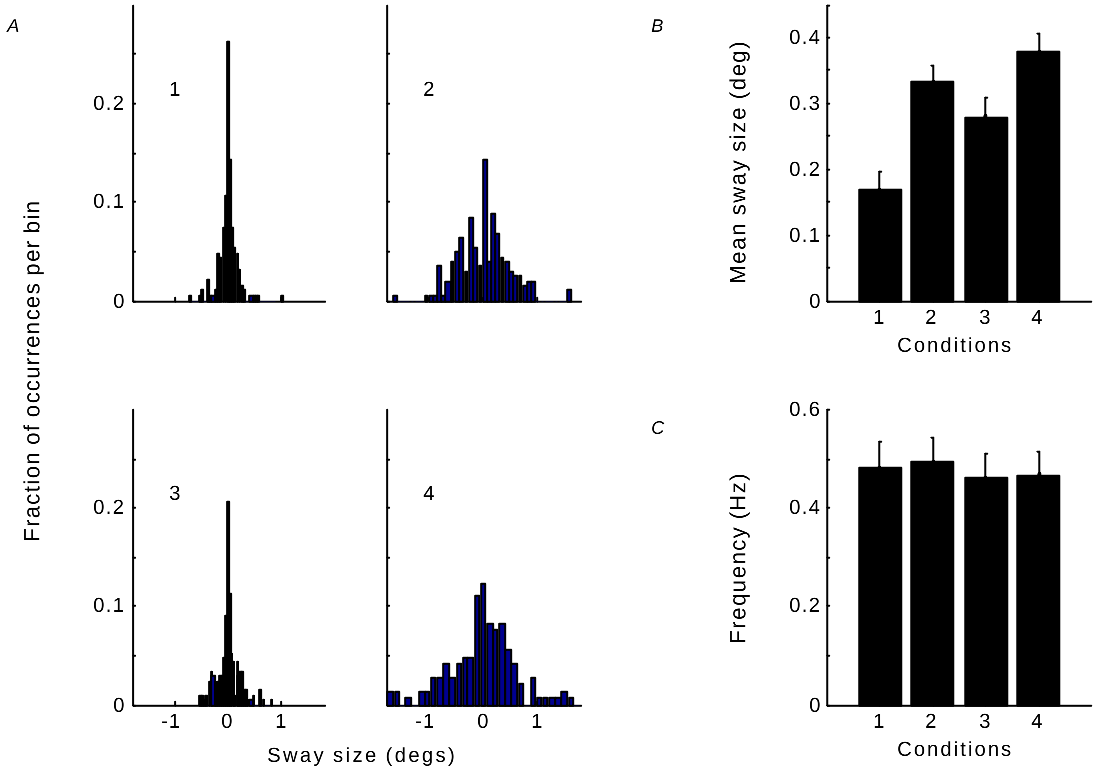

Figure 4. Effect of trial conditions on sway size.

Panel ***A*** shows histogram distributions of sway sizes for one subject under each of four trial conditions labelled 1, 2, 3, 4 as described in the methods section. Each trial lasted 200 s. A sway size was the angular displacement between successive turning points of the pendulum. Trial conditions were 1. Stand still with visual feedback. 2. Stand easy with visual feedback. 3. Stand still with no visual feedback. 4. Stand easy with no visual feedback. For each trial condition, panel ***B*** shows the mean, trial, sway size and panel ***C*** shows the mean sway frequency. For both panels, values were averaged over ten subjects for each of the four trial conditions. Error bars show 95% confidence intervals for the mean values.

Ankle impedance was not changed to bring about a reduction in sway size (Figure 5). Panel A shows representative data for one subject in each of the four trial conditions. Data were averaged from 1.3 s before to 1.3 s after all positive gradient 'line crossings' while the pendulum was falling. Data points are shown at 40 ms intervals. The same basic pattern of torque v angle is seen for each condition. The width of the pattern, e.g. inter-reversal change in position (0.07, 0.23, 0.09, 0.34 degrees respectively) ranks in the same order as mean trial sway size. The 'line crossing' impedance was approximately the same for each condition. The velocity at equilibrium was less when the sway size was less and the stationary points (a, b) were closer to equilibrium (less torque error) when the sway size was less.

The detailed shape of the curves in Figure 5A reveals several interesting insights into the balancing process. Unlike undamped simple harmonic motion, the pattern of ankle torque v angle was not a straight line. Indeed the pattern is not strictly an oscillation with the same starting and finishing point: it shows a net change in position of equilibrium which can be thought of as a positional step. The torque did not always change in phase with angle. This indicates modulation of ankle torque that does not depend solely on elastic forces and which is partly in phase with velocity with a consequent energy absorbing effect. The average velocity of the preceding and subsequent equilibrium ('line crossing') was reduced (almost zero for "stand still") which illustrates the effect of damping. However, there was no increase in gradient at maximum velocity (equilibrium) which shows that this is not simple viscous damping. This same pattern was repeated when the pendulum was rising rather than falling. No difference from this pattern has been seen in any trial.

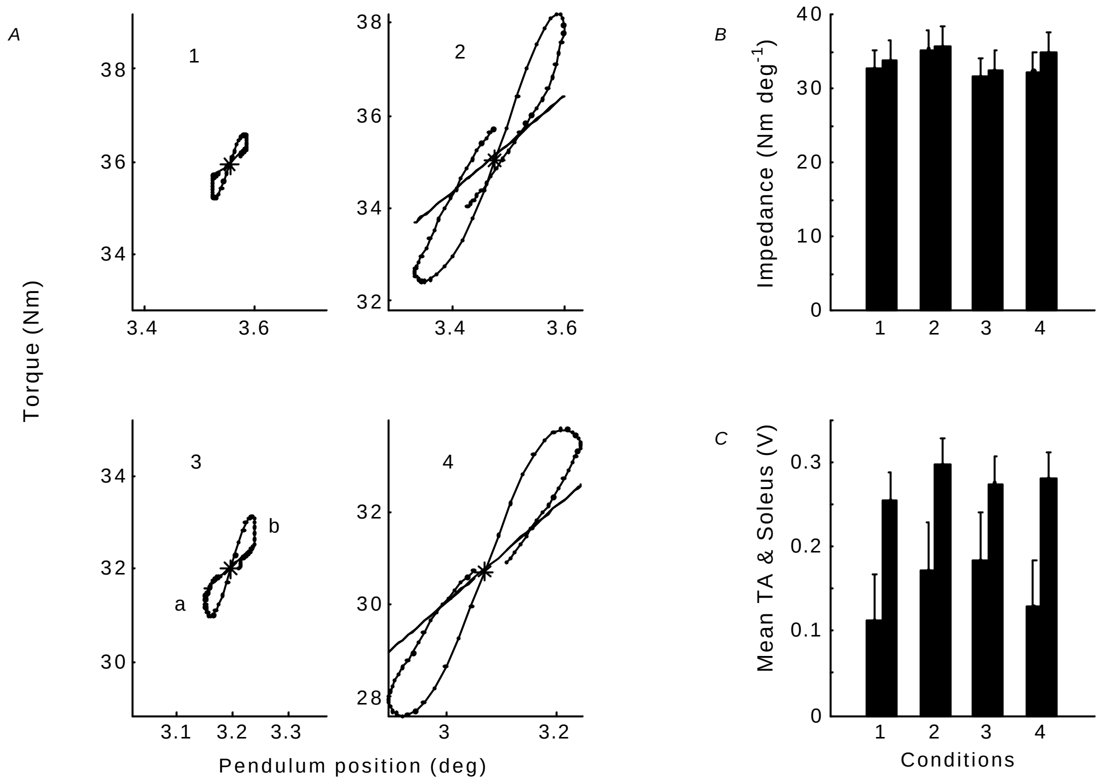

Figure 5. Effect of trial conditions on ankle impedance.

Panel ***A*** shows data averaged from 1.3 s before to 1.3 s after each positive gradient, equilibrium line crossing while the pendulum was falling: this is for one representative subject under each of four trial conditions labelled 1, 2, 3, 4. Combined ankle torque is plotted against pendulum position with the same scaling for each graph. Trial conditions are the same as Figure 3. Data points are at 40 ms intervals and proceed from label 'a' to 'b'. The line of equilibrium (ignoring pendulum friction) is shown as a dashed line. The asterisk marks the point of equilibrium and maximum velocity. For each trial condition, panel ***B*** shows the mean, positive gradient, line crossing impedance (pendulum falling - left bar, rising - right bar) and panel ***C*** shows the mean EMG activity summed over both legs (tibialis anterior - left bar, soleus - right bar). For both panels, values were averaged over ten subjects for each of the four trial conditions. Error bars show 95% confidence intervals for the mean values.

Figure 5 (panel B) shows there was no significant difference in ankle impedance between the conditions of the four trials. (Two way ANOVA, N=40, Falling F=1.67, P=0.20, Rising F=1.13, P=0.35). The impedance at equilibrium was averaged over ten subjects for each of the four trial conditions. We draw the important conclusion that reduced mean sway size was not caused by increased ankle impedance. If anything, there was a slight tendency for higher impedance when standing easy as opposed to standing still (2 v 1 and 4 v 3). It is clear from EMG recordings that, as a result of the trial conditions, there were no significant changes in mean activation levels of tibialis anterior, (two way ANOVA, N=40, F=1.61, P=0.21) or soleus (two way ANOVA, N=40, F=1.27, P=0.30) (panel C). This implies that co-contraction of muscles about the ankle joint was not a factor in bringing about changes in sway size.

Grouping the equilibrium line-crossings according to the duration of the sway they occur in shows that high frequency sways are associated with increased line-crossing ankle impedance (Figure 6A). However, investigation of the relationship between sway frequency and ankle impedance shows that the intent of the subject or the availability of visual feedback makes no significant difference to the ankle impedance at all frequencies (Fig. 6C). Five sway frequency categories were used. For each frequency category and for each trial record, the line-crossings were averaged and the rising and falling line crossing impedances calculated. These impedances were then averaged over ten subjects for each of the four trial conditions (panel B). Panel A shows that the ankle impedances for each of the four trial conditions are very similar across the frequency range of 0.1 to 1.5 Hz. As in figure 5, there is a slight tendency for the 'stand still' impedances to be lower than the "stand easy' impedances. For the frequency range 0.1 - 0.7 Hz, the ankle impedance was relatively frequency insensitive. This implies that any change in frequency at the low end of the range will make little difference to the ankle impedance. These ankle impedances are always higher than the pendulum impedance which is also shown. Panel B shows that for each frequency category, and for both the rising and falling line-crossing impedances, there are no significant changes caused by the four trial conditions.

Due to the inertia of the pendulum, most of the pendulum sways occur in the 0.1 - 0.7 Hz range as seen in Figure 6C. Thus we cannot say what happens to ankle impedance during sways of higher frequency and our conclusions are concerned with low frequency sway control. In any case, filtering of our data would have precluded observations at frequencies greater than 3 Hz. The subject's intent and use of visual feedback makes little difference to the frequency distribution of sways as well as no significant difference to the mean sway frequency (Fig 4C). For each frequency category there is no significant change in the number of sways except the 0.4 Hz category. The intention to 'stand still' using visual feedback results in slightly more sways in the 0.2 Hz category at the expense of the 0.4 Hz category. This slight non-significant reduction in sway frequency when 'standing still' would be consistent with the slight non-significant reduction in ankle impedance.

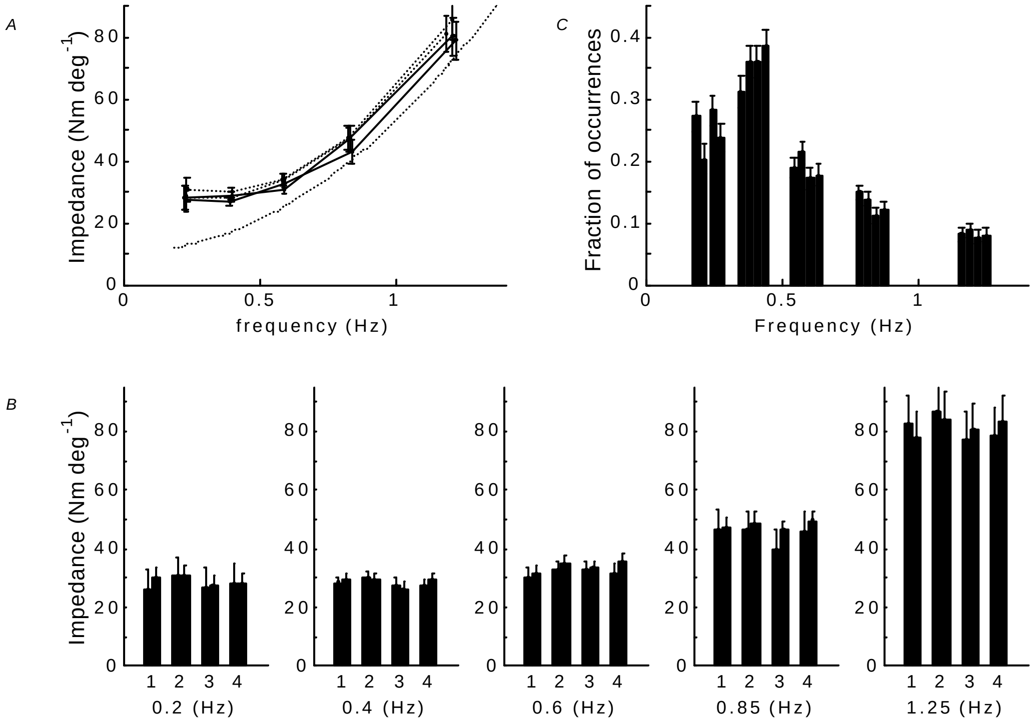

Figure 6. Relationship between ankle impedance and frequency.
The pendulum falling and pendulum rising, positive gradient, line-crossing equilibria were separately grouped into five frequency bins of 0.1-0.3 Hz, 0.3-0.5 Hz, 0.5-0.7 Hz, 0.7-1.0 Hz and 1.0-1.5 Hz. The frequency for each line-crossing was calculated from the duration of the associated sway. For each frequency bin for each trial, the mean falling and rising line-crossing impedances, and the mean frequency associated with the line-crossings, were calculated. The impedances were averaged over ten subjects for each of the trial conditions.

Panel ***A*** shows the mean line-crossing impedance for each of the four trial conditions plotted against mean binned frequency. (For this plot the rising and falling impedances have been combined and the points have been plotted at the mean frequencies rather than the central bin frequencies). The stand still conditions are plotted as solid lines and the stand easy conditions are plotted as dashed lines. The lowest dashed line is the load impedance. The load impedance was calculated using the formula $Z=abs(I*(jw)^2 -K_{tt} + b*jw)$ where I is the pendulum moment of inertia, K_{tt} is the load stiffness, b is the viscous drag of the pendulum, w is the angular frequency and j is the square root of -1 (Schwarzenbach & Gill, 1992). For each frequency bin and for each trial condition, panel ***B*** shows the mean, positive gradient, line crossing impedance (pendulum falling - left bar, rising - right bar). (Two way ANOVA, N=40, P= 0.80, 0.20, 0.54, 0.42, 0.48 for the falling impedances in order of increasing bin frequency and P = 0.52, 0.2, 0.1, 0.51, 0.75 for the rising impedances in order of increasing bin frequency). Panel ***C*** shows the fraction of occurrences populating each bin for each trial condition. A group of four trial conditions are shown (order 1, 2, 3, 4 from left to right) with the group centred at the mean frequency for each bin. (Two way ANOVA, N=80, P= 0.11, 0.0004, 0.24, 0.13, 0.81 for the five bins in order of increasing frequency). For all three panels, the error bars show the 95% simultaneous confidence intervals in the mean values.

The results in Figure 7 were generated using the model described in the Appendix. They reveal how independent changes in ankle stiffness, ankle viscosity and noise power would be reflected in our measures of 'line crossing' impedance, sway size and sway frequency. Figure 7A shows that, in conjunction with an appropriate noise source, typical values of ankle stiffness and viscosity for A/P standing sway (Winter et al. 1998) can produce a pattern similar to that seen in Figure 5, except that the gradient increases at equilibrium (maximum velocity) due to linear viscosity in the model. Figure 7B compared with 7A shows that a reduction in stiffness causes a decrease in sway frequency, an increase in sway size, a decrease in 'line crossing impedance' and a more apparent viscous increase in gradient at equilibrium. Figure 7C compared with 7A shows that decreasing the viscosity produces an increase in sway frequency, an increase in sway size, a decrease in 'line crossing impedance'. Figure 7D when compared to 7A shows that a four-fold increase in torque noise power produces no change in sway frequency and doubles the sway size without changing the 'line crossing impedance'. From the three factors which affect sway size, (stiffness, viscosity and noise), a change in noise power, i.e. torque error, gives the most comparable results to our sway size, sway frequency and ankle impedance data of figures 4 & 5.

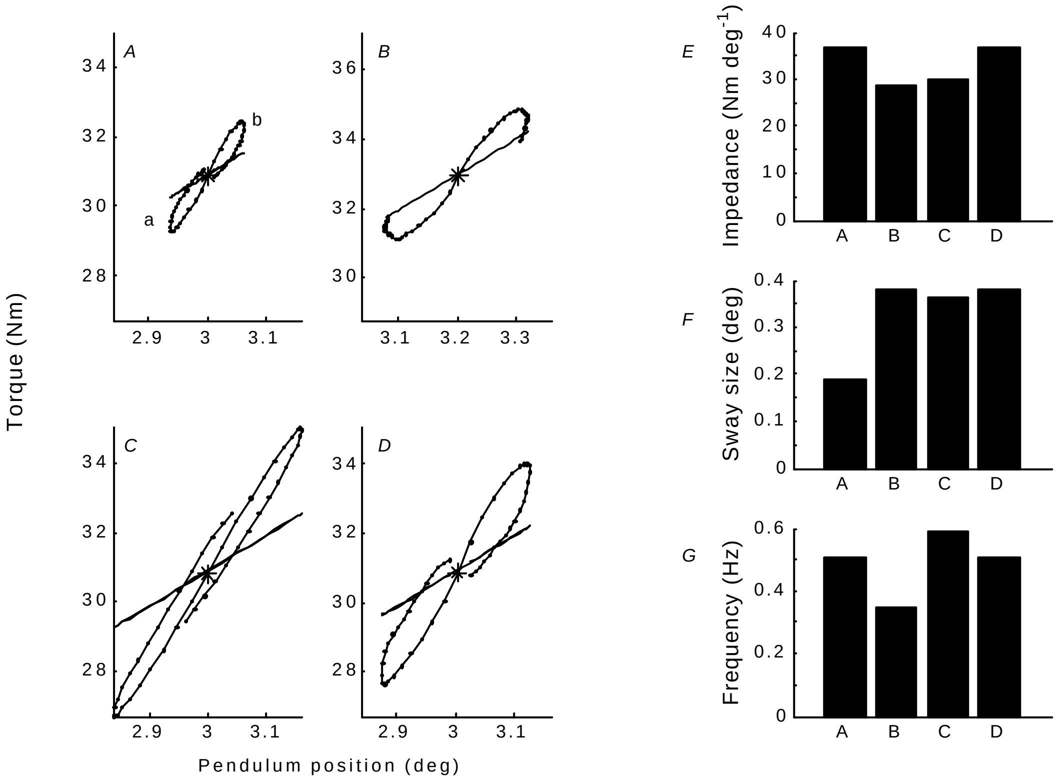

Figure 7. Effect of ankle stiffness, viscosity and torque noise on sway.

These results were generated using the model described in the Appendix. Panels ***A*** to ***D*** show data averaged from 1 s before to 1 s after each falling, positive gradient, equilibrium line crossing. The same scaling is used for each graph. Data points are at 40 ms intervals and proceed from label 'a' to 'b'. The line of equilibrium (ignoring pendulum friction) is shown as a dashed line. The asterisk marks the point of equilibrium and maximum velocity.

Panel ***A*** has normal values for ankle stiffness and viscosity for A/P sway taken from Winter et al. (1998), (K=1440 Nm/rad = 25.1 Nm/deg, B=350 Nm s/rad = 6.11 Nm s/deg). Panel ***B*** shows reduced stiffness (K = 600 Nm/rad = 10.5 Nm/deg) compared to ***A***. Panel ***C*** shows reduced viscosity (B=100 Nm s/rad = 1.75 Nm s/deg) compared to ***A***. Panel ***D*** shows four times the torque noise power as ***A***. Panel ***E*** shows the measured 'line crossing gradients' for the simulated trials shown in panels ***A*** to ***D***. These line-crossing gradients are inevitably higher than the stiffness as the impedance includes a viscous and noise component. Panel ***F*** shows the mean sway size and panel ***G*** shows the mean sway frequency for the same simulated trials ***A*** to ***D***.

Discussion.

The questions we are pursuing are (i) whether or not subjects can systematically reduce the sway size of the pendulum by their intention and/or use of visual feedback, (ii) whether ankle impedance, ankle stiffness or ankle viscosity are the means used to change sway size or whether an alternative mechanism must be sought?

We found unambiguously that subjects could systematically control mean pendulum sway size (Fig 4). Giving full attention to minimising sway ('standing still') was effective whether or not visual feedback was available. When subjects were 'standing still', visual feedback enabled sway size to be reduced more than without visual feedback. This result agrees with that of Fitzpatrick et al. (1992b) although the effect of vision need not be to increase reflex gain in the manner they suggested. Visual feedback made less difference to sway size when subjects were giving minimal attention to their sway ('stand easy'). This result may be relevant to work which showed that visual input had little effect or contradictory effects on sway (Fitzpatrick et al. 1992b; Collins & De Luca, 1995; Winter et al. 1998). If subjects are not intending to stand still, vision may have little effect on standing sway.

How do subjects achieve this systematic reduction in sway size? A crucial element of the answer is that in minimising sway size, subjects did not change the mean sway frequency (Fig. 4C). The second element of the answer concerns the ankle impedance. The simple result here is that ankle impedance was not changed by the subject's intent or use of visual feedback (Fig. 5B). However, this result is complicated by the fact that the line-crossing impedance is a biased measure that is sensitive to frequency as shown in Figure 6.

Impedance increases with frequency although at the low end of the frequencies encountered the effect is slight. In comparing the degree of frequency bias between the four trial conditions we note (i) the mean sway frequency did not change and (ii) that subjects did not alter the distribution of frequencies with which line-crossings are associated (apart from the 0.4 Hz category where there is little sensitivity to frequency anyway) (Fig 6C). Thus the frequency bias did not change between the four trial conditions and it is reasonable to compare the averaged line-crossing impedances in the manner shown in Figure 5B. Moreover, by comparing the line-crossing ankle impedance between the trial conditions at each frequency category, we could see that at all sway frequencies the ankle impedance is not changed by the intent of the subject to minimise sway, or by the use of visual feedback from the oscilloscope (Fig. 6). This confirms that changing ankle impedance was not the means used to alter sway size.

In theory, alteration of stiffness, viscosity and torque noise are three methods that could be used to control sway size. Our modelling results illustrate the effect of independent changes in these parameters on sway size, line-crossing impedance and sway frequency (Fig. 7). (i) Control of sway size by stiffness alone requires an ankle stiffness that is greater than the toppling torque per unit angle of the pendulum (K_{tt}). As the ankle stiffness is increased the sway size would decrease. If there is no damping, then the sway size is proportional to $K_e^{-0.5}$ where K_e is effective stiffness (ankle stiffness minus K_{tt}). Sway frequency would increase with stiffness. In our experiments we found no change in sway frequency between trial conditions so control of sway size by stiffness alone is ruled out. (ii) By increasing ankle viscosity alone, the sway size could be reduced without changing ankle stiffness. If viscous ankle torques were significant, the impedance (gradient on an ankle torque v angle plot) would increase at equilibrium ('line crossing gradient' where the velocity is greatest) and sway frequency would decrease. In our results sway frequency did not change when sway size was systematically reduced so viscous changes alone are not the cause of reduction in sway size. Furthermore, the line-crossing gradient was not velocity dependent (Fig 5A). (iii) If ankle torque noise power alone were decreased, sway size would be decreased with no change in impedance or sway frequency which is what we observed.

What can we conclude regarding ankle stiffness and viscosity? Sway size was systematically reduced with no increase in average 'line crossing gradient'. The simplest explanation of this is that ankle stiffness and viscosity do not change. However, it is possible that a simultaneous decrease in ankle stiffness and increase in viscosity could have produced a reduction in sway size with no change in ankle impedance. We reject this possibility because a decrease in ankle stiffness and an increase in ankle viscosity would each have caused a decrease in sway frequency whereas no significant change in sway frequency was observed. A possible increase in ankle stiffness and decrease in viscosity is also rejected because again there was no change in sway frequency. Moreover, these possibilities are unlikely since one would expect position gain and velocity gain to vary together. Changes in noise power do affect changes in sway size, but do not affect the sway frequency or the line crossing impedance. For this reason we suggest changes in torque noise are the most likely cause of changes in sway size.

How do our conclusions correspond with those of other authors? Our 'line crossing' measure of impedance relates the repeated changes in torque to the changes in position that occur during unperturbed balancing of the pendulum. These changes in torque are taken to be the result of mechanical, reflex and feedforward (predictive) components though the extent of each component is unknown. Our results apparently differ from Fitzpatrick et al. (1992b) who claimed that task dependent and sensory dependant increases in ankle (reflex) stiffness changed the amount of sway in response to perturbation. Fitzpatrick et al.'s measure of stiffness relates the average change in ankle position to the average change in ankle torque induced by a slow perturbation. The ankle mechanisms used to regulate balance were averaged out by their method of analysis and the position maintaining stiffness remained.

Since subjects maintain equilibrium by balancing both the gravitational torque and the perturbation torque, the stiffness measured by Fitzpatrick et al. (K_F) depends on the perturbation torque (ΔT_p), the toppling torque per unit angle of the pendulum (K_{tt}) and the angle through which the subjects are deflected (DA) such that $K_F = DT_p/DA + K_{tt}$ (M. Lakie, unpublished observations). The stiffness K_F describes the resistance to perturbation but gives no information on the mechanism used to produce that resistance. It does not discriminate between mechanisms based on stiffness control, viscous control, predictive momentum absorption or noise reduction. A predictive, active adjustment of torque to reduce deflection can be achieved without any increase in mechanical stiffness, reflex stiffness or line-crossing impedance. Therefore it is potentially misleading to interpret the reduced deflection under slow perturbation as an increased ankle stiffness. The stiffness measured by Fitzpatrick et al describes the end result of the processes resisting a change in position, though it does not describe the ankle mechanisms used to resist the change in position. If the source of ankle impedance is mechanical, the ankles have one impedance only and our results are incompatible with those of Fitzpatrick et al. However, if the nervous system is regulating one impedance associated with balance and a different impedance or gain associated with position maintenance, then our procedure provides a measure of the former and the procedure of Fitzpatrick et al may provide a measure of the latter. We agree with Fitzpatrick et al. that changes in sway size were not caused by alterations in muscle co-contraction (Fig. 5C).

Our results differ from the conclusions of Winter et al. (1998) who produced evidence that M/L (medial/lateral) sway in standing was inversely proportional to the square root of the ankle stiffness. However, it is clear from Winter et al.'s own data that there was substantial damping present. Using their typical A/P (anterior/posterior) measurements of stiffness, viscosity and inertia we calculate a damping ratio of $B/\sqrt{4IK_e}=0.76$ and successive sway amplitude ratio of $A_{n+1}/A_n=0.0006$, ($K_e=850$ Nm/rad, $B=350$ Nm s/rad, $I=62$ kg m^2 (Winter et al. 1998; Kreyszig, 1999)). This calculation suggests that sagittal oscillations are not "severely underdamped approaching the undamped condition" as Winter et al. claim. Rather, this measurement agrees with our findings that preceding and subsequent oscillations have small or negligible average velocity (Fig 5A especially 1, 3). Winter et al.'s data and our graphs indicate that damping of sway is highly effective. We think this degree of damping is unlikely to be mechanical/reflex in origin. The shape of our graphs does not indicate a simple damper where viscous torque is proportional to velocity. Also for small oscillations reflexly active cat soleus has a small damping ratio ($\xi < 0.1$) (Lin & Rymer, 2000). Damping produced by an active, predictive modulation of torque would seem more likely.

Our data are consistent with the feedforward modulation of ankle torque as shown by Gatev et al. (1999). Our graphs in Figure 5A would illustrate their idea that ankle torque changes with angle in a spring like manner though the muscle activity is modulated predictively. Our data do not support their hypothesis that ankle stiffness is controlled centrally so as to reduce sway size.

Since changes in sway size are not produced by changes in impedance, stiffness or viscosity (Figs 5, 6, 7) an alternative mechanism must be sought. A predictive method with momentum absorption is plausible (Morasso & Schieppati, 1999) and feedforward control is likely (Fitzpatrick et al. 1996; Gatev et al. 1999). Below we sketch a broad outline of the processes involved in controlling the pendulum.

Balance of the inverted pendulum is unstable and uncorrected errors in ankle torque grow until a sizeable sway results. In effect the pendulum is either dropped or thrown by too little or too much ankle torque. A planned impulsive pattern of torque is then used in an attempt to arrest the motion of the pendulum. The pendulum is caught. The torque error when the pendulum is caught causes the next sway. The drop and catch or throw and catch process repeats over and over. The equilibrium position of the pendulum is not fixed (unlike stiffness control) and each drop (throw) and catch results in a change in position of equilibrium. This drop and catch process appears more complex than the simple regulation of a gain (or impedance). We suggest that mechanical stiffness may provide some stabilisation, that mechanical/reflex viscosity is rather insignificant and that torque is actively controlled to achieve final control.

Reduction of sway depends on the following processes: (1) registering quickly and accurately when position has changed, and velocity and acceleration have increased; (2) judging torque impulses accurately to arrest the motion and return to balance; (3) accurately maintaining the torque close to that required for balance. These processes require fine control of muscular effort. Any random noise in the nervous activation of the muscle will tend to a loss of balance and an increase in sway. The accuracy or lack of noise with which torque can be delivered will provide a limit on reducing sway size. The sensory threshold for sway detection may also provide a limit on reducing sway size. We agree with Fitzpatrick *et al.* (1992) that the task does not require complex volitional modulation of muscle activity since the task can be performed quite easily while the subject is distracted or engaged in conversation. In seeking to "stand still" rather than "stand easy", the ratio of active to passive processes may be increased.

Our results are for balancing an inverted pendulum which is not the same thing as standing, yet we have compared our results with those concerning quiet standing. In standing the effect of multi-segmentation is to reduce sway (Fitzpatrick et al. 1994b). The involvement of the vestibular apparatus in the fine regulation of quiet standing is not thought to be significant (Fitzpatrick et al. 1994b). Since pendulum sways occur at low frequencies our method necessarily gives information about low frequency sway control. Our investigation shows that ankle impedance is not increased to reduce sway size. We deduce that ankle stiffness and viscosity are also not changed to control sway size. A reduction in torque noise via an active, predictive process, which provides damping, is more likely to be the cause of reduction in sway size.

Appendix

A model of the pendulum and the ankle torque applied to the pendulum has been constructed to assess the effect of changes in ankle stiffness, ankle viscosity and ankle torque noise on sway size, sway frequency and our measure of ankle impedance. The pendulum is modelled as a second order differential equation.

$$I\ d^2q/dt^2 + b\ dq/dt\ - K_{tt}\ \sin q = T_{ankle}$$

where I is the inertia, b is the viscous damping and K_{tt} is the gravitational toppling torque per unit angle of the pendulum. Values used were those for our own pendulum. I = 62.6 kgm^2, b = 0.061 Nm s/deg, K_{tt}=10.3 Nm/deg. q is the pendulum angle and T_{ankle} is the ankle torque applied to the pendulum.

The ankle torque generated by the subject was modelled as having a stiffness, viscous and noise component.

$$-T_{ankle} = K\ (q-q_0) + B\ dq/dt + w$$

where K and B are the ankle stiffness and viscosity respectively. q_0 is the offset angle for the ankle stiffness and w is the ankle torque noise. Typical values of K and B for A/P standing sway are taken from Winter et al. (1998). $K = 850 + K_{tt}$ = 1440 Nm/rad = 25.1 Nm/deg, B = 350 Nm s/rad = 6.11 Nm s/deg. Simulink (The MathWorks Inc.) was used to solve the equations and the model is shown in Figure 8. q_0 was chosen to give a mean equilibrium position of 3 degrees. A low frequency or band limited white noise with a sizeable (few Nm) random step change in torque every 0.4 s produces records similar to real data. Band limited white noise was used in conjunction with a first order low pass filter, $1/(1+ts)$ to reduce the step like nature of the noise. A noise sampling time t_{sample} = 0.4s, a noise power of 6.4, and a filter time constant $t = 0.5$ s were used. We investigated the effect of varying the ankle stiffness K, ankle viscosity B, and torque noise power. Simulation data were subjected to the same analysis procedures as real data. Selected simulation results are shown in figure 7.

This model is essentially that described by Winter et al. (1998), although we have given more prominence to the effect of noise. The ankle stiffness and viscosity could represent mechanical values as suggested by Winter. The model could not represent reflex values of ankle stiffness and viscosity because there is no incorporation of a time delay, though the model could represent feedforward values in which it is assumed that time delays have been perfectly eliminated.

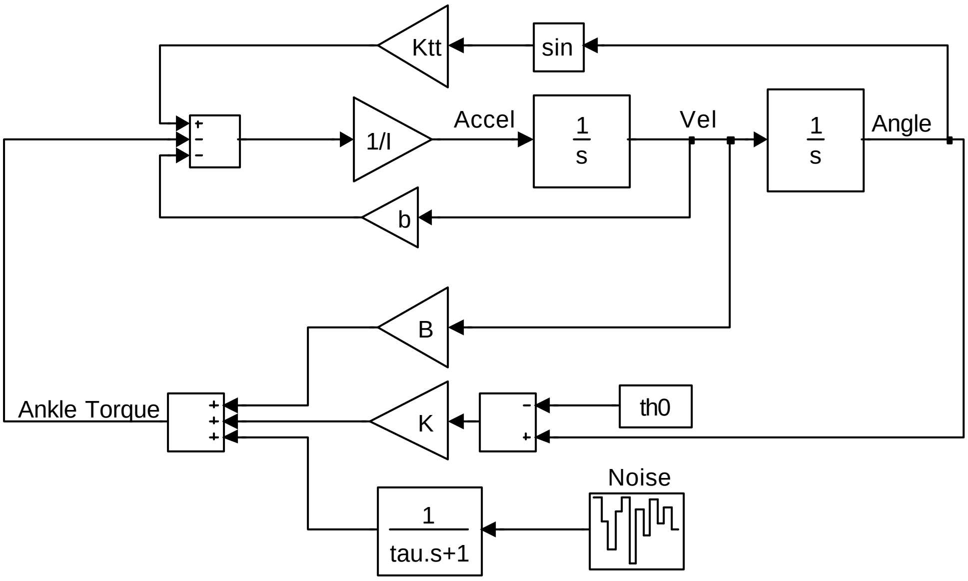

Figure 8. Model of inverted pendulum and ankle torque.
The Simulink model used to represent the pendulum and the ankle torque is shown. I is the pendulum moment of inertia, b is the viscous damping and K_{tt} is the gravitational toppling torque per unit angle of the pendulum. Values used were those for our own pendulum. I = 62.6 kgm^2, b = 0.061 Nm s/deg, K_{tt}=10.3 Nm/deg. K and B are the ankle stiffness and viscosity respectively. *th0* is the offset angle for the ankle stiffness. Typical values of K and B for A/P standing sway are taken from Winter et al. (1998). $K = 850 + K_{tt}$ = 1440 Nm/rad = 25.1 Nm/deg, B = 350 Nm s/rad = 6.11 Nm s/deg. Band limited white noise was used in conjunction with a first order low pass filter, $1/(1+ts)$ to reduce the step like nature of the noise. A noise sampling time t_{sample} = 0.4 s, a noise power of 6.4, and a filter time constant t = 0.5 s were used.

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CHAPTER 3.

HUMAN BALANCING OF AN INVERTED PENDULUM: POSITION CONTROL BY SMALL, BALLISTIC-LIKE, THROW AND CATCH MOVEMENTS.

Summary.

In standing, there are small sways of the body. Our interest is to use an artificial task to illuminate the mechanisms underlying the sways and to account for changes in their size. Using the ankle musculature, subjects balanced a large inverted pendulum. The equilibrium of the pendulum is unstable and quasi-regular sway was observed like that in quiet standing. By giving full attention to minimising sway subjects could systematically reduce pendulum movement. The pendulum position, the torque generated at each ankle and the soleus and tibialis anterior EMG were recorded. Explanations about how the human inverted pendulum is balanced usually ignore the fact that balance is maintained over a range of angles and not just at one angle. Any resting equilibrium position of the pendulum is unstable and in practice temporary; movement to a different resting equilibrium position can only be accomplished by a biphasic "throw and catch" pattern of torque and not by an elastic mechanism. Results showed that balance was achieved by the constant repetition of a neurally generated ballistic-like biphasic pattern of torque which can both control position and sway size. A decomposition technique revealed that there was a substantial contribution to changes in torque from intrinsic mechanical ankle stiffness; however by itself this was insufficient to maintain balance or to control position. Minimisation of sway size was caused by improvement in the accuracy of the anticipatory torque impulses. We hypothesise

that examination of centre of mass and centre of pressure data for quiet standing will duplicate these results.

Introduction.

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 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 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; Alsonso-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 here the essential and additional features only will be reported.

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). The subject stood on two footplates with their ankles positioned to be co-axial with the axis of rotation of an inverted pendulum. The 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 toppling torque per unit angle of the pendulum was measured to be 10.2 ± 0.4 Nm deg^{-1} and the moment of inertia was 62.6 ±2 kg m^2 (means ± SD). 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 EMG 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 encapsulated preamplifiers. These signals containing the entire bandwidth were then amplified, passed through an analogue full-wave rectifier and 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 0.04V.

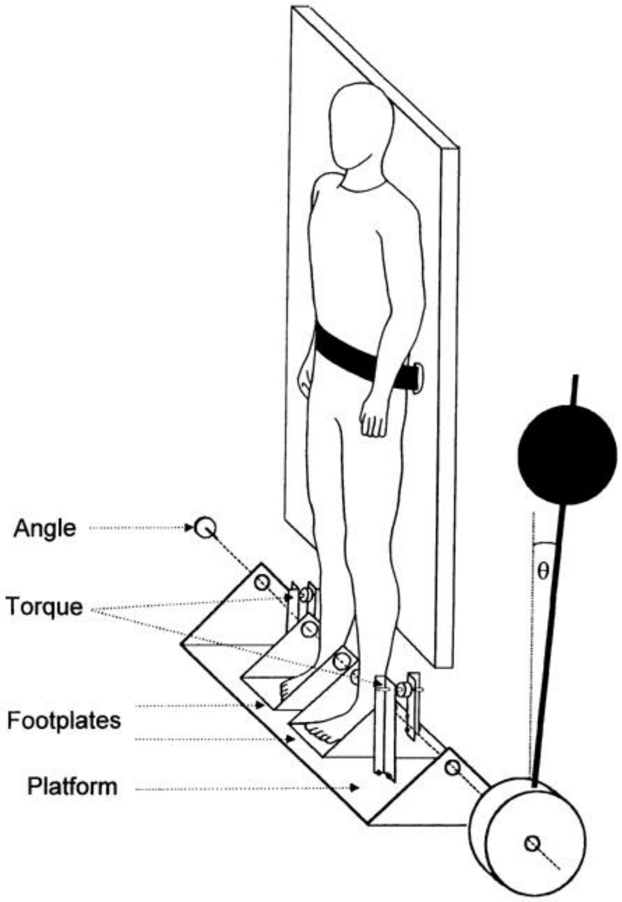

Figure 1. Inverted Pendulum Apparatus.
Subjects balanced a backward leaning, real inverted pendulum of mass and inertia equivalent to a medium sized woman. The subjects were unable to sway since they were strapped round the pelvis to a fixed vertical support. The rotation of the pendulum, platform and footplates was co-axial with the subject's ankles. Force exerted by the subject's ankle musculature onto each footplate was transmitted by horizontally mounted load cells. A contactless precision potentiometer measured sway of the pendulum.

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 the pendulum 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 degrees from the vertical so as to approximate standing sway. Both instructions were carried out once using visual feedback which was provided using an oscilloscope and once without using visual feedback. Thus there were four separate trial conditions which were:

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 200s.

When subjects were "standing still" they were encouraged to give their full attention to the oscilloscope when that was turned on 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 paper we are studying the effect of subject intention on pendulum movement and are 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 is 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 is 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 is 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 Figure 2.

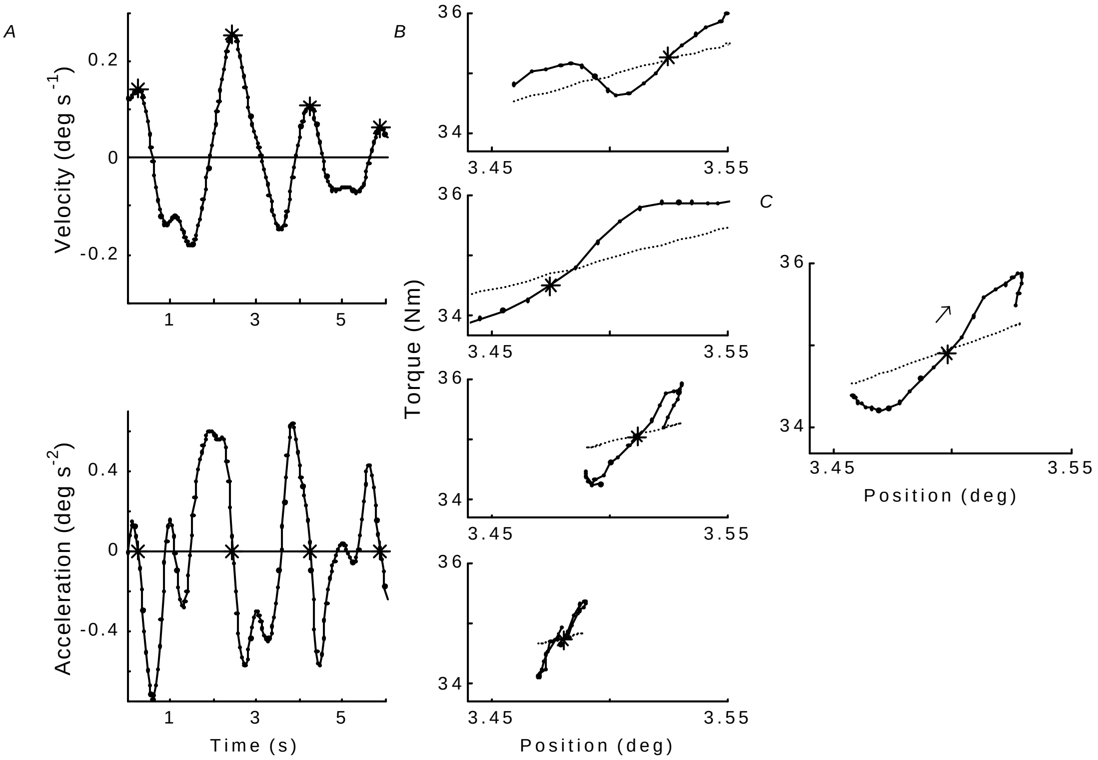

Figure 2. Illustration of the line-crossing averaging process.
Panel ***A*** shows a six second record of angular velocity and angular acceleration against time for a representative subject. Equilibrium times are identified by interpolating between the pairs of acceleration data points that cross zero. From these equilibrium times are selected those that occur while the acceleration is passing from positive to negative and while the velocity is positive (i.e. the pendulum is falling). These equilibrium times are shown as an asterisk. Ankle torque and pendulum position records are sampled at 0.04 s intervals before and after these selected equilibrium times. The four selected equilibrium times in panel ***A*** are shown in panel ***B***, together with ± 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 panel ***C***. In a 200 s trial, over 100 examples would be averaged. The rising, positive-gradient line-crossings are selected and averaged in an analogous manner.

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, 2.0 deg s^{-1} and these were chosen to equalise the numbers grouped in each bin for each trial condition as much 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, 4.0 deg s^{-2} 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 eighty values from ten 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" v "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 the 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 degrees for 20 seconds. 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 Nm 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 shows the same general characteristics with all subjects under all conditions. A representative example is shown in Figure 3A. The pendulum is not confined to one angle and there is no one position of equilibrium. Rather the pendulum makes small sways to and fro of irregular duration, size and speed. Movement of the pendulum is controlled by an ankle torque that always keeps close to that required for balance and which attains equilibrium transiently every time the line of equilibrium is crossed. Most of these line-crossings are spring like (torque increases as angle increases) with a positive gradient. Less commonly negative gradient line-crossings can also be seen (X).

When subjects gave their full attention to keeping the pendulum still as opposed to giving minimal attention to keeping the pendulum 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 (1 *vs.* 3) and when it was not (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 mirrors the mean sway size for all four conditions.

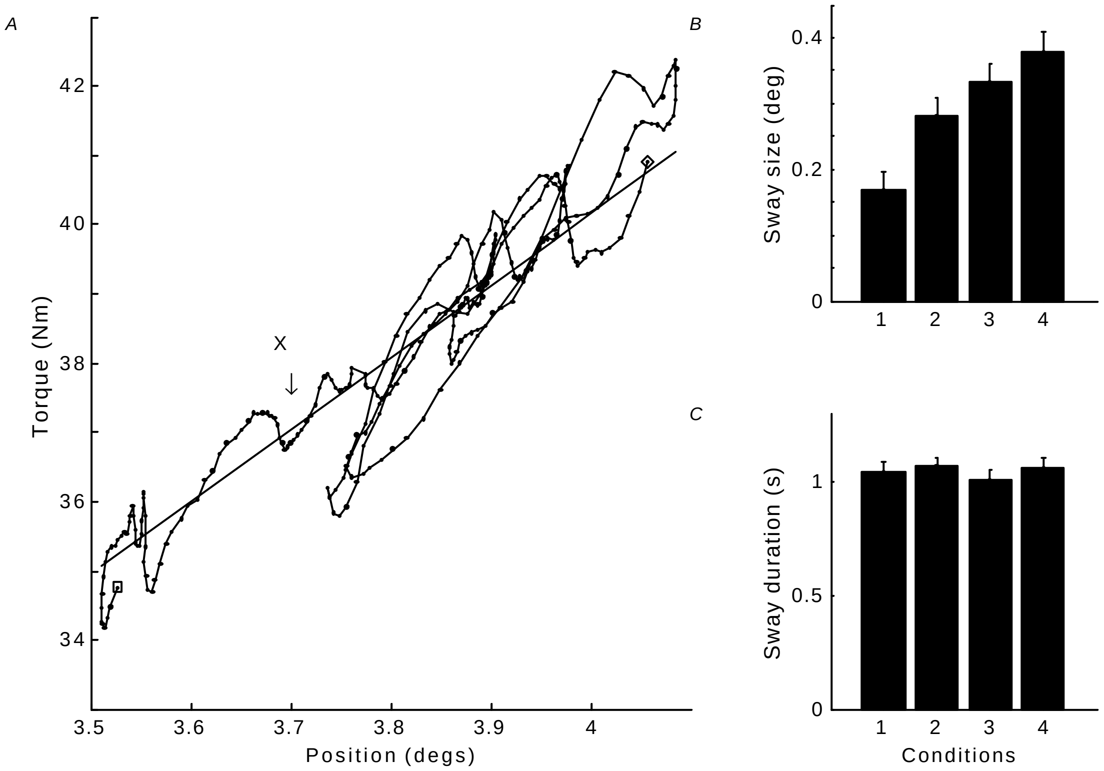

Figure 3. Pendulum sway.

Panel ***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, panel ***B*** shows the mean sway size and panel ***C*** shows the mean sway duration. For both panels, values were averaged over ten subjects for each of the four trial conditions. Error bars show 95% simultaneous confidence intervals for the mean values. As described in the methods section, a sway was 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. 4. Stand easy with no visual feedback.

In balancing the pendulum, an underlying process is repeated over and over. Figure 4 shows the entry into and departure from equilibrium averaged for all occurrences for all ten subjects. The pendulum falling, positive gradient cases have been selected under still and not still conditions with visual feedback available (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 sways from one reversal point (a) to the next (b). From (a) there is initially an increase in torque error (difference between torque applied and that required for equilibrium), (the "drop"), followed by a decrease in torque error leading to equilibrium and the maximum speed of the pendulum (starred). This is followed by an opposite increase then decrease in torque error which decelerates 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. In Figure 4, panels A and B, the reversal point at the start of the sway where the pendulum stops moving to a smaller angle, and starts moving to a larger angle is shown by (a). The data points proceed at 40 ms intervals, in the direction of the arrow, through the equilibrium point shown by the star, and towards the change of direction at the end of the sway shown by (b). The decrease in torque immediately after the sway begins at (a) is not caused mechanically/elastically because the muscle-tendon is being stretched at this point (dorsiflexion). Neither is mechanical viscosity the cause because the speed is increasing. In any case the mechanical and reflex viscosity (~0.02 and ~0.07 Nm s deg^{-1} respectively per leg operating at 15 Nm, (Mirbagheri et al. 2000)) are too small to produce major changes in torque for these averaged sways which only reach maximum speeds of 0.2 and 0.5 deg s^{-1} respectively. Thus the decrease in torque must be caused by neural modulation and this is confirmed in Fig. 4B where it can be seen that the soleus EMG is decreasing in the 140 ms preceding the reversal point at (a). The nervous system does not anticipate the initiation of the sway as the soleus EMG only rises as the sway begins at (a) leading to a delayed increase in ankle torque that will catch the falling pendulum.

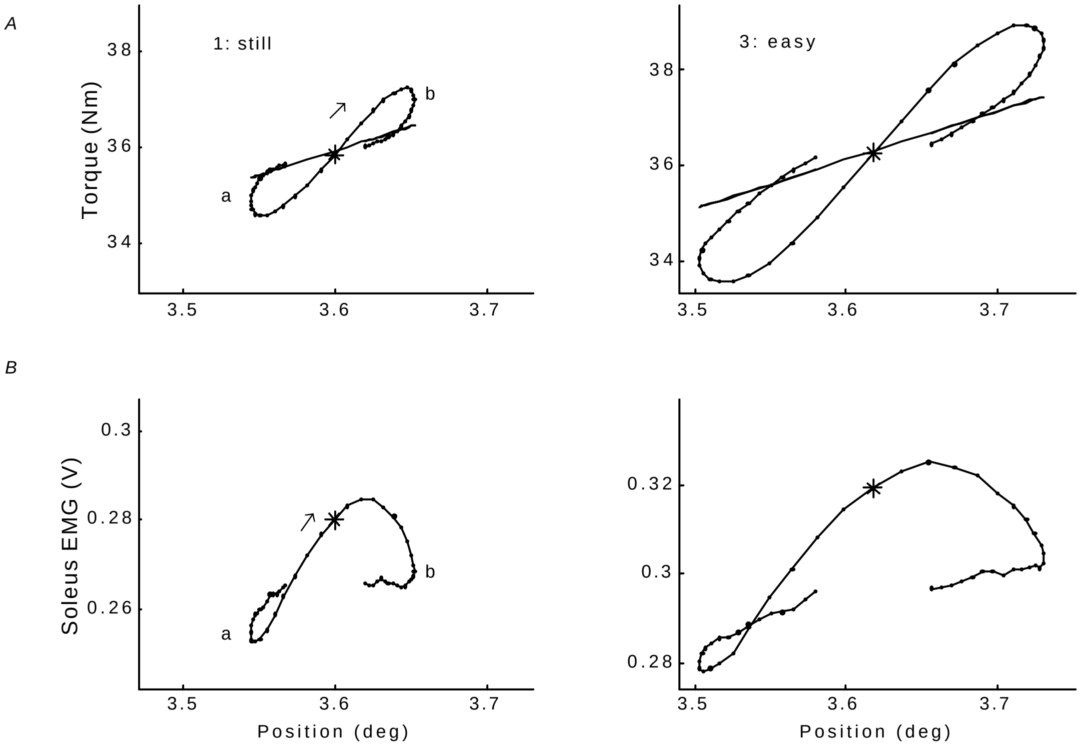

Figure 4. Effect of intention on averaged spring like line-crossings.

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 ten subjects in the "still" (1) and "easy" (3) conditions while visual feedback was available. Panel ***A*** shows combined torque from both legs *vs.* pendulum position. Panel ***B*** shows 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 panel ***A***. The asterisk marks the instant of equilibrium and maximum velocity.

Similarly, the decrease in torque after equilibrium (star) and before the reversal point (b) is also not mechanically elastic or mechanically or reflexly viscous in origin for the same reason as above. This decrease in torque corresponds 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 anticipates 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 with soleus. But 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 (Figure 8A).

One can clearly see that the sway size between (a) and (b) is less for the still condition than the easy condition. There is no change in line-crossing gradient which indicates no change in the operational stiffness or viscosity of the ankle mechanisms (Loram et al. 2001). Since each dot occurs at 40 ms intervals it is 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 minimise the absolute acceleration of the pendulum. It can also be seen that at the reversal point (b) there is less torque error in the still condition than the easy condition and this would alter 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 a different distribution 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, Figure 5C shows the actual changes in ankle torque for trial condition 3 averaged over all ten subjects (*c.f.* 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 Figure 5 panels A 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) is approached (the catch).

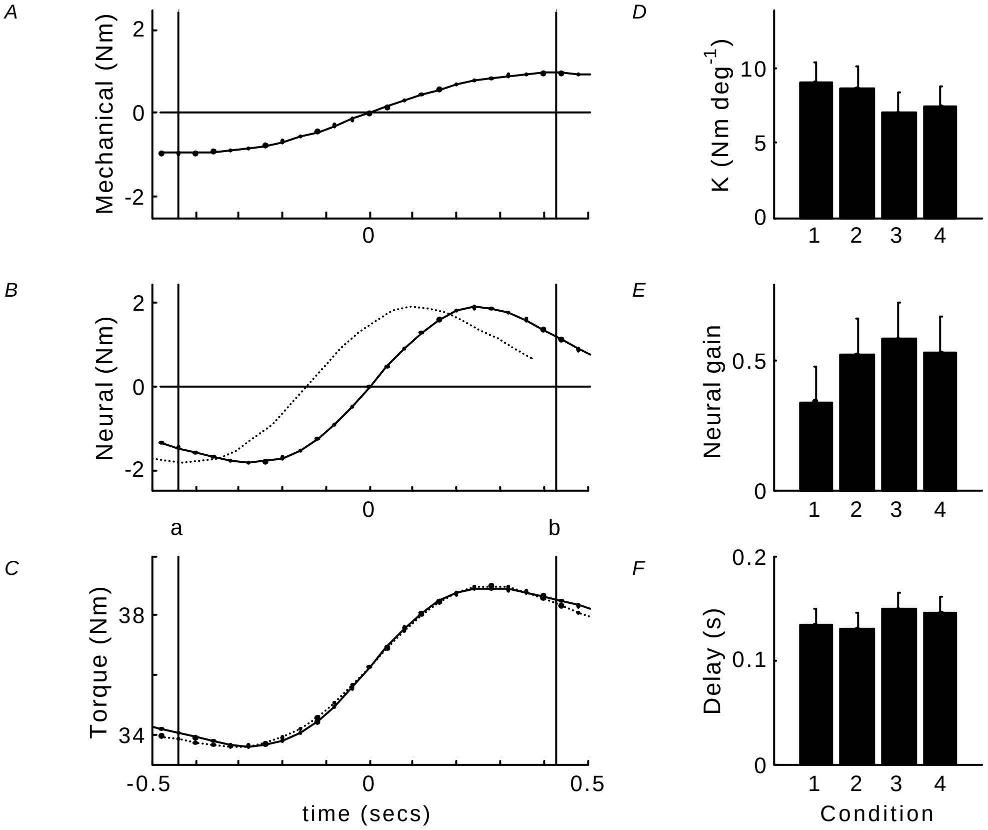

Figure 5. Decomposition of ankle torque according to our model.
The decomposition of ankle torque into intrinsic elastic and neurally modulated components is shown. The model is described in Appendix A. Panels ***A* - *C*** show illustrative data averaged from all subjects for falling, positive gradient equilibrium line-crossings in condition 3. Panel ***A*** shows the changes in torque arising from stretching and releasing of the elastic components. Panel ***B*** shows the variation in torque resulting from neural modulation (solid line) and the preceding variation in soleus EMG (dashed line). Panel ***C*** shows the actual variation in torque as well as the modelled variation (dashed line). Panels ***A***, ***B*** and ***C*** show the same position, EMG and torque data as Figure 4 (easy). Data points are at 40 ms intervals and the reversal points (a) and (b) corresponds 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, panel ***D*** shows the mean intrinsic mechanical stiffness, panel ***E*** shows the mean neural gain and panel ***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 (Nm V^{-1}). Parameter values were averaged over ten subjects for each of the four trial conditions. Error bars show 95% simultaneous confidence intervals for the mean values.

The average intrinsic mechanical stiffness of all subjects was ~8 ± 4 Nm deg^{-1} (± SD) and this value is just less than the gravitational toppling torque per unit angle of the pendulum (10.2 Nm deg^{-1}), (Fig. 5D). The averaged neural gain of all subjects was ~0.5 ±0.3 (±SD) which is expressed relative to the isometric gain in Nm V^{-1} 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 (±SD), (Fig.5F). For the mechanical stiffness, the neural gain and the electromechanical delay, there were no significant differences resulting from the intention of the subject (two way ANOVA, N=80, F=2.8, P=0.1; F=0.8, P=0.4; F=3.6, P=0.06 respectively).

When subjects were minimising pendulum movement they minimised the torque error as the pendulum was brought to rest at the end of a "catch". Fig.6A shows that at all velocities, apart from the lowest, the torque error when the pendulum is first brought to rest following the line-crossing is 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.

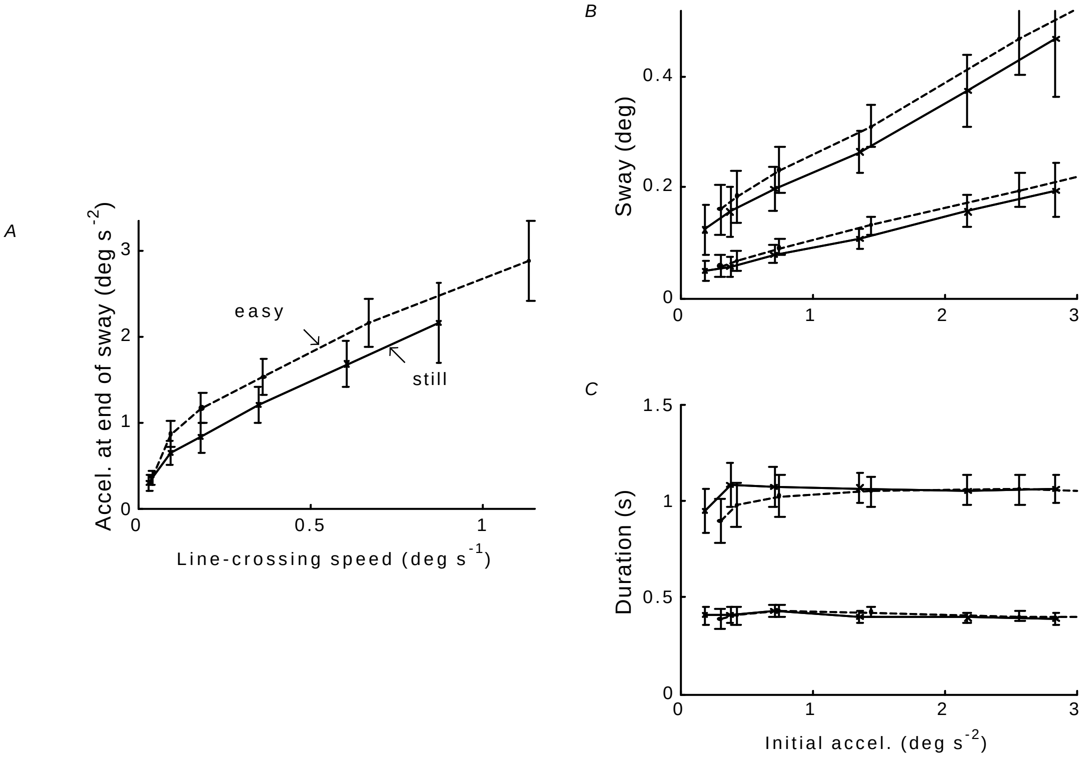

Figure 6. Effect of intention on binned sways.
Panel ***A*** shows the effect of intention on sways sampled and grouped according to their velocity at the first positive gradient line-crossing. ***A*** shows the acceleration at the end of the sway. Panels ***B*** and ***C*** show the effect of intention on sways sampled and grouped according to their initial acceleration. ***B*** shows the sway size to the first positive-gradient line-crossing equilibrium and to the reversal point at the end of the sway. ***C*** shows the duration to the first line-crossing and to the end of the sway. For all panels the "still" results (crosses on solid 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.

To assess the benefit of minimising the initial acceleration of a sway, we need 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) increases with the initial acceleration for both the “still” and “easy” conditions. For the three lowest acceleration bins the difference between the conditions is not significant though taking all five bins together the fact that the sway sizes are always less for the “still” condition is significant (N=5, P=1/2^5=0.03). This figure confirms that in minimising pendulum movement there is a benefit from minimising the initial acceleration of a sway. Fig.6C shows that for all initial accelerations the duration of a complete sway is virtually unchanged at 1s, and the duration till equilibrium is unchanged at 0.4s and the intention of a subject makes no difference to either of these times. Clearly, in the “still” condition there is an improvement in the efficacy but not the rapidity of the movement minimising process.

The size of a sway is clearly associated with the maximum speed of the pendulum at the spring like line-crossing in the middle of the sway and this relationship is 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 is eventually brought to rest and then there is 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 is significantly and substantially less in the “still” compared to the “easy” condition. The intention of the subject to minimise movement has great effect by minimising the 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 results purely from minimising the initial acceleration of the sway; and half also results from the intention of the subject during the whole course of the subsequent sway.

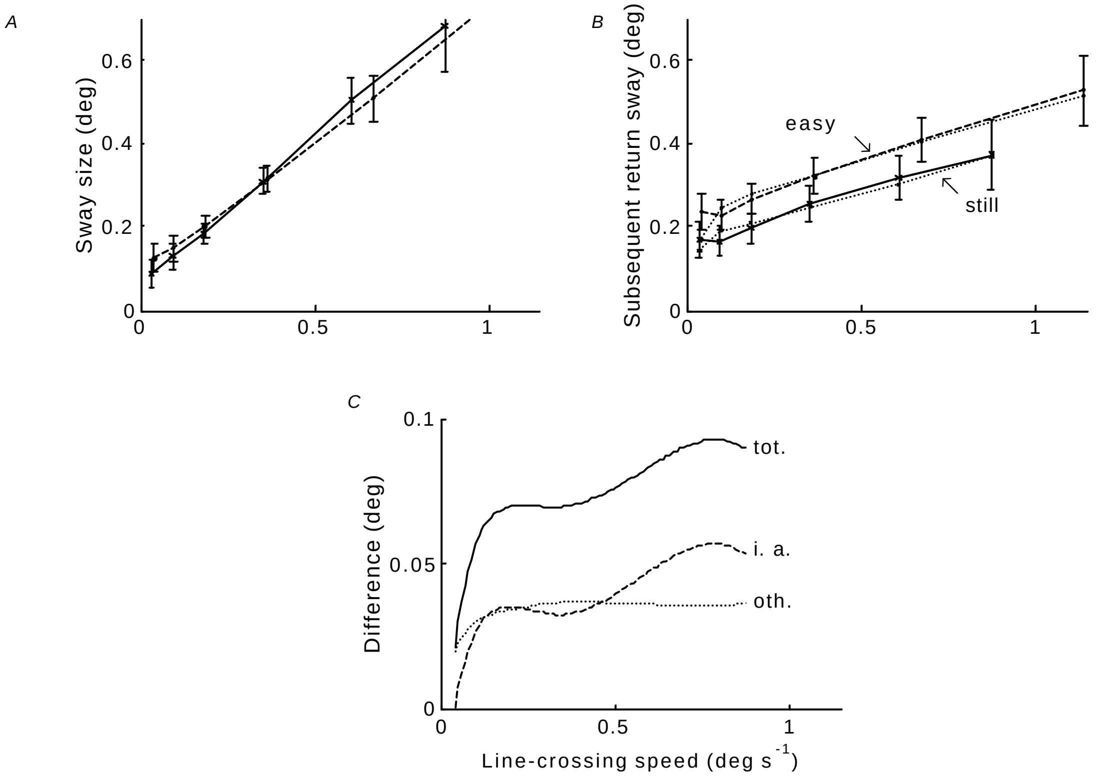

Figure 7. Effect of intention on the current sway and subsequent return sway. For every sway the pendulum starts from transient rest, passes a positive gradient line-crossing (a speed maximum) and ultimately comes to a reversal point where it changes direction. For each bin, panel ***A*** shows the mean 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. Panel ***B*** (solid 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 Figure 6 (A,B). The "still" results (crosses on solid 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.

Panel ***C.*** For a range of current line-crossing speeds the solid 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. 6 (A,B) not Fig. 7B.

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 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 Figure 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 was 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 - Figure 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, 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 Figure 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 result in a low frequency positional drift. This may accord with the knowledge that positional changes in quiet standing follow a third order model comprised of an underlying first order decay and a damped spring oscillator all buffeted by noise (Johansson et al. 1988; Kiemal et al. 2001).

(ii) What is the relationship between the intrinsic elastic and neural origins of changes in ankle torque? As seen in Figure 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 (Figure 5). The basis for this decomposition is given in Appendix A. On this interpretation, the mean intrinsic, mechanical stiffness for all subjects (~8 ± 4 Nm deg^{-1}, ± SD) is less than the minimum stiffness required for stability (the toppling torque per angle of the pendulum (10.2 Nm 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 Nm include (≈> 1.5Nm/deg (de Zee & Voigt, 2001)), (≈> 2.5Nm/deg (Hof, 1998)), (≈ 4 to 6 Nm/deg (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, ±SD) 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 ~140ms calculated for soleus from Fig. 2, Bawa and Stein (1976). The modulation of soleus EMG shown in Figure 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 and 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) 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). Purely of itself this mechanism accounts 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 Figure 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 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 (*c.f.* Figure 2, 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 (Figure 5).

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

$$T(t) = T(q(t), E(t - t))$$

We have eliminated velocity as a variable because the mechanical and reflex viscosity 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 twelve 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).

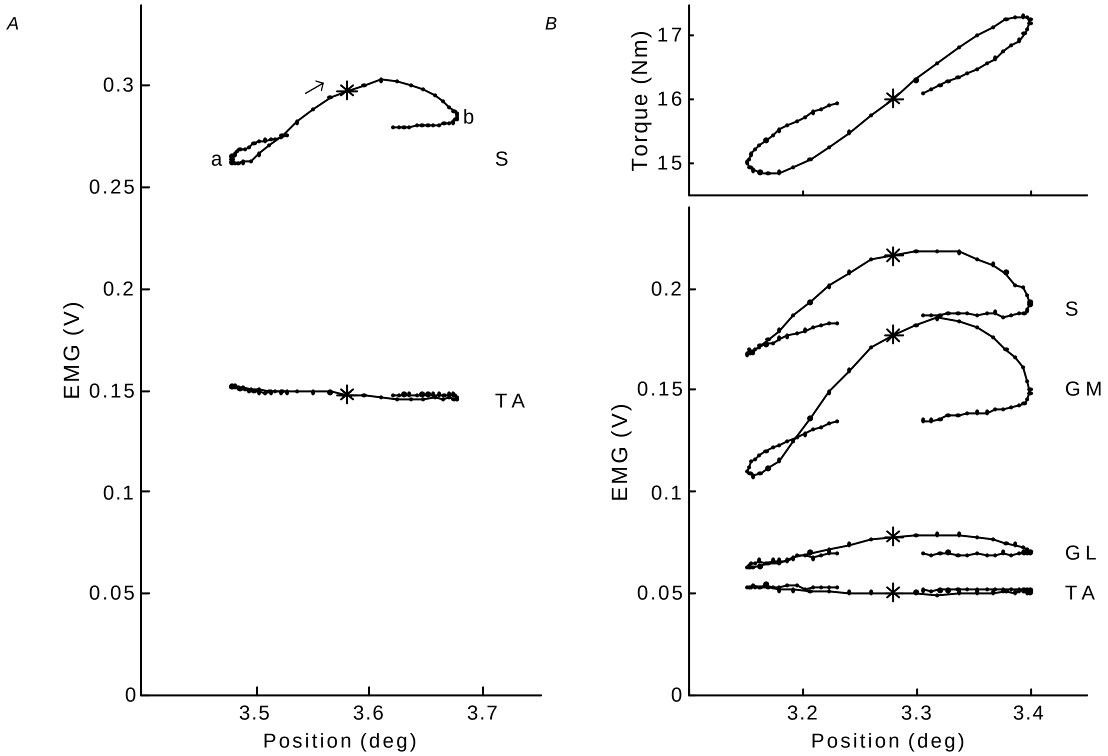

Figure 8. Comparison of soleus, tibialis anterior and gastrocnemius EMG.
For both panels averaged data are shown from 1 s before to 1 s after all positive gradient, equilibrium line-crossings while the pendulum was falling. Data points are at 40 ms intervals and proceed from reversal points (a) to (b) via the arrow. The asterisk marks the instant of equilibrium and maximum velocity. Panel ***A*** shows soleus (S) and tibialis anterior (TA) EMG from both legs *vs*. pendulum position. These data were averaged over all ten subjects and over all 40 trials including "still" and "easy" condition both with and without visual feedback. Panel ***B*** shows (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 to 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.

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).

$$\frac{dT}{dt} = \frac{\partial T}{\partial q}\Big|_E \cdot \frac{dq}{dt} + \frac{\partial T}{\partial E}\Big|_q \cdot \frac{dE}{dt}$$

The coefficients of the linear terms are intrinsic angular stiffness, $K = \frac{\partial T}{\partial q}\Big|_E$, (change in torque per unit angle at constant EMG), and isometric neural gain, $c = \frac{\partial T}{\partial E}\Big|_q$, (change in torque per unit EMG at constant angle).

Hence $$\frac{dT}{dt} = K\frac{dq}{dt} + c\frac{dE}{dt} \qquad (1)$$

We use (1) to predict the rate of change of torque (*dT/dt*) from the pendulum velocity (*dq/dt*) and the rate of change of soleus EMG (*dE/dt*) with an electromechanical 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 ± 3 - 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 are 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 t 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 $0 < K < 30$ Nm/deg, $0 < c < 800$ Nm/V, $0 < t < 0.4$ s. This was established by starting searches from a variety of locations including the eight vertices of the above 3-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 t from the 40 trials varied with respective standard deviations of ±4 Nm deg^{-1}, ±0.3 (relative to the isometric value in Nm V^{-1}) and ±40 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, ± 3 Nm deg^{-1}, ±0.09 and ± 20 ms for coefficients K, c and t 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 ($\%VAF$) by the model was calculated by

$$\%VAF = 100\left(1 - \frac{\sum^{N}(y - Y)^2}{\sum_{1}^{N} y^2}\right) \quad \text{where } N \text{ is the number of points.}$$

For all 80 results, the mean $\%VAF$ was 97%. Since this decomposition rests on assumptions, the results need to be 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 (Figure 9).

The equation of motion for the pendulum is

$$I\frac{d^2q}{dt^2} = mghq + T \qquad (1)$$

where I is the moment of inertia, q 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

$$t = 0,\ q = q_0,\ v = v_0,\ T = -mghq_0 .$$

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,

$$t = t_e,\ q = q_e,\ v = 0,\ T = -mghq_e .$$

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.

$$T = -mghq + T_I \quad (2) \qquad \text{where} \quad \int_{t=0}^{t=t_e} T_I\,dt = -Iv_0$$

Any function for T_I that satisfies the impulsive requirement will suffice, but for simplicity and realism we will choose $T_I = A\sin(\frac{pt}{t_e})$. Solution of the integral yields the value of the constant $A = -I\frac{v_0 p}{2t_e}$.

Substitution of *T* into the pendulum equation of motion (1) gives,

$$\frac{d^2q}{dt^2} = -\frac{v_0 p}{2t_e}\sin(\frac{pt}{t_e})$$

Double integration between $t=0$ and $t=t$ and gives

$$q = q_0 + \frac{v_0}{2}(t + \frac{t_e}{p}\sin(\frac{pt}{t_e}))$$

and substitution into the equation for subject torque (2) gives

$$-T = mghq_0 + mgh\frac{v_0}{2}(t + (\frac{t_e}{p} + \frac{I}{mgh}\frac{p}{t_e})\sin\frac{pt}{t_e})$$

where -*T* is plotted in Figure 9.

The temporal symmetry of the problem allows the full drop and catch to be calculated using the range $-t_e \le t \le t_e$ and the double drop and catch (Fig. 9D) to be calculated using $-t_e \le t \le 3t_e$.

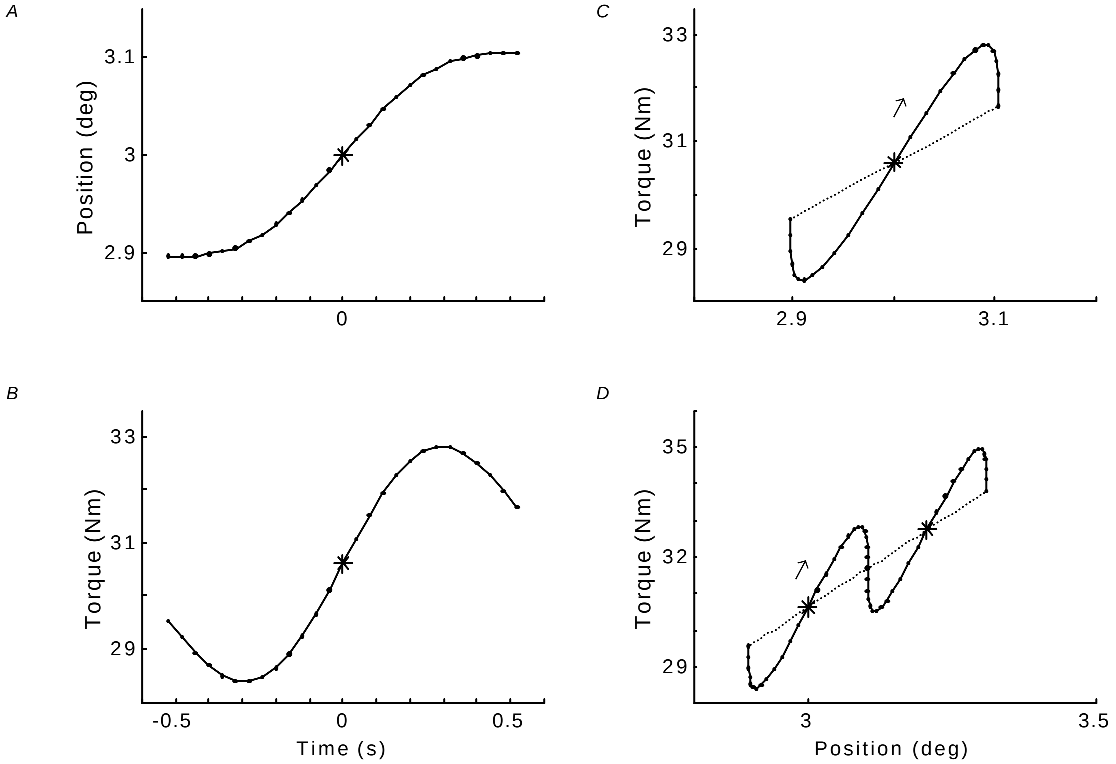

Figure 9. An ideal, perfect drop and catch pattern.
Panels ***A, B*** *&* ***C*** show a perfect biphasic drop and catch pattern that will move the pendulum from rest and equilibrium at one position to rest and equilibrium at a new position. ***A***. Pendulum position *vs*. time. ***B***. Torque *vs*. time. ***C***. Torque *vs*. position. Panel ***D*** shows 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.

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CHAPTER 4.

DEVELOPMENT OF THE ANKLE STIFFNESS MEASUREMENT APPARATUS.

This chapter presents a chronological description of the design and development of the ankle stiffness measuring apparatus. The finished product, the method and the results that were obtained from it are described in **Chapter 5** and elsewhere (Loram & Lakie, 2002a). Here I outline the initial idea, the problems that were envisaged and encountered and how those problems were dealt with.

Aim and purpose of the stiffness measurement technique.

Throughout my PhD project, I had available measurements of ankle torque, ankle angle and EMG signals from the triceps surae muscles and tibialis anterior. These measurements enabled me to observe the changes in ankle torque in relation to the changes in muscle activation and ankle angle. With this information I could relate the change in ankle torque to changes in ankle angle irrespective of whether those changes in torque result from neural modulation or stretching of the mechanical structures. On this view (not necessarily a correct view) there is one true input variable, ankle angle and other variables such as muscle activation and muscle length are internal system variables. I call the transfer function that relates changes in ankle torque to changes in ankle angle the operational ankle impedance to allow for the fact that the changes in torque are produced by a black box system over whose workings nothing is known. The method, results and conclusions of these measurements are presented in **Chapter 2** and elsewhere (Loram *et al.*, 2001; Loram & Lakie, 2001). Secondly I could construct a mathematical model that would relate the changes in ankle torque to the neural modulation and changes in ankle rotation. I constructed the simplest model possible and used this to calculate both a neural gain and an intrinsic mechanical stiffness. The method, results and conclusion from this model are presented in **Chapter 3** and elsewhere (Loram & Lakie, 2002b). The problem with this simple model and any more sophisticated models I might want to construct is that they all rest on assumptions. What I really wanted was a direct measurement of the intrinsic mechanical stiffness both in balancing the inverted pendulum and in quiet standing. By intrinsic mechanical stiffness I mean the stiffness that operates instantly, without change in neural modulation, to generate a torque in response to any angular perturbation of the ankle. To reflect the stiffness operating in standing, these measurements should be carried out without disturbing the process of pendulum balancing or quiet standing. As far as I was aware, this had not been done before.

Whilst I recognised that the mechanical structure surrounding the ankle joint is a complicated compound system comprising of muscles, tendons and ligaments I thought that the muscle stiffness was the relevant property that determined the instantaneous stiffness of the ankle in standing. This idea was based on the assumption that the tendons were substantially stiffer than the muscle. I assumed that the muscle was the most compliant linkage in the chain passing from the foot through the Achilles tendon and triceps surae muscle to the skeleton of the leg. Indeed, papers about the role of ankle joint stiffness tend to refer to muscle stiffness in general terms and thus perpetuate the idea that muscle stiffness specifically is the relevant quantity that provides instantaneous torque response to a perturbation (Grillner, 1972; Morasso & Schieppati, 1999; Zatsiorsky & Duarte, 2000; Winter *et al.*, 2001).

My aim was to measure muscle stiffness in standing. One question I wanted to test was whether or not the intrinsic ankle stiffness in standing was sufficient to stabilise the human inverted pendulum. Another hypothesis I wanted to test was the theory that the intrinsic ankle stiffness caused by tonic muscle activity is varied to control sway size (Winter *et al.*, 1998). And finally, I wanted to see whether the directly measured intrinsic muscle stiffness could be related to muscle models such as the distribution moment model of muscle activity (Cholewicki & McGill, 1995). To investigate these latter questions it would be necessary to measure muscle stiffness sufficiently frequently to record of changes of stiffness through time. Ideally I wanted to measure the stiffness every 40 ms to provide a continuous record of stiffness that could be related to the torque, angle and EMG measurements. This requirement dictated that the perturbation actuator would need a very rapid response time since it would need to complete a cycle of perturbation and recovery within approximately 40 ms.

Choice of actuator.

Since it was necessary to produce rapid perturbations of the ankle it was obvious that I needed to displace the foot relative the body and leg rather than the other way round because of the large inertia of the body and the small inertia of the foot. Ankle stiffness can be measured either by applying a known position perturbation to the foot and measuring the torque response or by applying a known torque perturbation to the foot and measuring the angular deviation of the foot. Results from both methods will not necessarily yield the same answer (Hunter & Kearney, 1982) not least because it is difficult to produce a rapid deflection of the foot using a torque perturbation from say a torque motor. Elastic torque will only be generated when the mass of the foot is moved and the elastic structures surrounding the ankle are stretched. The applied torque would have to be very large to produce sufficient acceleration to move the foot and footplate an appreciable distance in a short time. The measurement of torque response to angular perturbations more closely corresponds to the natural situation in quiet standing where angular changes in ankle angle will be responsible for stretching the associated muscles and producing a mechanical torque response.

I considered the possibility of using a hydraulic mechanism to displace the foot. This would without doubt displace the foot by whatever angle I chose but I thought it would not be easy to control the speed of the perturbations or the waveform. It would also be expensive. The idea of using a piezo-translator was suggested because of its rapid response and recovery times measured in micro to milliseconds, its high intrinsic stiffness and its controllability. The disadvantage of piezo-translators is that they have very small displacements measured in microns. I found a manufacturer, (P.I., Germany), who produced a translator capable of producing throws up to 0.1 mm. Positioned at 0.2 m from the ankle this would give 0.03 degrees of rotation. Provided this rotation could produce a measurable torque response it offered the possibility of studying muscle stiffness under dynamic conditions. If necessary I could average many perturbations to increase the signal to noise ratio and use time locked averaging to see how the stiffness changes as the body or pendulum sways forwards and backwards. Alternatively, by binning the perturbations into groups of similar ankle torque, ankle velocity or EMG, I would be able to study the effect of these variables on muscle stiffness.

An advantage of small perturbations is that they will not disturb the standing or pendulum balancing task too much. Given that median ankle deflections in quiet standing and pendulum balancing are about 0.1 degree and 0.2 degrees respectively then an ideal range of displacements would have been from 0 to 0.25 degrees. With a distance of 0.2 m from the ankle this would required a piezo-translator with a maximum throw of 1 mm. From the best quality supplier that I could find, (P.I., Germany), piezo translators with a throw of 0.2 mm are available but they are more expensive. They also required a 1000V power supply and amplifier which would also have pushed the cost up and have been more difficult to construct in house. So I opted for the cheaper translator with a throw of 0.1 mm and with a power supply requirement of 100V. Translators came in open loop form with an approximate translation per unit voltage input and closed loop form which included a position sensor and feedback servo which precisely controlled the amount of translation that actually occurred. While the closed loop form might be appropriate for precision adjustment of a telescope lens, for my purpose precision control was not important so long as I could measure the amount of translation that actually occurred. Fortunately I already possessed a highly sensitive contactless displacement sensor that could be used to measure the amount of displacement caused by the piezo translator. Also, given that the open loop form also offered the fastest response times I chose an open loop translator (LVPZT P-840.60, P.I., Germany)

Experimental set-up of the translator and footplate.

Figure 1 - 'Series arrangement of PZT' shows the schematic series arrangement of the pendulum, the piezo translator (PZT), the load cell and the leg and body of the subject. Figure 1A in **Chapter 5** shows a more accurate diagram of the footplate and Figure 1C

shows the general experimental set-up including the pendulum, platform, footplate and subjects. At the front end the PZT mechanically contacts the load cell which is attached to the platform and pendulum via a stiff strut. At the back end the PZT contacts the footplate and foot which is mechanically linked to the lower leg of the subject by the musculature and ligaments surrounding the ankle joint. If the subject is balancing the pendulum the subject's body is stabilised by being strapped to a fixed vertical support. If the subject is standing freely, the pendulum is fixed in position. The PZT had to be in series with the load cell in order to measure the torque response to ankle rotation. The PZT could not be attached between the toe end of the footplate and the platform otherwise the torque applied to the footplate by the subject would be shared between this parallel arrangement of the load cell and the PZT. Upon expansion of the piezo-electric ceramic the pendulum/platform and the footplate are pushed in opposite directions. If the subject is strapped balancing the pendulum this expansion pushes the pendulum to the vertical and causes the toe end of the footplate to move up, although virtually all the movement occurs at the footplate. If the pendulum is fixed and the subject free to move, the subject is pushed more to the vertical and the toe end of the footplate is still pushed up. Again virtually all the movement occurs at the footplate.

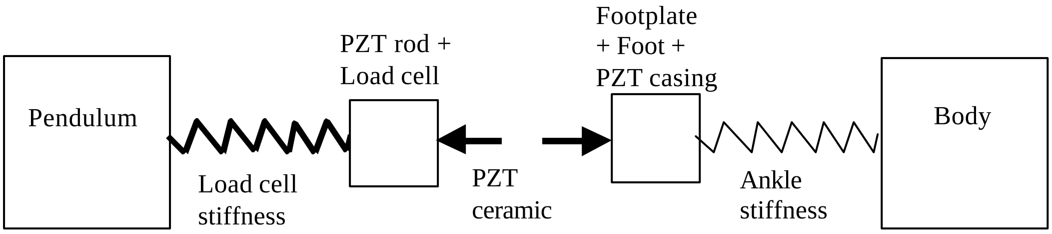

Figure 1. Series arrangement of PZT.
The casing of the piezo translator (PZT) is attached to the footplate. When a voltage is applied the piezo ceramic expands and the PZT rod moves and pushes onto the load cell which is attached to the pendulum. The movement of the rod is resisted by the very high stiffness of the load cell (which generates a force reading) and the high inertia of the pendulum. The PZT casing is pushed back onto the footplate. The footplate and foot rotate through virtually the full throw of the PZT and are resisted by the stiffness of the ankle and the inertia of the human body. Either the subject's body is fixed by being strapped to a vertical support, or the pendulum is fixed by being locked in position.

How much of the PZT throw would translate into rotation of the footplate?

The ceramic PZT is immensely stiff and was expected to behave as a true position actuator. However, not all of its small throw (0.1 mm) would translate into the rotation of the footplate. I was considering the case where the pendulum was balanced by a strapped subject. I expected there to be some loss of footplate rotation due to movement of the pendulum and compression of the torque cell. The pendulum has a very large moment of inertia (63 kg m^2), perhaps a thousand time the combined moment of inertia of the footplate and foot (~0.04 kg m^2) and was expected not to move much in the short duration of the translation (~ 40 ms). At the other end of the series mechanical system shown in Figure 1 - Series arrangement of PZT' is the body of the subject which is strapped to the fixed support when the pendulum is balanced. Given the inertia of the subject and the fact that they were strapped, I considered this end of the system to be fixed. As the PZT expands, based on a static displacement analyses, the displacement of the footplate relative to the subject will depend on the stiffness of the ankle and the stiffness of the load cell. The load cell was expected to compress in response to the translation thus recording a torque. From the technical fact sheet I estimated the load cell as having a stiffness (k) of more than 9000 N/mm. With the load cell at its current distance from the ankle (l) of 0.2 m this gave an angular stiffness (K) of 360,000 Nm deg^{-1} (using $K = k\ l^2$). I wasn't sure what ankle stiffness to expect but based on published data I anticipated a value of around 5 Nm deg^{-1} for each leg. On this basis I expected the static translation of the torque cell to be 72,000 less than the translation of the footplate. I concluded that effectively all the translation of the PZT would be transmitted into rotation of the footplate.

The likelihood of successfully measuring ankle stiffness.

With this method the ankle rotations were likely to be extremely small. The load cell was mounted 0.2 m from the ankle axis so a PZT throw of 0.1 mm would result in an angular rotation of 0.03 degrees. My first concern was what would actually be displaced by this rotation. It was possible that only the soft tissues on the sole of the foot would be deformed and that there would be no actual rotation of the ankle and no stretching of the Achilles tendon. It was imperative that I developed a method to measure what was actually being stretched or deformed. Ideally I wanted to measure the elongation of the triceps surae muscle in the same manner that Gurfinkel et al had done in their 1994 paper on foot deformation (Gurfinkel *et al.*, 1994). These authors measured the change in distance between the heel and the calf using a thread and a strain gauge.

My second concern was whether the torque response to the footplate rotation would be measurable. I anticipated that there would be a visco-elastic torque response from the ankle musculature, passive ankle structures and the foot itself. I expected the viscous torques to be significantly lower than the elastic torque (Hunter & Kearney, 1982; Kearney & Hunter, 1982; Mirbagheri *et al.*, 2000). Since I wasn't sure what ankle stiffness to expect I assumed a lower estimate of 2.5 Nm deg^{-1} per leg. This gives a maximum elastic torque response of 0.075 Nm. The resolution of the load cell was 0.01 Nm but the noise level was 0.03 Nm so if the elastic ankle stiffness was low I would need averaging to improve the signal to noise ratio. There is also an inertial torque component from the acceleration of the load cell, footplate, and foot. I didn't calculate the inertial torque and really should have done. With a PZT frequency of 25 Hz to complete a cycle in 40 ms the maximum angular acceleration would be 2 rad/s^2 ($accel = w^2 * A = 2*\pi*25^2 * 0.005 = 2$ rad/s^2). An inertia of 0.04 kg m^2

gives a maximum inertial torque of 0.08 Nm which is equal to the elastic component and which would tend to obscure the elastic response. I also didn't anticipate the viscous component of the apparatus which probably originates in the bearings.

Initial tests on the footplate apparatus.

As shown on Figure 1A in **Chapter** 5, a contactless variable reluctance displacement sensor was installed to measure footplate rotation by sensing changes in the position of the toe end of the footplate relative to the platform. I applied perturbations to the footplate in contact with the load cell by tapping the toe end of the footplate with a metal object and by hand. On the torque and footplate rotation signals this produced a ringing at around 40 Hz which would interfere with the measurements of the torque response to the PZT translations. By hand I could feel that the whole footplate was vibrating at about that frequency. By eye I could see that the biggest source of compliance was the strut that connected the load cell to the footplate. By applying loads I noted the toe end of the footplate was deflecting by about 150 µm for 12 Nm of applied torque giving an angular stiffness of 300 Nm deg^{-1}. With a resonance at 40 Hz this gave an estimate of the footplate inertia of 0.25 kg m^2. Taps applied to the torque cell alone also produced ringing in the torque signal around 800 Hz which I attributed to resonance of the torque cell itself determined by its high stiffness and low mass. This resonance indicated that the mass of the footplate coupled to the 'spring' of the load cell could also be a possible source of unwanted vibration.

Redesign of the footplate and strut to increase the frequency of resonance.

The resonant frequency of the footplate apparatus needed to be raised so that it did not interfere with the stiffness measurements. This was done by increasing the stiffness and decreasing the inertia of the footplate and strut assembly. The load cell was moved from 0.2 to 0.1 m from the ankle axis. This shortened the flexing strut which should make it stiffer. However this also subjects the load cell to forces which are twice as large which effectively increases the sensitivity and decreases the range of the load cell. The increased sensitivity is good for registering changes in ankle torque produced by the PZT which were likely to be near the noise level. The decreased range is bad because the offset would have to be adjusted for each subject and possibly for each trial to ensure that the voltages produced by the load cell remain in the -5 to +5V range required by the analogue to digital sampler. The range was now only about 0 - 15 Nm ± offset which corresponds to a 3 degree range of pendulum movement. This relocation also subjects the load cell to forces which are nearer to its limit of destruction. If someone stood on the outer part of the left footplate with their full body weight they would probably break the load cell. In the interest of preserving the fairly expensive load cell I did not mount the load cell and PZT any closer to the ankle axis than 0.1m.

Since the PZT is mounted in line with the load cell, the relocation of the load cell doubled the angular rotation of the footplate from 0.03 to 0.06 degrees. The increased rotation of the footplate means that the stiffness can be measured in response to a more meaningful ankle rotation. However it also means that the acceleration of the footplate during a PZT translation is twice as great causing excitation of inertial torques and resonances in the apparatus.

The footplate itself was redesigned and rebuilt. The original footplate made from thin steel was replaced with a smaller, aluminium footplate that was cross braced for added rigidity.

Aluminium was chosen for its lightness and holes were drilled in appropriate places to reduce the weight further. A repeat of the tapping experiments showed that the new footplate in contact with the load cell and strut revealed a resonance at around 100 Hz which was an improvement on the previous design.

The 100V amplifier for the PZT was constructed in house. Unfortunately, the amplifier transmitted high bandwidth noise to the PZT which produced an audible hissing. The hissing was unacceptable in case it reduced the life of the ceramic PZT. A smoothing capacitor connected to the output of the amplifier was used to substantially reduce this hissing.

The waveform used for PZT translations.

A signal generator was available for producing a variety of waveforms to drive the PZT. I expected a reflex response to arrive after 70 ms and so I wanted to complete the stiffness measurement within that time. Initially I used an uneven sinusoidal waveform that started at its minimum value, rose to its peak value in 70 ms and then decreased more rapidly. The rapid decrease was to shorten the time taken between perturbations. Only one cycle was given. I found this produced large transients in the torque signal at 70 ms when the direction of the footplate reversed and at the end of the perturbation when the footplate was decelerated to rest. These large accelerations at the end of the perturbation could be registered by the subject and had a destabilising effect. So I modified the waveform to an even sinusoidal form that had equal rise and fall times of 70 ms. This minimised the latter transients caused by acceleration of the footplate.

The size of the perturbations.

I wanted the largest perturbation and ankle rotation possible so that the rotations corresponded to the amount of ankle rotation that occurs during normal standing. The safe operating voltage of the PZT is 0 to 100V for long periods of time and -20V to 120V for short periods. Sustained voltages out side the 0 to 100V range do shorten the life of the PZT. The available oscilloscope had a maximum range of 80V so I estimated a pattern that looked like it was going from -10V to about 100V. I used the footplate rotation sensor to confirm that that this was giving a footplate rotation of about 0.05 to 0.06 degrees.

The duration of the perturbations.

I wanted to determine the range of perturbation durations that could be used to measure stiffness. To provide a known elastic resistance to perturbation I secured the toe end of the footplate to the platform using a spring. I placed separators from a feeler gauge in between the PZT and load cell to rotate the footplate and extend the spring by a variety of static displacements and calculated the angular stiffness of the spring from the best fit linear relationship between the readings of torque and footplate rotation.

Using the uneven sinusoidal waveform mentioned above I varied the rise time from 27 to 110 ms, taking an average of 10 perturbations. The shorter the rise time, the greater the inertial and viscous components of the torque response and more importantly the greater the amplitude of the transient resonance. The transient vibration of the footplate is excited during the three periods of maximum acceleration (i) when the footplate is accelerated initially, (ii) when the footplate velocity is reversed and (iii) when the footplate is decelerated to rest. These transient vibratory components obscured the inertio-visco-elastic component I wished to measure. The larger transients indicated that the PZT translation was really causing the footplate and strut assembly to vibrate internally and not to rotate as a single unit. This was confirmed by the lack of fit between the torque response and the footplate rotation measured by the sensor at the toe end of the footplate. I decided to relax the original aim of measuring the stiffness every 40 ms and concentrate on measuring the stiffness within the 70 ms period before the arrival of the short latency stretch reflex. So I opted for a rise time of 70 ms which would minimise the non-elastic components and allow a complete ankle stretch within the reflex free period. Also I experimented with a rise time longer than 70 ms which would minimise the non-elastic components further although this would not allow such a large stretch to be completed within the 70 ms period. On subjects I experimented with rise times of 50, 70, 83 and 100 ms which corresponded to single pulses at 10, 7, 6 and 5 Hz. The rise time of 50 ms did not give a measurable inertio-visco-elastic response because it was obliterated by the apparatus resonance. The rise time of 100 ms gave the clearest elastic response with the least transients and gave rotation speeds which were most akin to the ankle rotation speeds which occur in quiet standing.

When the subject is balancing the pendulum or balancing their own body centre of mass there is a related fluctuation in ankle torque. During the rotation of the footplate there is an additional torque component arising from the stretching of the ankle musculature. To calculate the ankle torque arising from the perturbation the underlying balancing torque has to be subtracted away. The longer the duration of the perturbation, the longer is the period of uncertainty during which the underlying balancing torque has to be estimated. Perturbations with a 100 ms rise times had a 200 ms period of uncertainty during which I didn't know the underlying changes in balancing torque. I wanted to be able to calculate the stiffness from individual unaveraged translations and so needed the minimum uncertainty in balancing torque during the perturbation. For this reason I decided to compromise on the duration of the perturbation and opted to use 7 Hz pulses with a rise time of ~70 ms and a complete pulse time of ~140 ms. This also allowed me to make a full stretch in the 70 ms reflex free period.

Calculation of the visco-elastic torque response of the ankle joint.

The method of calculating the torque response to the perturbation by subtracting the underlying balancing torque is described in **Chapter** 5. The torque response that remained still contained transients and inertial components that were associated with the footplate and not with the visco-elastic properties of the ankle joint. In an attempt to calculate and remove these extraneous footplate components I applied PZT translations to the footplate when no subject was standing on the apparatus. The gravitational moment of the footplate generated a small torque on the load cell. Symmetrical, single pulse PZT translations of 5, 6, 7, 8, 9 and 10 Hz were applied and the torque response was recorded and averaged over ten perturbations. For the 70 ms rise time pulse the torque response was small, largely inertial with little evidence of transient ringing, thus it corresponded to acceleration of the footplate as a single mass. This largely inertial response of the footplate was subtracted from the torque response when a subject was standing on the apparatus. However, the relative lack of transients indicated that the footplate does behave differently when a subject is standing on it. Perhaps the added inertia due to the subject's foot decreases the resonant frequency of the footplate and foot so that the transient ringing is more strongly excited.

Calculation of the elastic stiffness.

In **Chapter 5** I describe how I fitted a second order equation to the subject's torque response to the perturbation including the period from 0.4 seconds before the perturbation to 0.6 seconds after the start of the perturbation. From this fit I calculated the inertial, viscous and elastic parameters of that torque response. The small error in that fit appeared to be accounted for by the remaining transient vibrations. I wanted to test whether the stiffness in the reflex free period (70 ms) was the same as the stiffness in the complete period of the perturbation (140 ms) and thereby whether reflexes were influencing the torque response in the latter half of the perturbation. I subtracted the relatively small inertial and viscous components from the torque response and plotted the remaining torque response including residual transients against footplate rotation. The stiffness was calculated from the best fit straight line fitting the first 70 ms of the perturbation. Results showed little difference between this value of stiffness and the parameter calculated from the one second window surrounding the perturbation which indicated that reflexes are not influencing the stiffness in the second half of the perturbation.

Apparatus for measuring ankle deflection and foot deformation.

Since the size of the perturbations was rather small I wanted to establish whether the footplate rotation was actually causing the ankle to rotate and was stretching the Achilles tendon. I wanted to measure the true displacement of the heel relative to the back of the calf in the manner of Gurfinkel et al 1994 who used thread and a strain gauge. I obtained a football shin pad with straps which was modified to use as an apparatus support. Metal arms were added on which a strain gauge could be attached and from which a thread could be tied. A metal plate was made that I could attach to the heel using dental wax and tape. I proceeded as far as connecting the shin pad arm to the heel plate using a sensitive mercury filled elastic cord used to measure changes in limb circumference due to blood flow. It became clearly apparent that this and any similar method would not have the desired frequency response for measuring changes in length on the time scale of milliseconds. Next I decided to use a miniature laser range finder which could be secured to the footplate during perturbations and reflected off the heel plate. This is shown in Figure 1B in **Chapter 5**. This sensor was ideal because it had rapid response, high resolution, was insensitive to target and laser rotation and was contactless. The main practical difficulty was keeping the subject heel in position so that the laser remained on target on the heel reflector plate. This method revealed that the heel was left behind by a few tens of microns as the footplate rotated downwards beneath it. This proved that the foot was deforming and that not all the footplate rotation was transmitted through the ankle. However, calculation of the actual ankle rotation involved making assumptions about whether the shin was moving backwards in response to the perturbation thereby also reducing the ankle rotation. Eventually I took the step of strapping the laser to the lower leg using the modified shin pad and directly measuring the changes in position of the heel relative to the calf. These changes in distance mirrored the footplate rotation and thus confirmed that the ankle was being rotated and unrotated by the footplate. The laser was mounted below the main belly of the calf muscles because of the range of the laser but also so as to reduce movement of the laser due to muscle contraction. The insensitivity of the laser to rotation means that even if the position of the laser moved laterally due to muscle movement, the range between the laser and the heel would be unaltered. Ankle rotation measured using the laser actually provided a better fit to the transients of the torque response than did the rotation of the footplate measured at the toe end of the footplate. This suggests that the transient vibrations are occurring predominantly at the strut, load cell and heel end of the footplate rather than the toe end of the footplate.

Evaluation of PZT translator method of measuring ankle stiffness.

The use of this technique has enabled me to measure the intrinsic ankle stiffness directly with an appropriate size perturbation for natural ankle movements in quiet standing and in a way that does not disrupt the standing process. To my knowledge this is the first time this has been done.[1]

One main limitation of the method is the resonance of the apparatus which limits perturbations to a minimum rise time of 70 ms and a minimum cycle time of 140 ms. Since I have not isolated the source of these vibrations I cannot be sure how to solve this problem. The most likely source of vibration is oscillation of the footplate on the strut so an improvement can probably be gained by further stiffening the strut and decreasing the mass of the footplate. The other limitation is the maximum throw of the PZT. A greater range of throws would enable investigation of the intrinsic stiffness through a range of ankle rotations from below to above those encountered in quiet standing. It would also enable the thresholds for the onset of the stretch reflex to be investigated.

If I was repeating the experiments with the same apparatus I would have used perturbations with a 100 ms rise time throughout. A rise time of 70 ms was chosen mainly to minimise the uncertainty in estimating the torque applied to balance the pendulum during the perturbation and thus enable the stiffness to be calculated from individual perturbations. The original aim

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[1] I have subsequently found out that Gurfinkel et al (1974) carried out an experiment that was in principle similar to my own.

was to give a temporal record of changes in stiffness. However the stiffness values from individual perturbations have not yielded any useful results. The values are very noisy and within a single trial do not appear to correlate with anything meaningful such as ankle torque, ankle position, muscle, EMG or ankle rotation velocity. (To obtain less noisy stiffness measurements from individual perturbations I would need to shorten the duration of the perturbations). Thus this technique at present only works with averaged perturbations. With averaging one would expect the uncertain fluctuation in balancing torque to average out to zero. Results given in **Chapter 5** (Figure 3) show that the perturbations actually produce a neurally modulated reaction with an torque onset latency of about 140 ms as well as a small apparatus vibration that is transmitted to the right footplate. The unexpected result is that this neurally modulated reaction torque is much smaller with 100 ms translations than with 70 ms translations so the underlying changes in balancing torque are actually smaller with the slower, 5Hz translations. And indeed there was no stretch reflex response after 70 ms as reported in **Chapter 5**. So I would have obtained more certain stiffness measurements by using 100 ms translations throughout.

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CHAPTER 5.

DIRECT MEASUREMENT OF HUMAN ANKLE STIFFNESS DURING QUIET STANDING:

THE INTRINSIC MECHANICAL STIFFNESS IS INSUFFICIENT FOR STABILITY.

Summary.

During quiet standing the human "inverted pendulum" sways irregularly. In previous work where subjects balanced a real inverted pendulum, we investigated what contribution the intrinsic mechanical ankle stiffness makes to achieve stability. Using the results of a plausible model, we suggested that the intrinsic ankle stiffness is inadequate to provide stability. Here, using a piezo-electric translator we have applied small, unobtrusive mechanical perturbations to the foot while the subject was standing freely. These short duration perturbations had a similar size and velocity to movements which occur naturally during quiet standing, and they produced no evidence of any stretch reflex response in soleus, or gastrocnemius. Direct measurement confirms our earlier conclusion; intrinsic ankle stiffness is not quite sufficient to stabilise the body or pendulum. On average the directly determined intrinsic stiffness is 91% ± 23% (mean ±SD) of that necessary to provide minimal stabilisation. The stiffness was substantially constant, increasing only slightly with ankle torque. This stiffness cannot be neurally regulated in quiet standing. Thus we attribute this stiffness to the foot, Achilles tendon and aponeurosis rather than the activated calf muscle fibres. Our measurements suggest that the triceps surae muscles maintain balance via a spring-like element which is itself too compliant to guarantee stability. The implication is

that the brain cannot set ankle stiffness and then ignore the control task because additional modulation of torque is required to maintain balance. We suggest that the triceps surae muscles maintain balance by predictively controlling the proximal offset of the spring-like element in a ballistic-like manner.

Introduction.

In quiet standing, the body mass is generally regarded as being inherently unstable with the body centre of mass (CoM) maintained a few centimetres in front of the ankle joint. Close examination reveals quasi-random alternating movements of the centre of mass in the sagittal plane. The gravitational torque on the centre of mass is generally considered to increase linearly with ankle angle (Winter *et al.*, 1998; Morasso & Schieppati, 1999) and we refer to this relationship as the toppling torque per unit angle. Forward collapse of the body is prevented by ankle torque produced by activity of the triceps surae muscles.

The activated triceps-surae musculature generates an intrinsic mechanical stiffness across the ankle joint. Such stiffness provides an instant torque response to any change in ankle angle without any intervention required from the nervous system (Grillner, 1972; Horak & MacPherson, 1996; Winter *et al.*, 1998). What is the extent of this free restoring force?

If the ankle stiffness is less than the toppling torque per unit angle then the body CoM is mechanically unstable and an active neural modulation of ankle torque is required to produce stability (Morasso *et al.*, 1999; Morasso & Schieppati, 1999). Conversely, if the ankle stiffness is greater than the toppling torque per unit angle, the body CoM is in principle at least marginally stable. However, to account for the relatively high frequency of small sagittal oscillations which make up the sway pattern, stiffness would have to be greater still. Based on a mean frequency of 0.5 Hz, Morasso et al have suggested a value of 200% relative to the toppling torque per unit angle (1999; Morasso & Schieppati, 1999). A similar value can be derived from Winter et al (1998). Thus active neural modulation of ankle torque is still required if the intrinsic mechanical stiffness is less than 200% of the toppling torque per unit angle.

In experiments where subjects balanced a human proportioned inverted pendulum (Loram & Lakie, 2002), ankle torque changes were shown to result from anticipatory neural modulation as well as the intrinsic mechanical ankle stiffness. Using a plausible model it was predicted that the intrinsic ankle stiffness was just insufficient to provide marginal stability of the real inverted pendulum or the body CoM in standing.

In this study we have measured the intrinsic mechanical ankle stiffness using a method which allows the activity of standing quietly to continue undisturbed and which uses perturbations which are comparable to the ankle movements normally experienced in quiet standing. The essence of the technique is that it measures the intrinsic stiffness which is the stiffness before the nervous system has time to produce any change by reflex or other means. Gurfinkel et al (1974) attempted something similar but it is not clear from their paper that their technique can precisely distinguish between the intrinsic and neurally generated stiffness. With subjects lying on their backs, ankle stiffness has been measured for a variety of torque levels using a pseudo-random binary sequence of relatively large perturbations (Hunter & Kearney, 1982; Mirbagheri *et al.*, 2000), (5 degs, 1.7 degs resp.). These measurements also showed that the ankle stiffness increased markedly as the size of the perturbation decreased (Kearney & Hunter, 1982) although the perturbation size was not decreased to values comparable with typical ankle deflections in quiet standing. Using a rapid release ergonometer, the series elastic stiffness of the triceps-surae has been measured at different torque levels for subjects sitting (de Zee & Voigt, 2001) and for subjects standing with the availability of a bar (Hof, 1998). The size and speed of these releases were very large (30 degrees at 860 degs s^{-1}, > 25 degrees at 800 degs s^{-1}) and much greater than values encountered in normal standing movements. It has been claimed that the mechanical, series elastic ankle stiffness has been measured directly during standing using simple regression of ankle torque against ankle angle for an extended time period (Winter *et al.,* 2001). However this, and the author's previous method (Winter *et al.*, 1998) are invalid because they do not take into account the changes in torque caused by changes in muscle activation. For their claims to be true it is necessary to demonstrate that all the changes in ankle torque over an extended period of 10 s are caused by mechanical stretching of the muscle without any sway related neural modulation. Others have suggested that torque is generated as a result of reflex activity in standing and that the gain of these reflexes can be altered thus changing the effective stiffness (Fitzpatrick *et al.*, 1992a; Fitzpatrick *et al.*, 1992b; Fitzpatrick *et al.*, 1996).

The novel method presented in this paper uses a piezo-electric translator to apply small perturbations to the foot while the subject is standing freely or balancing an equivalent inverted pendulum. These perturbations have a similar size and velocity to the ankle movements that are normally encountered during this activity. As well as measuring the operative intrinsic, mechanical stiffness it also allows study of the reflex response relevant to the small, slow ankle movements that are normally present in quiet standing.

We address five questions. 1. What is the effective intrinsic, mechanical ankle stiffness during quiet standing and when balancing the inverted pendulum? 2. Is ankle stiffness a neurally controlled parameter or is it a biomechanical constant? 3. Can this stiffness be partitioned into foot and true ankle components? 4. Is there evidence of reflex activity during quiet standing? 5. What mechanism for controlling the body CoM is suggested by these results?

Methods.

Subjects.

Fifteen healthy people, eight male, aged between 20 and 68 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.

Experimental protocol

Subjects performed three tasks and we measured the intrinsic mechanical ankle stiffness of the left leg during each activity. First, they stood freely and quietly with their eyes open for a minimum period of 200 s. Second, they were strapped to a fixed, vertical support and shown an oscilloscope displaying the level of torque they were generating. They were asked to maintain a constant level of torque for 40 seconds. This experiment was repeated for randomised, different, torque levels varying between 5 Nm and 25 Nm. Third, they were strapped to a fixed vertical support and asked to balance a human proportioned inverted pendulum for a minimum period of 200 s. Subjects could see the pendulum position displayed on the oscilloscope and were asked to maintain the pendulum at 3 degrees so as to approximate the level of ankle torque applied during quiet standing. All subjects could perform this task after minimal familiarisation.

Figure 1. Ankle stiffness measuring apparatus

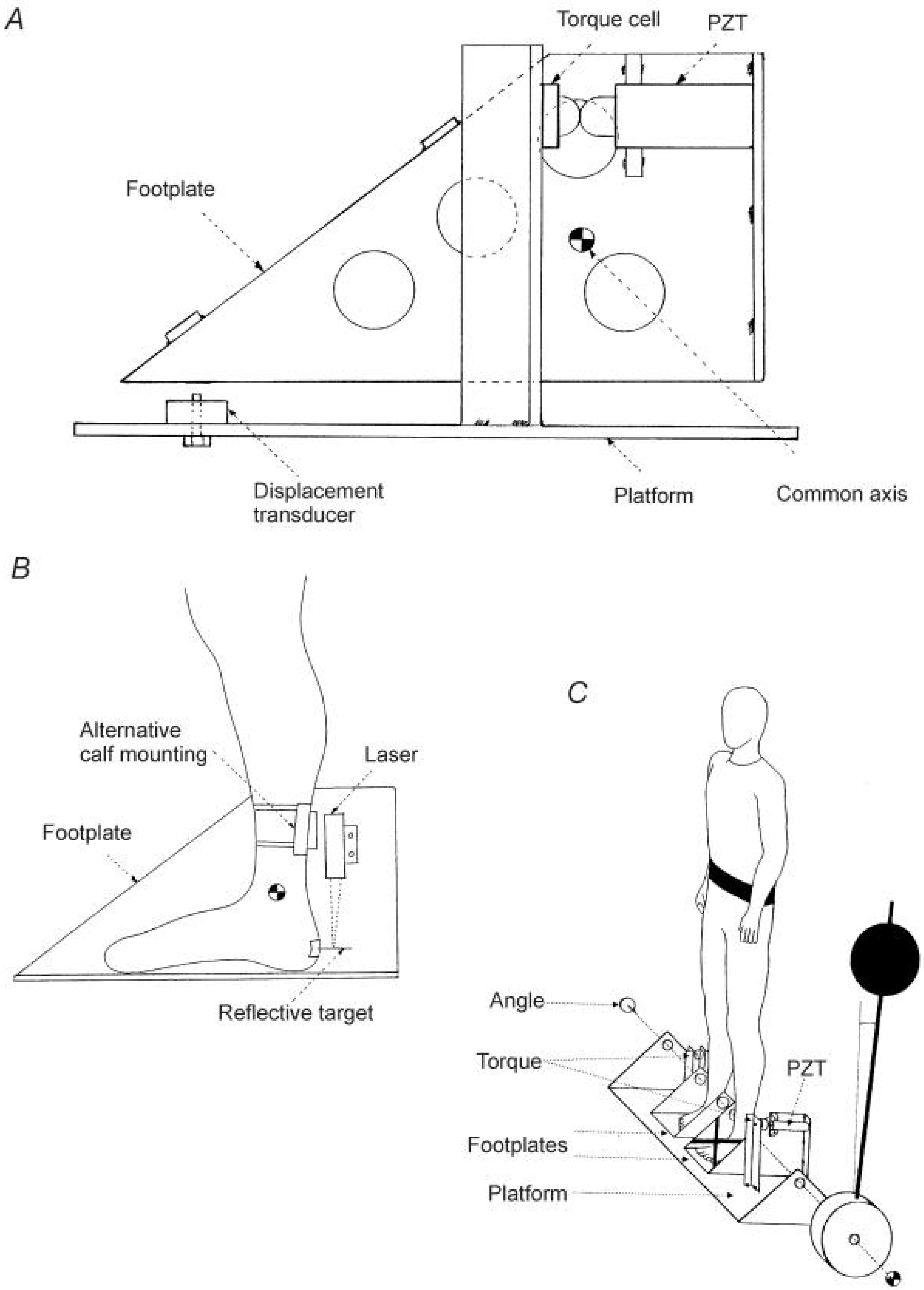

Figure 1. Ankle stiffness measuring apparatus.

Panel ***A***. Ankle stiffness measurement. The left footplate is constructed of aluminium alloy, lightened by holes and cross-braced for rigidity. A Piezo-electric transducer (PZT) produces a translation which rotates the footplate relative to the platform. Lengthening of the element raises the toe end of the footplate. The footplate rotation is registered by the contactless displacement transducer. The resulting force change is recorded by the torque cell. The contact face of the PZT is spherical to minimise off-axis forces.

Panel ***B***. Ankle rotation. A miniature laser range-finder operating by triangulation and insensitive to rotation measures the linear distance to a target attached to a mount moulded to the subject's heel with dental wax. The laser can be attached to the footplate as shown or alternatively mounted on a snugly fitting calf mould securely taped to the leg.

Panel ***C.*** General view. The subject stands on two footplates. Both footplates are coupled to the platform by horizontally mounted load cells which record ankle torque. The platform is rigidly coupled to a heavy inverted pendulum. A piezoelectric, vibrating gyroscope mounted under the platform measures angular velocity. In free standing the platform and pendulum are immobilised and the apparatus remains stationary while the subject sways. In the torque generation and pendulum balancing tasks the subject is strapped at pelvis height to a solid back support (not shown) that prevents body movement. During pendulum balancing the pendulum and platform sway while the subject is static. The backward lean of the pendulum mimics the normal forward inclination of the body and is measured by a contactless, precision potentiometer. The ankles, platform and footplates have a common axis. Alignment and support are provided by six precision ball races and a substantial steel framework (omitted here for clarity).

Apparatus.

The subject stood on two footplates with the centre of their ankles approximately 22 cm apart. Their ankles were positioned to be co-axial with the axis of rotation of an inverted pendulum (Figure 1C). The apparatus and sensor instrumentation for balancing the inverted pendulum has been reported fully elsewhere (Loram *et al.*, 2001) and is shown in Figure 1C. The left footplate was fitted with a piezo-electric translator (LVPZT P-840.60, P.I., Germany) which applied a rotation to the footplate causing dorsi-flexion of the ankle joint (Fig. 1A).

While standing, subjects stood freely without any mechanical contact or support and the pendulum and platform were locked to provide a horizontal surface. Subjects adopted their own standing position and their mean ankle angles ranged from 1.5 to 4 degrees. While subjects generated constant levels of torque the pendulum and platform were locked and a vertical support was moved forwards so the subject could be strapped at their normal standing position. While balancing the inverted pendulum the subject was strapped to the same fixed back support while the pendulum and platform were free to move coaxially with the footplates and feet. An appropriate pendulum mass was used for each subject. This was usually 60 kg although 40 or 50 kg was used for smaller, lighter subjects. The distance of the pendulum centre of mass from the axis of rotation was 0.94 m.

The piezo-electric translator had a maximum throw of 100 μm and was positioned to give 0.055 degrees of footplate rotation. The maximum throw was used for all experiments. For each perturbation a raised cosine waveform was used to minimise footplate acceleration and the accompanying reactive inertial torques. This method works on the assumption that there is no responsive change in muscle torque during the perturbation. Isolated perturbations with a rise time of 70 ms and a period of 140 ms (7 Hz cosine wave) were used. The average speed of these perturbations is 0.7 degrees s^{-1}. The effect of the piezo-electric translator is simultaneously to push the toe end of the footplate up and the pendulum or human centre of mass to a more upright angle. The inertia of the footplate and foot is less than 1/1000 times that of the pendulum or the human body so there is negligible deflection of these latter during the perturbation. A few subjects could feel the perturbations clearly when they were standing freely and when they were strapped to the support generating low constant levels of ankle torque. Others were never able to perceive the perturbations. For all subjects, the perturbations merged into the background when ankle torque levels were high and when torque fluctuations were greater such as when balancing the pendulum. There were no auditory cues perceptible

A contactless variable reluctance displacement sensor (Model 502-F, NS020, EMIC, France) with a sensitivity of 1mV μm^{-1} and response time of < 0.1 ms recorded rotation of the left footplate relative to the platform. A piezoelectric vibrating gyroscope measured velocity of the platform relative to the ground. A Hall effect precision potentiometer measured pendulum position and horizontally mounted miniature load cells recorded left and right ankle torque. The piezo-translator was mounted in series with the left torque cell. The footplate rotation and left torque were sampled at 1000 Hz. Other sensors were sampled at 25 Hz.

Rotation of the footplate produces deformation of the foot and rotation of the ankle joint (Fig 7A). In order to measure the relative contributions of each it is necessary to measure either lengthening of the Achilles tendon or deformation of the foot relative to the heel. To assess whether the Achilles tendon was being stretched by the perturbations a laser range finder (YP05MGVL80, Wenglor Sensoric, Germany) with a resolution of < 2 μm and a response time of 5 ms was used to measure vertical changes in distance 'y' between the heel and the calf. The laser was strapped to the back of the lower leg approximately 10 - 15 cm above the heel and reflected off a small metal plate firmly attached to the back of the heel close to the calcanean tuberosity (Fig. 1B). Each heel plate was individually fitted using a dental wax moulding. The approximate horizontal distance 'x' between the ankle joint and the point of reflection of the laser behind the heel was measured. The variation in ankle angle, 'q', was calculated using $q = y/x$. There was some inevitable uncertainty in determining the exact position of the ankle axis of rotation. The absolute accuracy of q is limited by the absolute accuracy of x to ± 20%. To measure deformation of the foot, the laser was fixed to the footplate and reflected off the same heel plate. This measured deflection of the heel relative to the footplate resulting from deformation of the foot. The measurement assumes that the body centre of mass is not raised while the footplate rotates. Rather, the heel is left behind by the downward movement of the footplate underneath it. The laser signal was sampled at 100 Hz. The loss of ankle rotation caused by foot deformation was calculated by dividing the variation in heel height by 'x' as above.

EMG activity was recorded on the left leg from soleus, tibialis anterior, gastrocnemius medialis and gastrocnemius lateralis using bipolar surface electrodes with encapsulated preamplifiers. These signals containing the entire bandwidth were then amplified and passed through an analogue full-wave rectifier. During preliminary trials these signals were sampled at 1000 Hz. During later experiments they were passed through a r.m.s. averaging filter with a time constant of 100 ms and then sampled at 25 Hz. The sampled EMG level corresponding to the "noise floor" was 0.04V.

Methods of data analysis.

Perturbations were usually given regularly at rates of approximately 0.8 Hz. Experiments showed that giving perturbations irregularly made no difference. Perturbations were then averaged as shown in Figure 2. Figure 2A shows the averaged footplate rotation and figure 2B shows the velocity of the perturbation.

During standing, the torque response to individual perturbations is less than the natural fluctuations in ankle torque which are associated with balance (Fig 2C). Averaging reduces the size of the unrelated fluctuations relative to the mechanical response to the perturbation (Fig 2D). Typically for standing or balancing the pendulum 200 perturbations were averaged. When the subject was generating a constant torque 30 perturbations were averaged.

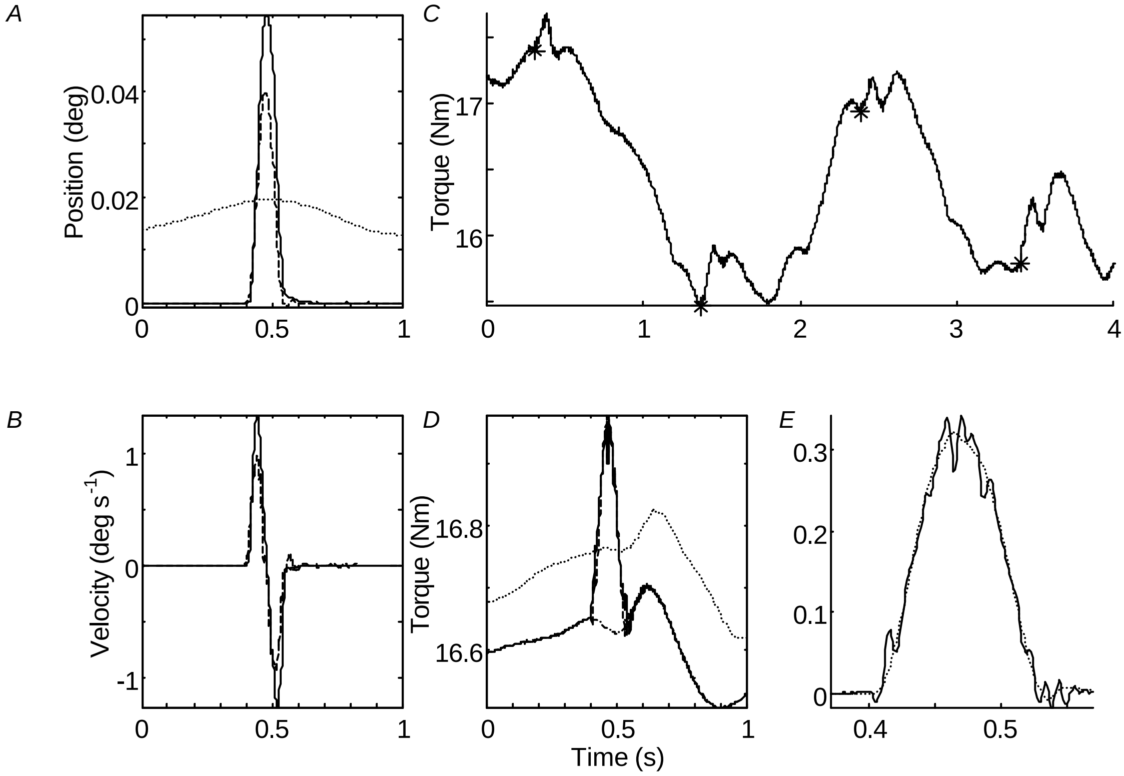

Figure 2 Averaging and calculation of mechanical response.

For one representative standing subject, panel ***A*** shows the averaged footplate rotation (solid), rotation of heel (calcanean tuberosity) about the ankle measured using the laser (dashed), and movement of the body CoM (dotted). The zero degree position is arbitrary. The footplate rotation starts at 0.4 s. Panel ***B*** shows the averaged velocity of the footplate (solid) and the heel about the ankle (dashed). Panel ***C*** shows the unaveraged time record of left ankle torque. The asterisks mark the beginning of footplate rotations. Panel ***D*** shows the averaged record of left torque (solid), left torque after subtraction of footplate component (dashed), the interpolated background torque during the perturbation (dot-dashed) and the right torque (dotted). Panel ***E*** shows the averaged mechanical response to the footplate rotation (solid) and the torque computed from the elastic, viscous, inertial model (dotted).

Next the small torque response of the footplate is subtracted (Fig 2D). This eliminates the inertia, viscosity and gravitational moment of the footplate itself. Some high frequency vibration at around 100 Hz remains which we attribute to resonance of the apparatus. For the 140 ms duration of the perturbation from 0.4 to 0.54 sec, we need to separate the mechanical torque response to the perturbation from the unrelated changes in ankle torque. By eye we draw a line between the left ankle torque at 0.4 s to the torque at 0.54 sec as shown in Figure 2D. Mathematically we use cubic spline interpolation to draw the line because a cubic spline matches both the value and the gradient at the two end points. A complication is that the remaining noise and 100 Hz vibration can throw out the gradient of the spline at the end points. So some smoothing is needed before applying the spline. First the left torque record in Figure 2D was replaced by a linear interpolation between 0.4 and 0.54 seconds. Then the torque was low pass filtered at 7Hz. Then the values between 0.4 and 0.54 seconds were replace by a cubic spline interpolation. The result is the dashdot line shown in Figure 2D and this is our estimate of the changes in ankle torque unassociated with the mechanical response to the perturbation. This estimate was subtracted from the torque record in Figure 2D to give the mechanical response to the perturbation which is shown in Figure 2E.

The mechanical response to the footplate rotation was modelled as having elastic, viscous and inertial components according to the equation $T = KA + Bv + Iaccel$ (Fig 2E). T is the mechanical torque response, A is the angle, v is the angular velocity and $acce$l is the angular acceleration. Linear least squares regression was used to estimate the parameters K, B and I which are the stiffness, viscosity and moment of inertia respectively. On average the percentage variance accounted for by this model (%VAF) was 99.0% and the torque response was predominantly elastic. The parameters *K*, *B* and *I* can be estimated with a high degree of certainty. The mean 95% confidence intervals were ± 1%, 2% and 20% respectively. These confidence intervals do not reflect the true uncertainty of the parameters because the torque response to which the parameters have been fitted is itself uncertain due to the interpolation procedure. We estimated that uncertainty by applying our interpolation procedure to the right ankle torque which is known throughout the perturbation (0.4 to 0.54 seconds). We calculated the difference between the known right torque and the interpolated right torque. We added this difference to the mechanical response of the left ankle torque in figure 2E to assess the effect of this uncertainty on the elastic, viscous and inertial parameters. The elastic, viscous and inertial parameters changed on average by 4%, 5% and 89%. The uncertainty in the inertial parameter is high because the inertial component is a small part of the mechanical response and indeed is only of minor interest in this study.

A 30th order linear phase FIR filter was used to differentiate the averaged footplate position and then to differentiate the velocity record. Using the Parks-McClellan algorithm, the filter was designed to differentiate the signal up to frequencies of 300 Hz (Ingle & Proakis, 1997). The differentiation was followed by a low pass FIR filter with a passband of 100 Hz.

The averaged records of footplate rotation, platform rotation and laser deflection also contained some residual variation unassociated with piezo-electric translation. These variations were subtracted using the same interpolative method as for the averaged torque record. A very small movement of the platform was subtracted from the rotation of the footplate relative to the platform to calculate the true rotation of the footplate relative to the ground. The torque response to true ankle rotation was modelled using the laser measurements from the back of the heel (mean %VAF= 99.5%). Likewise, the foot stiffness was modelled from the heel deflection relative to the footplate measured using the laser (mean %VAF=95.5%).

For each subject the approximate toppling torque per unit angle of the CoM was calculated using $m \times g \times h$ where m is the mass of the subject above the ankles, g is the gravitational field strength and h is the height of the CoM above the ankles (Table 1). Each subject was weighed and a corrective fraction of 0.029 (Patla *et al.*, 2002) corresponding to the mass of the feet was subtracted. The approximate position of the centre of mass was measured by lying subjects on a horizontal board and measuring the moment produced across a pivot (Page, 1978). The height of the ankles above the ground was subtracted from the height of the centre of mass.

During free standing, an estimate of the CoM angle was calculated from the ankle torque record using a low pass filter with a frequency cut off of 0.5 Hz (Caron *et al.*, 1997). We replaced Caron *et al*'s filter with a 1st order Butterworth filter since verification with real inverted pendulum data shows that this provides an improved estimate (Loram, unpublished observations).

Results.

When a dorsiflexion of 0.055 degrees is applied to the foot, approximately 70% of that rotation is transmitted through the foot and ankle joint to the distal end of the Achilles tendon and thereafter to the triceps-surae muscle-tendon complex (Figure 2A). The true angular deflection of the ankle shown in Figure 2A has been calculated from the change in length between the laser on the back of the lower leg and the plate attached to the heel. During the perturbation itself the angular deflection of the CoM is negligible (Fig2A). This means that approximately 30% of the footplate rotation is absorbed in the foot and is not transmitted to the ankle joint.

For an ankle stretch reflex, raw gastrocnemius EMG would have a latency of ~40 ms and the torque response would have an onset latency of ~75 ms reaching a peak value after ~170 ms (Stein & Kearney, 1995). The response would be in one leg only. The perturbations that we applied produced no evidence of a stretch reflex either during preliminary trials when the raw EMG was sampled at 1000 Hz or in subsequent trials when EMG was integrated and sampled at 25 Hz. Figure 3B shows averaged integrated EMG records while subjects were strapped to the fixed vertical support and were generating constant levels of ankle torque. In the left leg there was no unambiguous EMG response in soleus, gastrocnemius medialis, gastrocnemius lateralis or tibialis anterior. There was also no evidence of any neural modulation of torque occurring solely in the left leg (Fig 3A). However, in both the right and left legs there is evidence of a small, damped oscillatory variation in torque. This fluctuation in torque may be a small vibration transmitted through the apparatus or it may be an attenuated version of the balancing reaction described below. When the subjects were standing freely there was evidence of a small reaction in left soleus, gastrocnemius medialis and possibly tibialis anterior (Fig 3D). In the integrated EMG record this reaction starts approximately 100 ms after the start of the dorsiflexion and reaches a peak approximately 200 ms after the start of the dorsiflexion. For the standing activity there was a corresponding torque reaction in both the right and left leg (Figure 3C and Fig. 2D). Because the reaction occurs in both legs it is not a stretch reflex. The onset and peak of the torque reaction are ~140 ms and ~270 ms after the start of the perturbation respectively. It can be seen that the neurally modulated torque response begins as the mechanical response from the perturbation finishes and does not interfere with the calculation of the intrinsic mechanical stiffness. This reaction is inappropriate as it has a destabilising effect on the CoM which is accelerated to a more vertical position as seen in Figure 2A. The same EMG and torque reaction is seen when balancing the pendulum. The reaction is therefore present only when the intention is to balance an unstable load.

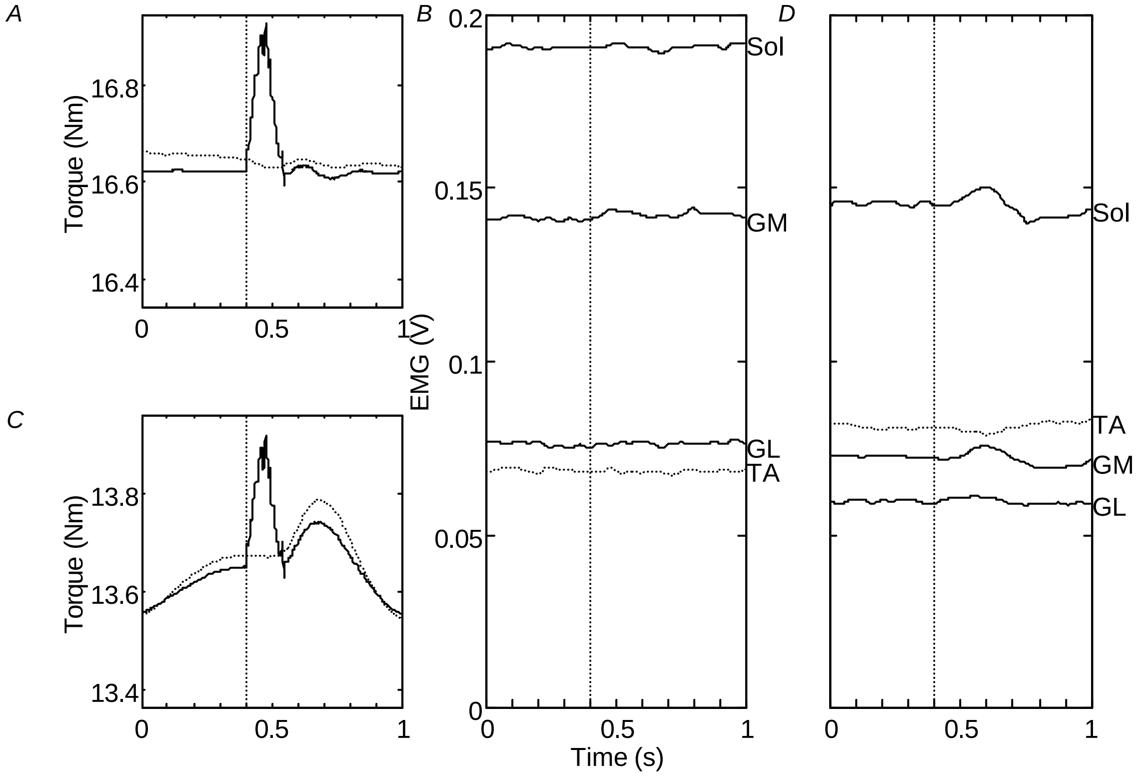

Figure 3. Neurally modulated responses to perturbation.
Panel ***A*** shows the left torque (solid) and right torque (dotted) record averaged from nine subjects while they maintained a variety of constant torque levels. Panel ***B*** shows the corresponding averaged integrated EMG records from soleus (Sol), gastrocnemius medialis (GM), gastrocnemius lateralis (GL) and tibialis anterior (TA) for the left leg. For the same nine subjects, panel ***C*** shows the averaged torque records while standing freely and panel ***D*** shows the corresponding averaged EMG records. The perturbations start at 0.4 s.

The accuracy of the measurement method was assessed by measuring the stiffness of a calibrated spring. The toe end of the footplate was fastened to the platform by a tension spring. Manual static displacement of the spring through small, calibrated distances provided an estimate of the angular spring stiffness using readings from the load cell and footplate rotation sensor. Measurement of the stiffness using the piezo-electric translator produced a value (4.6 ± 0.2 Nm deg $^{-1}$, mean ± SD) that was 18% higher than the static stiffness of the spring (3.9 ± 0.05 Nm deg $^{-1}$, mean ± SD). The coefficient of variation, 5%, was calculated by taking repeated measurements of the same spring. Using this spring a viscosity of 0.04 Nm s degree $^{-1}$ was measured which should be attributed to the apparatus. For four subjects the consistency of their ankle stiffness was assessed by repeating measurements after an interval of six months. For the quiet standing activity, their mean difference in ankle stiffness was 10% or 0.5 Nm deg $^{-1}$.

For fifteen subjects, the mean intrinsic mechanical ankle stiffness for the left leg during quiet standing and balancing the pendulum is 5.2 ± 1.2 (mean ±SD) and 4.7 ± 1.0 (mean ±SD) Nm deg $^{-1}$ respectively (Fig. 4A). The difference in stiffness between these tasks is significant but small (unbalanced two way ANOVA, N=48, F=11.6, P=0.002). By assuming both ankles have the same stiffness, we have calculated the combined ankle stiffness relative to the static toppling torque per unit angle of the body CoM or the pendulum CoM. The mean relative stiffnesses are 91% ± 23% (mean ±SD) and 80% ±19% (mean ±SD) respectively for standing and balancing the pendulum (Fig. 4B). The individual subject values of body mass, body height, height of CoM above ankle joint '*h*', toppling torque per unit angle '*mgh*', left ankle stiffness and combined relative ankle stiffness are shown in Table 1. For thirteen subjects the mean sway size and sway speed are shown (Fig 4C and D). For standing, the footplate rotation caused by the translator is 40% of the mean subject sway size and the mean footplate rotation speed is 5 times faster than the mean sway speed. While balancing the pendulum the sway size and speed are 2.5 times larger than while standing. Averaging over all subjects, the footplate rotations are larger than 29% of sways in quiet standing and are faster than all sways in quiet standing.

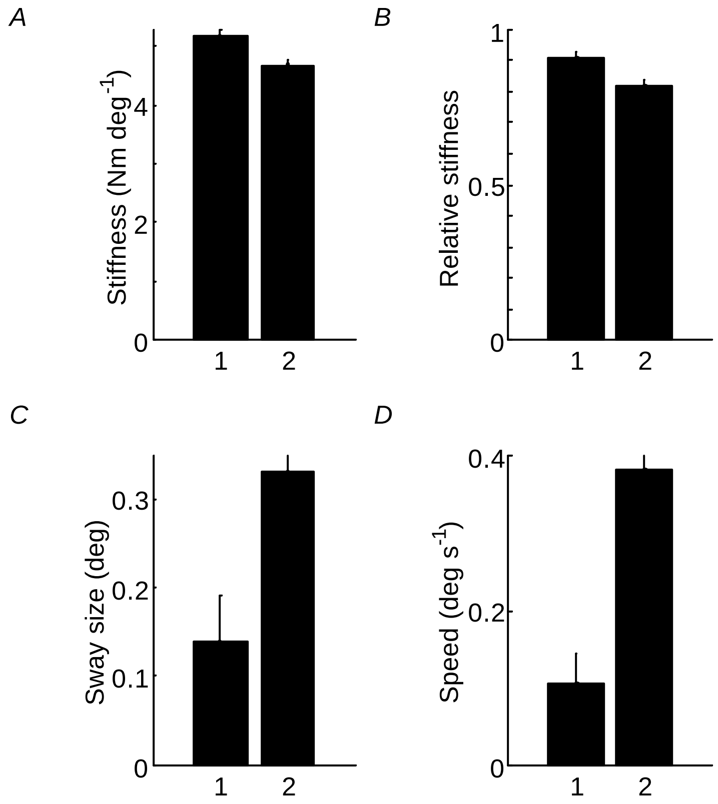

Figure 4. Stiffness during standing and balancing the pendulum.
The intrinsic, mechanical left ankle stiffness, averaged from fifteen subjects while they stood freely (1) and balanced the pendulum (2) is shown in panel ***A***. Panel ***B*** shows the combined stiffness of both legs relative to the toppling torque per unit angle of (1) the body CoM while standing (2) the pendulum CoM while balancing the inverted pendulum. A sway is defined as a unidirectional movement from one reversal point to the next. The median sway size and median sway speed, averaged from thirteen subjects are shown in panels ***C*** and ***D*** respectively for standing (1) and balancing (2). The uncertainty bars represent standard errors in the mean values.

Tables.

Table 1. Subject values of toppling torque per unit angle and ankle stiffness.

Subject	**Mass (kg)**	**Height (m)**	**h (m)**	**mgh (Nm/deg)**	**Left leg stiffness (Nm/deg)**	**Combined relative stiffness**
JH	78.0	1.83	0.97	12.63	5.6	0.89
NH	69.1	1.82	0.93	10.69	5.9	1.11
LT	77.4	1.62	0.86	11.11	4.9	0.87
NC	72.6	1.80	0.96	11.58	6.5	1.13
JR	70.9	1.60	0.87	10.20	1.9	0.37
MK	86.5	1.72	0.93	13.34	5.3	0.80
FO	83.1	1.73	0.96	13.30	5.0	0.75
DG	81.5	1.66	0.92	12.48	4.0	0.63
MH	80.4	1.71	0.90	11.96	6.0	1.01
JHa	73.5	1.65	0.89	10.81	5.0	0.92
KW	59.1	1.67	0.88	8.67	4.4	1.02
JW	60.5	1.74	0.88	8.87	6.0	1.35
SP	79.5	1.73	0.91	12.01	5.0	0.84
IL	69.4	1.78	0.97	11.18	5.8	1.03
ML	90.9	1.85	0.97	14.70	6.5	0.88
Mean	75.5	1.73	0.92	11.57	5.2	0.91
S.D.	8.9	0.08	0.04	1.64	1.2	0.23

Measurement of the intrinsic ankle stiffness while subjects maintained a variety of constant torque levels showed only very slight variation of stiffness with ankle torque (Fig. 5A). As ankle torque is increased from 5 to 25 Nm, the mean ankle stiffness rises from 5 to 6 Nm deg^{-1} at around 20 Nm and then slightly decreases. The large increase in muscle activation shown in Figure 5C is unable to produce much change in intrinsic mechanical ankle stiffness. For this reason, the ankle stiffness measured is not attributed to muscle fibre stiffness. The source of the measured stiffness is more likely to be the combination of aponeurosis, tendon and foot which will not change greatly with muscle activation. The measured viscosity increases slightly with ankle torque from 0.06 to 0.09 Nm s deg^{-1} (Fig. 5B). Up to 0.04 Nm s deg^{-1} is likely to result from the apparatus and not the subject's ankles. The inertia of the foot showed little variation with ankle torque as would be expected (Fig 5D). The mechanical response was mostly elastic. For each parameter and for all parameters combined the mean percentage variance accounted (%VAF) was 90% (elastic), 9% (viscous) and 0% (inertial), 99% (combined). The inertial component of our mechanical response is very low. As described in the Methods, estimates of the moment of inertia have a high relative error of 89% and are included to show an approximate value for the foot. The larger elastic component has a small relative error of 4%.

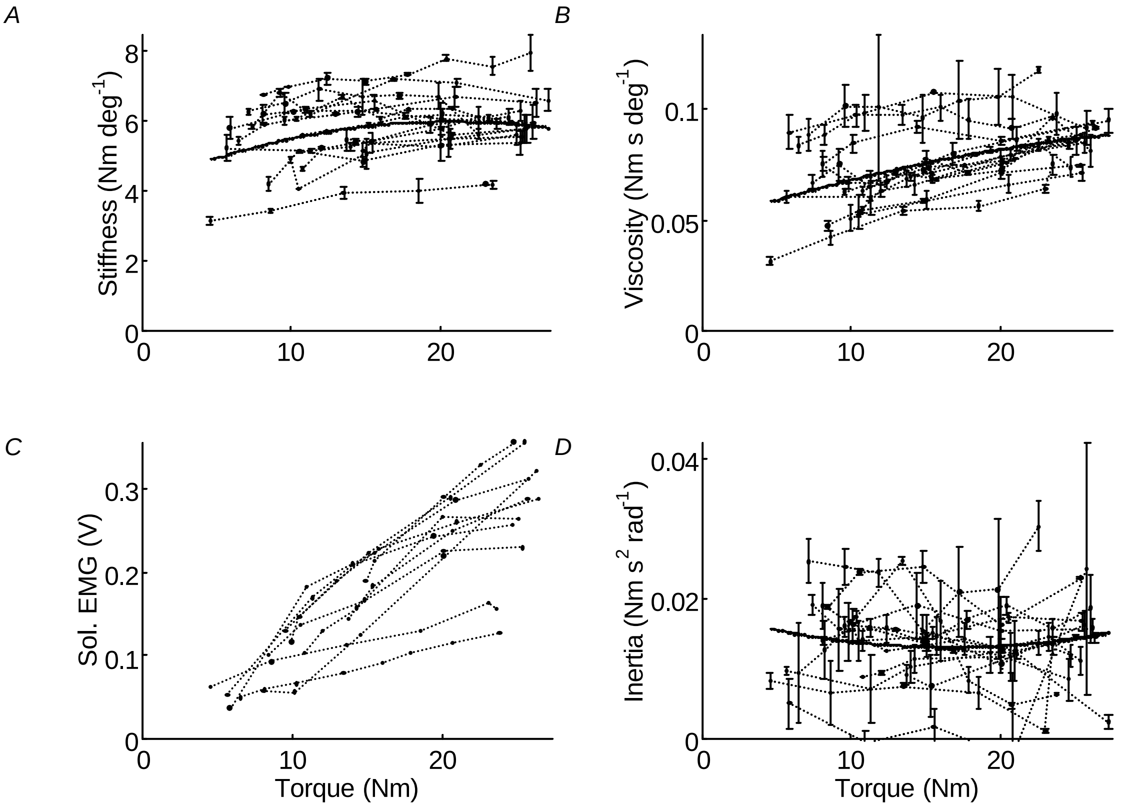

Figure 5. Variation of stiffness with ankle torque.

For fifteen subjects, the variation of intrinsic mechanical left ankle stiffness with ankle torque is shown in panel ***A***. Panels ***B*** and ***D*** show the variation with ankle torque of viscosity (ankle and apparatus) and inertia (foot) respectively. The error bars show the uncertainty associated with estimating the background torque during the perturbation. The solid lines represent the mean quadratic line of best fit. Panel ***C*** shows the variation of soleus EMG with ankle torque for eleven of the fifteen subjects. For these experiments subjects were strapped at their normal standing ankle angle to a fixed vertical support. Subjects maintained constant ankle torque using visual feedback from an oscilloscope and repeated this at different levels of torque.

Figure 6A shows that the perturbations are a sizeable fraction of the normal sway size during pendulum balancing and thus also during the smaller sways of quiet standing (Fig 4C). It can also be seen that most of the perturbation is transmitted to the distal end of the Achilles tendon though some is lost in the foot. The torque resulting from rotation of the footplate allows the calculation of the overall or combined stiffness. The overall stiffness is usually (as here) referred to as ankle stiffness. Figure 7 and the explanatory legend should be consulted for a definition of terms. Rotation of the footplate results in a sum of foot deformation and rotation at the ankle joint. The true ankle rotation is less than the footplate rotation because of the foot deformation. Depending on the position of the laser, the laser measurements can be used to calculate foot stiffness or the true ankle stiffness. For nine subjects where the laser measured deflection of the heel from the footplate, the foot stiffness is around 23 ±13 Nm deg^{-1} (mean±SD) compared with their overall stiffness of 4.7 ± 1.2 Nm deg^{-1} (mean ±SD), (Fig 6B). For six subjects where the laser measured deflection of the calcanean tuberosity relative to the calf, the true ankle stiffness during quiet standing is around 9.6 ± 1.3 Nm deg^{-1} (mean ±SD) compared to their overall stiffness of 5.9 ± 0.6 Nm deg^{-1} (mean ±SD), (Fig. 6B). Both foot stiffness and true ankle stiffness show some dependency on ankle torque. The true ankle stiffness increases with ankle torque from 8 Nm deg^{-1} at 5 Nm to a peak and plateau of approximately 11 Nm deg^{-1} at around 25 Nm (Fig. 6C). Conversely, the foot stiffness decreases with ankle torque plateauing at a mean value of 21 Nm deg^{-1} at around 20 Nm (Fig. 6D). During standing the foot stiffness is approximately twice as much as the true ankle stiffness. If the overall ankle stiffness is a series combination of foot and true ankle stiffness, then approximately one third of the compliance occurs in the foot and two thirds of the compliance occurs in the Achilles tendon and associated muscle.

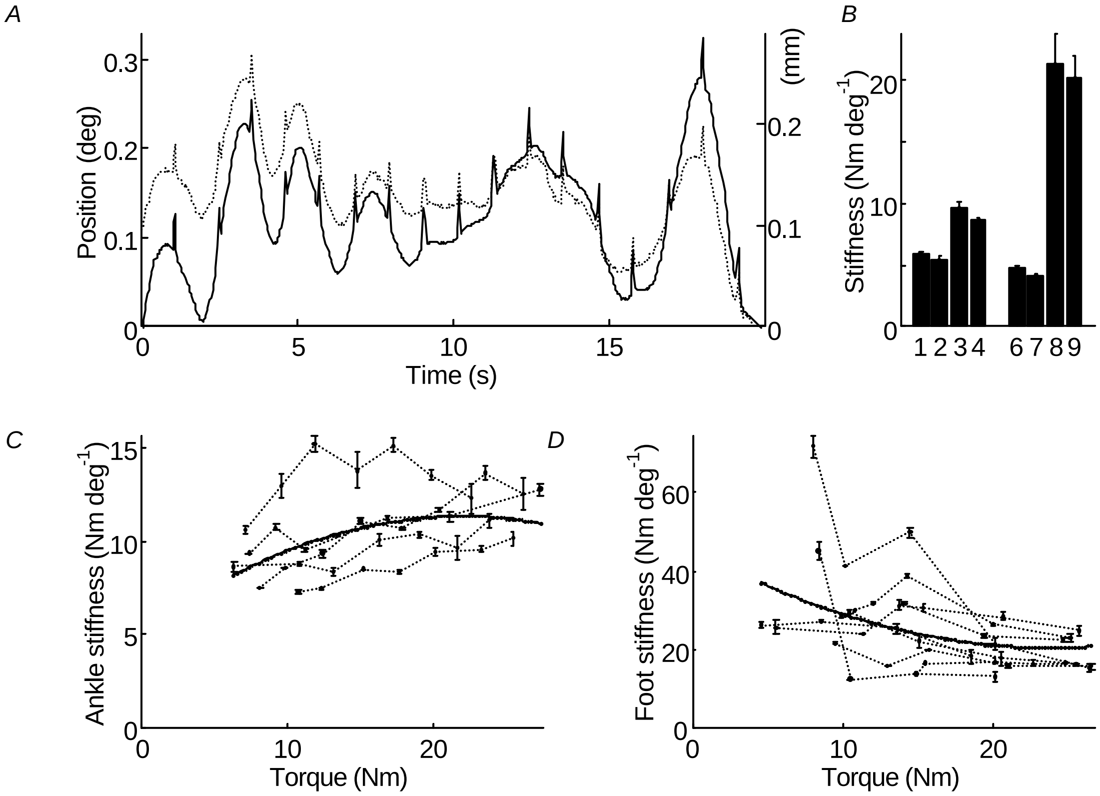

Figure 6. Partitioning stiffness into ankle and foot components.
For one representative subject balancing the inverted pendulum, panel ***A*** shows the unaveraged record of footplate angle (solid) and heel rotation about the ankle measured using the laser (dotted). The left axis scale shows angular changes of the footplate and heel in degrees about an arbitrary zero and the right axis scale shows linear movements of the heel relative to the calf in millimetres. Panel ***B***. For six subjects the mean, intrinsic, overall ankle stiffness while standing (1), and balancing the pendulum (2) is shown. The true ankle stiffness measured using the laser is shown for standing (3) and balancing the pendulum (4). For nine subjects the overall ankle stiffness while standing (6) and balancing the pendulum (7) is shown. The foot stiffness measured using the laser is shown for standing (8) and balancing the pendulum (9). For five subjects, panel ***C*** shows the variation of true ankle stiffness with ankle torque. Panel ***D*** shows the variation of foot stiffness with ankle torque for 9 subjects. For ***C*** and ***D*** the error bars show the uncertainty associated with estimating the background torque during the perturbation. The solid lines represent the mean quadratic line of best fit.

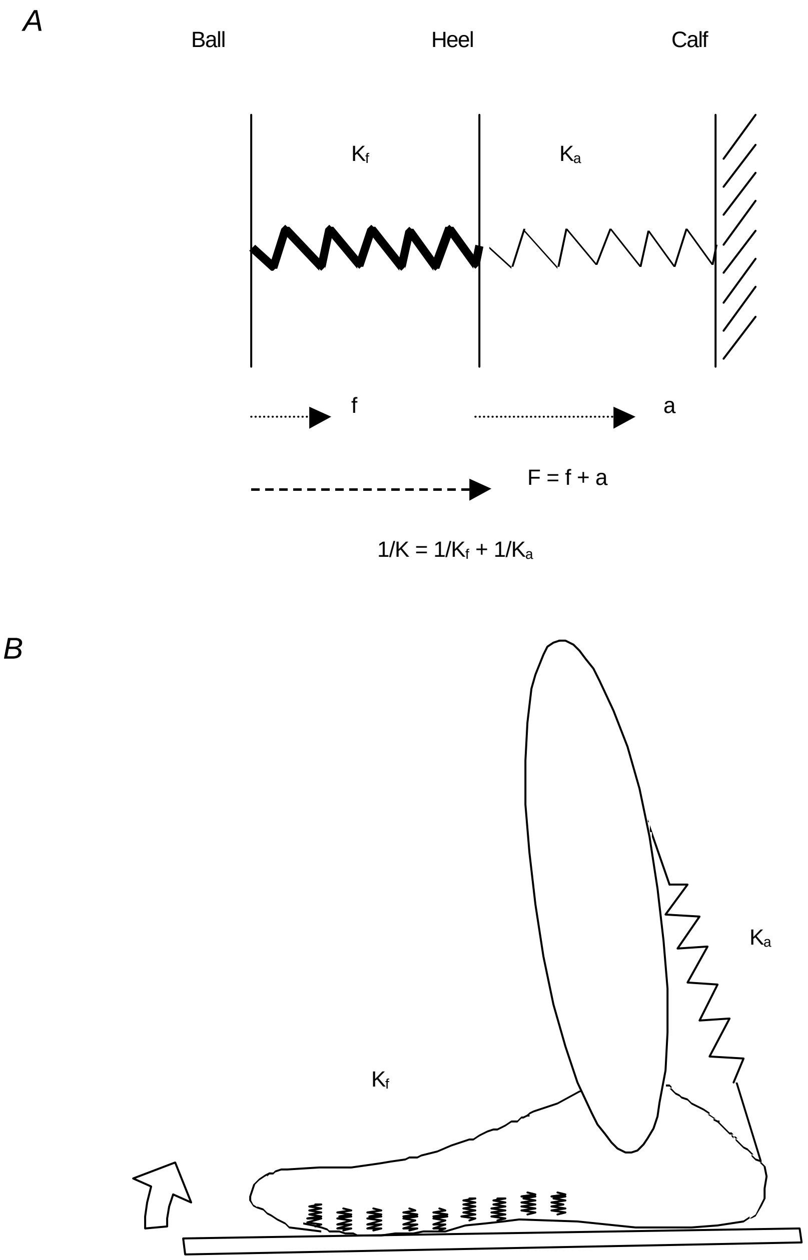

Figure 7. Stiffness and compliance of the foot and ankle.
Panel ***A***. Rotation of the footplate relative to the calf, *F*, is taken up by angular deformation of the ball of the foot relative to the heel, *f*, and rotation of the heel relative to the calf, *a* such that $F = f + a$. It is assumed that the calf does not move during the applied rotation of the footplate. The stiffness of the foot, K_f can be thought of as being in series with the stiffness of the ankle, K_a. In both series elements, the torque, *T*, is the same. The foot stiffness is calculated from the torque increment per unit foot deformation, $K_f = DT/Df$. Likewise the ankle stiffness is the torque increment per unit ankle rotation, $K_a = DT/Da$. The combined stiffness, *K*, is the torque increment per unit footplate rotation, $K = DT/DF$.

Compliance is the inverse of stiffness. The combined compliance, $1/K$, is the sum of the foot compliance, $1/K_f$, and ankle compliance, $1/K_a$, so $1/K = 1/K_f + 1/K_a$. Accordingly, the combined stiffness, K, of the two series springs is less than the weakest spring in the chain.

Usually researchers do not partition footplate rotation into foot and ankle components. The combined stiffness, K, is usually regarded as ankle stiffness. So that we may compare our results with previous work we will refer to the combined stiffness, K, as the ankle stiffness and we refer to the stiffness related purely to ankle rotation, K_a, as the true ankle stiffness.

Panel ***B***.
We think the foot stiffness, K_f, most likely resides in the soft tissues of the foot as well as the arch.
The true ankle stiffness, K_a, includes all components acting in parallel at the ankle joint.

Discussion

In order of discussion we are considering five questions. 1. Is there evidence of reflex activity during quiet standing? 2. What is the intrinsic mechanical ankle stiffness during quiet standing and balancing the inverted pendulum? 3. Is this stiffness a neurally controlled parameter or is it a biomechanical constant? 4. How does this stiffness partition into 'true ankle' and foot components? 5. What mechanisms of human balance are implied by these results?

Is there evidence of reflex activity during quiet standing?

Studies of reflex activity in the ankle joint musculature have commonly used perturbations that are large and rapid compared with movements that are normally encountered during quiet standing (Stein & Kearney, 1995; Mirbagheri *et al.*, 2000). Although the perturbations used in these experiments were smaller than a typical ankle movement during quiet standing they were approximately five times faster. Accordingly they might be expected to elicit reflex responses. In all experimental conditions we found no EMG or torque evidence of a stretch reflex in all fifteen of our subjects and thus conclude that stretch reflexes are probably not relevant to quiet standing (Fig. 3). Further investigation of the size and velocity thresholds required to elicit the reflex is necessary to consolidate this conclusion. Our results are consistent with previous research (Gurfinkel *et al.*, 1974). These authors used perturbations up to 0.2 degrees at 0.6 degs s^{-1} and from visual inspection of the raw EMG they also found no evidence of stretch reflexes.

However, when subjects were standing or balancing the inverted pendulum but not when strapped and maintaining constant levels of torque, there was a very interesting longer latency reaction in triceps-surae and tibialis anterior (Fig 3C,D). The reaction appears to be "approved or sanctioned" by the decision to balance (Berthoz, 2000) and is not a reflex in the classic Sherringtonian sense. This response was transmitted to both legs and given the latency is possibly of central origin. It was also inappropriate and destabilising causing the CoM to sway to the upright. It might have been a response to the proprioceptive illusion of falling generated by the increased pressure on the sole of the left foot and the dorsiflexion of the left ankle. The vestibular or visual senses would not have been stimulated because only the foot was moved. If the angular movement about the ankle joint corresponded to a head movement below the visual and vestibular thresholds then the movement was open to misinterpretation by the nervous system. It is a useful reminder of how balancing reactions can be inappropriate and destabilising (Diener *et al.*, 1984) and a source of sway generally (Loram & Lakie, 2002).

<u>The intrinsic mechanical ankle stiffness during quiet standing and balancing the pendulum.</u>

During the 140 ms period of the perturbation, there was no neural modulation of ankle torque and thus this technique measures the intrinsic, mechanical stiffness of the ankle. This stiffness is assumed to include components from the foot, the parallel elastic stiffness of the ankle, the Achilles tendon and the triceps surae musculature.

In quiet standing the ankle stiffness is approximately 91% of the static toppling torque per unit angle of the CoM (Fig. 4) and is thus insufficient to stabilise the human "inverted pendulum". This conclusion is valid whether or not human standing is in fact a true, rigid inverted pendulum provided that for the body CoM (i) the vertical acceleration is negligible, (ii) the angular acceleration about the ankles is proportional to the horizontal acceleration and (iii) the moment of inertia of the body about the ankles is constant (Morasso *et al.*, 1999). Essentially the same ankle stiffness was found when subjects were balancing the human proportioned inverted pendulum and this was 80% of that required for minimal stability of the pendulum used, although of course this comparison is dependent on the size of the pendulum used. The viscous component was found to be a very small part of the total response.

While stability and positional control are not provided by the intrinsic ankle stiffness alone, there is some useful instant mechanical torque response to any perturbation or change of ankle angle. In effect the action of gravity on the CoM is nearly cancelled out and so a reduced balancing problem is presented to the central nervous system. Although the system is still unstable and the nervous system has to generate an appropriate ankle torque for every ankle angle, errors in torque are not subjected to so much destabilising positive feedback. The situation is equivalent to having to balance an inverted pendulum of approximately 9% the actual toppling torque per unit angle although the inertia of the system is unchanged.

Is this stiffness a neurally controlled parameter or is it a biomechanical constant?

It has been argued that the nervous system sets or modulates ankle stiffness to control body sway (Winter *et al.*, 1998; Carpenter *et al.*, 1999; Gatev *et al.*, 1999). According to our measurements the intrinsic stiffness changes little with ankle torque despite a large change in muscle activation and so muscle fibre stiffness is unlikely to be the source of stiffness (Fig 5A,C). This leaves the aponeurosis, tendon, and foot as the likely sources of stiffness in standing. We have found that it is remarkably difficult to bring about any significant change in the intrinsic mechanical ankle stiffness. Our conclusion is that in quiet standing intrinsic ankle stiffness is not under neural control but is a biomechanical constant so far as the nervous system is concerned.

Comparison of our stiffness values with previous measurements.

Our values of intrinsic stiffness increase from 5 Nm deg^{-1} at 5 Nm ankle torque to a broad maximum of 6 Nm deg $^{-1}$ at 20 - 25 Nm torque (Fig 5A). Using perturbations, others have measured this stiffness over the same range of ankle torques (Hunter & Kearney, 1982; Mirbagheri *et al.*, 2000). They give values increasing from 2 and 3 Nm deg $^{-1}$ respectively at 5 Nm ankle torque to 5 and 6 Nm deg $^{-1}$ at 25 Nm ankle torque. Also most recent measurement of the triceps surae series elastic stiffness which, excludes the parallel ankle stiffness, shows an increase from 2 Nm deg^{-1} at 5 Nm to 4.5 Nm deg $^{-1}$ at 25 Nm (de Zee & Voigt, 2001).

At low torques our values are higher than those previously measured using perturbations and while our values follow the same increasing trend with ankle torque our trend is less pronounced and has a slight plateau or decrease at the higher torque values. Why has this difference in results occurred? An important explanation is that these authors used substantially larger and faster perturbations than we have used. Their earlier measurements have indicated that ankle stiffness increases substantially as perturbation size decreases though their measurements did not extend to the movement range experienced in quiet standing (Kearney & Hunter, 1982). There was the possibility that the small movements occurring in quiet standing are subjected to high short range stiffness caused by the friction-like and stiction-like properties of passive joint complexes (Winters *et al.*, 1988). Our perturbations were 40% of the median ankle movement during standing and are thus appropriate for measuring the stiffness encountered during quiet standing. Our results give weight to the idea that ankle stiffness is higher when the ankle movements are smaller and slower. This idea is illustrated by the fact that while standing the intrinsic ankle stiffness is 10% higher than while balancing the inverted pendulum (Fig 4A). Balancing the pendulum was associated with greater ankle movement (Fig 4C) and was not even remotely associated with any corresponding changes in ankle torque, EMG or ankle angle. Based on our absolute accuracy check using a calibrated spring, a second explanation of the difference between our measurements and previous measurements is that our values could be 18% too high though this would not account for the lack of dependency on ankle torque.

With a combined relative stiffness of 91%, the partial stabilisation achieved by the intrinsic, mechanical ankle stiffness is greater than has been predicted (Morasso *et al.*, 1999; Morasso & Schieppati, 1999). These authors have argued, rightly in our view, in favour of active stabilisation mechanisms. However we think they have tended to overestimate the human toppling torque per unit angle (Table 1). They also underestimated the intrinsic ankle stiffness by neglecting the fact that people have two legs.

In a similar experiment to our own Gurfinkel et al estimated the mean intrinsic ankle stiffness of five subjects to be 7.6 Nm deg^{-1} per leg. These gave a combined stiffness of 112% relative to the toppling torque per unit angle of their subjects. Subjects were standing on a force platform which was rotated toes upwards at 0.6 degs s^{-1} by up to 0.2 degrees. These unidirectional perturbations are the same order of magnitude as movements which occur naturally in quiet standing and like our own experiments the perturbations appear to have allowed the standing process to continue. Perturbations were averaged and ankle stiffness was calculated from the change in ankle torque divided by the change in ankle angle measured over an unspecified period up to 0.3 s following the perturbations. Our reservation about this work is that from their Figure 1 the subjects were clearly thrown backwards by the perturbation. This can be seen from the difference between the platform rotation and the change in ankle angle. The modulation of ankle torque associated with the change in position of the body centre of mass was not subtracted from the total change in ankle torque. In fact it is not clear that the change in torque was wholly mechanical, especially given the unspecified duration of the measuring period, the movement of the body centre of mass, the unidirectional nature of the perturbations and the raw EMG records shown in Figure 1. If these factors were included the estimate of stiffness would be reduced but we cannot say by how much.

Recently it has become possible by using dynamic ultrasonography to measure tendon stiffness in vivo without using perturbations. Measurement of the gastrocnemius tendon gives a mean stiffness of 3.4 Nm deg^{-1} rising to a value of 6.5 Nm deg^{-1} near maximum voluntary contraction of the muscle (Maganaris, 2002). These values exclude other parallel components to ankle stiffness such as other tendons and connective tissue. It is therefore expected that the mean value will be lower than our value for quiet standing.

Partitioning of compliance into foot and ankle components.

Our laser measurements of foot deflection and true ankle rotation allow us to partition the ankle stiffness into foot and true ankle components. Our finding for quiet standing that approximately one third of the angular compliance occurs in the foot and two thirds occurs at the ankle joint is in good agreement with a previous study of this question (Gurfinkel *et al.*, 1994). These authors found similar values of foot stiffness and a similar partition ratio for larger ankle displacements including up to two degrees.

Mechanisms of balance control.

Our direct measurements of ankle stiffness confirm our previous model based predictions showing that the intrinsic ankle stiffness is insufficient to provide stability of the human or the artificial, human proportioned inverted pendulum (Loram & Lakie, 2002). These results are consistent with our hypothesis that in quiet standing the human inverted pendulum is controlled by a neurally modulated, ballistic-like throw and catch process as was found for human balancing of the inverted pendulum (Loram & Lakie, 2002). Our results also complement the finding that operational ankle joint impedance (from intrinsic mechanical sources or neural modulation) is not a controlled parameter in the regulation of sway size (Loram *et al.*, 2001). Our results in this paper contradict the theories of Winter et al (1998; 2001). These authors argue that the nervous system sets the muscle tone sufficiently high to create muscle stiffness safely greater than the gravitational spring. They maintain that the nervous system then leaves the intrinsic stiffness to do its job.

When one is measuring the stiffness combination of springs in series such as the muscle fibres and the tendon, the value of stiffness is limited by the weakest spring. It is a common misconception (initially shared by ourselves) that the tendon has greater stiffness than the muscle. The misconceived idea is of an actuator (the muscle) connected to a lever (the heel of the foot) by a steel wire (the tendon). In the context of quiet standing, it seems from the lack of change in stiffness with ankle torque and muscle activation, that the combined inactive components of tendon, aponeurosis and foot are less stiff than the muscle fibres. It makes mechanical sense for the tendon to be more compliant than the muscle fibres for large as well as small movements. If the tendon and foot were stiffer than the muscle fibres then a sudden, forceful deflection of the foot, such as when walking up a step onto the ball of the foot, might provide a rapid, damaging pull on the muscle fibres. A more compliant tendon would buffer the muscle fibres from the perturbation by reducing the sudden lengthening of the muscle. However the increased tendon stiffness at high levels of ankle torque must be enough to generate the large torques required during running and jumping (Hof, 1998; de Zee & Voigt, 2001).

This conclusion suggests a simple model of standing. It suggests that the muscle fibres act as a stiff actuator which has the ability to change its length as a result of neural modulation. This actuator transmits torque to the ankle joint via a relatively weak spring. The spring has its length and tension altered by changes in position which occur at the distal end (the heel) and the proximal end (the tendo-muscular junction). The triceps-surae muscle controls the position of the proximal end of the Achilles tendon and thereby controls the tension in the spring and indirectly the position of the body CoM. As stated by Roberts, it should be remembered that the tension in a spring is a function not only of its stiffness but also its offset (Stein, 1982). Our conclusion is that the horizontal projection of the centre of mass is controlled by a spring offset control mechanism, not by a stiffness control mechanism. This mechanism requires that the proximal end of the weak spring (length of muscle fibres) be controlled in an anticipatory manner by the nervous system. It may be significant that the muscle spindles are well positioned to register the length of the muscle fibres and thus the spring offset. This idea and its physiological implications are to be discussed more fully in a subsequent article.

> A caveat.
>
> Compliance of the foot means that the axis of rotation of the body CoM is not a fixed centre through the ankle joint. As visual observation will confirm, the axis of rotation moves forward as the body sways forward and more torque is transmitted through foot. This may mean that for small sways close to the vertical the toppling torque per unit angle is less than it would be if the centre of rotation did not move. Thus for such sways the intrinsic mechanical stiffness could confer more stability than our calculations show. This possibility requires further investigation.

In conclusion, we find that in quiet standing the intrinsic, mechanical ankle stiffness is around 5 Nm deg $^{-1}$ per leg which for both legs amounts to 91% of the static toppling torque per unit angle of the body CoM. This stiffness is relatively constant and is not under neural control. One third of the compliance occurs in the foot and two thirds occurs at the ankle joint. We predict that the body CoM is controlled by anticipatory modulation of the proximal offset position of the weak spring which is the Achilles tendon. Our evidence is that stretch reflexes are not relevant to quiet standing.

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CHAPTER 6.

CONCLUSIONS AND HYPOTHETICAL MODEL.

From the studies reported in **Chapters 2**, **3** and **5**, a certain amount has been learned about ankle stiffness and about how a human proportioned inverted pendulum is balanced and controlled using the ankle musculature. In this final chapter I will present eleven conclusions that can be substantiated from my data and from the analysis I have carried out. While the conclusions concerning pendulum balancing primarily relate to the task concerned, the closest application to this task is quiet standing and so I will use my results to make predictions concerning sagittal sway in quiet standing. If these predictions are confirmed for standing that lends support to the idea that quiet standing approximates to the inverted pendulum model. If these predictions are not validated for quiet standing that indicates inadequacies in the inverted pendulum model of quiet standing. Finally, my conclusions have led me to a speculative yet plausible model of the ankle mechanisms used in quiet standing. This "dynamic bias" or "ballistic bias" model has interesting properties which I will describe at the end of this chapter.

Conclusion 1.

There is no one angle at which the pendulum is stabilised (Loram & Lakie, 2002b). In any one trial there is a mean angle at which the pendulum is maintained. With minimal familiarisation, most subjects can stabilise the pendulum for a sustained period at mean angles ranging from 1.5 degrees to 9 degrees of more. Moreover, equilibrium is only achieved instantaneously and is never achieved at exactly the same angle twice. Thus models concerning human balance of an inverted pendulum need to move away from the idea of maintaining a single sustained position of equilibrium either at the vertical or some other fixed angle. The implication for standing is that within anatomical constraints, inverted pendulum dynamics do not dictate any particular angle at which the body CoM should be maintained.

Conclusion 2.

In choosing to maintain the pendulum at a certain mean angle there is a trade off between the amount of effort required and the amount of control required. Balancing the pendulum further from the vertical requires greater effort. Balancing the pendulum close to the vertical (zero to one degree) also requires finer control than balancing the pendulum in the middle of the range (three to four degrees). Also, balancing the pendulum close to the vertical tends to involve modulation of tibialis anterior as well as triceps surae. It is an anatomical fact that subjects have a greater region of stability in front of the ankle joint than behind it and can apply plantar flexing torque more easily than they can apply dorisflexing torque. (i) The foot lever in front of the ankle is longer than the foot lever behind the ankle. (ii) The plantar flexor muscles are stronger than the dorsiflexing muscles. (iii) The moment arm of the plantar flexors is greater than the moment arm of the dorsiflexors. Thus bringing the pendulum close to the vertical means bringing it close to a boundary beyond which it cannot be controlled and so the margin for error is reduced. In quiet standing, individuals appear to maintain a characteristic angle for the centre of mass and in doing so they are making a certain choice in the trade off between effort, control and possible activation of tibialis anterior. It is interesting to question whether there is any criterion for deciding whether one choice of angle is better than another.

Conclusion 3.

Balancing the pendulum is an active process that requires sway related neural modulation of triceps surae (Loram & Lakie, 2002b). Balance cannot be maintained if muscle activation remains at a constant level. I predict that quiet standing also requires similar, sway related, neural modulation of soleus and gastrocnemius. This goes beyond the idea that soleus is merely activated in a constant tonic manner to generate torque and or stiffness (Horak & MacPherson, 1996; Winter *et al.*, 1998). When balance is lost, neural modulation tends to be reactive i.e. a response is triggered by the loss or impending loss of balance. The braking of movement, bringing the pendulum to instantaneous rest tends to be anticipatory.

Conclusion 4.

While balanced by a subject the inverted pendulum exhibits continuous, irregular sway. The origin of this sway is inappropriate torque applied by the subject in balancing an unstable load (Loram & Lakie, 2002b). Any torque imbalance will either accelerate or decelerate the unstable pendulum. If the subject is intending to move the pendulum then a deliberate torque imbalance is generated. If the subject is intending to keep the pendulum as still as possible then the torque imbalances are unintentional and will produce the small sways that result. For quiet standing there are generally no external perturbations as such but there are internal perturbations from the heartbeat (Sturm, 1980) and from respiration (Hodges *et al.*, 2002). It seems that respiration is usually compensated for by movement at a variety of joints and consequently has no effect on the CoM (Hodges *et al.*, 2002). The effect of the heartbeat on motion of the CoM is unknown. My prediction for standing is that irregular sway of the CoM will result from inappropriate ankle torque and this may in fact be the dominant cause of standing sway.

Conclusion 5.

When keeping the pendulum still and while standing quietly mean sway sizes are of the same order of magnitude as published values of sensory thresholds. When subjects were keeping the pendulum as still as possible, the subject's average, mean sway size was 0.17 degrees using visual feedback via the oscilloscope and 0.28 degrees using proprioceptive feedback alone (Loram *et al.*, 2001). In quiet standing, the average, median sway size was around 0.13 degrees (Loram & Lakie, 2002a). The perception threshold for ankle movements is reported as being 0.17 degrees at 0.06 degrees s^{-1} and less at higher velocities which are more normally encountered in standing sway (Fitzpatrick & McCloskey, 1994). Proprioceptive information from the legs provide the most sensitive feedback and vestibular thresholds were ten times visual and proprioceptive thresholds during normal standing sway. These estimates may be upper limits because they are the thresholds at which perception could be reported, not the threshold at which perception could be used in balance control. I hypothesise that for balancing the pendulum and for quiet standing, sway sizes cannot be reduced below a limit set by sensory thresholds. For movements below these thresholds the subject cannot actually tell whether the CoM is moving and in which direction it is moving. Thus for these small movements modulations of ankle torque are just as likely to be destabilising as they are to be stabilising. This hypothesis will be valid if compensation for small sways is provided by anticipation and reaction of the nervous system and will be invalid if small sways are accommodated by intrinsic mechanical properties of the ankle joint.

Conclusion 6.

For balancing the pendulum, sway size is not minimised by increasing the effective ankle stiffness where I define effective ankle stiffness as the rate of change of ankle torque with ankle angle (Loram *et al.*, 2001; Loram & Lakie, 2001). Sways of the pendulum are relatively slow and last one second on average. It is theoretically possible that during a sway the change in ankle torque per unit change in ankle angle could be increased to minimise sway size. For example, the equilibrium point stiffness hypothesis (Bizzi *et al.*, 1976; Polit & Bizzi, 1979) predicts that by varying levels of cocontraction at the ankle joint, a variable stiffness can be set for any ankle torque. This could be used to control sway size. For particular examples, Winter et al (1998) advocated that the CNS sets a certain amount of ankle plantar-flexor muscle tone and intrinsic ankle stiffness to control the amount of sway. Carpenter et al (1999) suggested that the CNS decreases sway by increasing the ankle stiffness. Gatev et al (1999) suggested that the effective ankle stiffness is centrally modulated in a predictive manner to control sway. Fitzpatrick et al (1992a; 1992b) claimed that sway size was controlled by altering reflex ankle stiffness. Regardless of the preferred mechanism, my results for balancing the inverted pendulum show that stiffness modulation in any of these forms is not the way that sway size is minimised (Loram *et al.*, 2001). I predict that in quiet standing, stiffness modulation is not relevant to the regulation of sway.

Conclusion 7.

For balancing the inverted pendulum and for quiet standing the intrinsic mechanical ankle stiffness is insufficient for stability. By direct measurement I found that during quiet standing, the intrinsic mechanical stiffness in response to very small perturbations was 91% ± 23% (mean ±SD) of the subject's toppling torque per unit angle (Loram & Lakie, 2002a).

When subjects were balancing the pendulum, their intrinsic ankle stiffness was 10% lower still. On average from all my subjects, the perturbations were larger than 29% of sways in quiet standing. It is expected that a smaller stiffness would be measured if the size of the perturbation was increased to the larger sway sizes encountered in quiet standing (Kearney & Hunter, 1982). Using a simple model that calculated the intrinsic mechanical stiffness from ankle torque, ankle angle and EMG data, very similar results were obtained for subjects balancing the pendulum (Loram & Lakie, 2002b). For both quiet standing and pendulum balancing this stiffness is insufficient to give even marginal stability and is greatly insufficient to account for the mean rate at which the CoM reverses direction.

Conclusion 8.

Pendulum sway resembles a series of intermittent, ballistic-like throws and catches of a almost entirely passively balanced system (Loram & Lakie, 2002b). Each throw and catch requires a biphasic torque pattern that is neurally modulated. If the intrinsic mechanical stiffness is 80 - 90 % of the gravitational toppling torque per unit angle then the inverted pendulum has been reduced to one with an effective gravitational spring of 10 - 20 % its actual value. Each sway is an intermittent step whose duration is uncorrelated with the duration of the previous step. The fact that successive sways do not maintain a constant phase relationship means that frequency domain analysis can miss features of this process. For example, each sway size depends to a large extent on the torque imbalance at the start of the sway (Loram & Lakie, 2002b). I predict that the same process occurs in quiet standing particularly in the sagittal plane. Inspection of Figure 2 in Winter et al (2001) lends support to this prediction.

Conclusion 9.

Sway size is minimised by modulating ankle torque more accurately (Loram & Lakie, 2002b)

The random like movement of the pendulum can be thought of as a repetition of a basic throw and catch process. Sway is minimised (i) by catching the pendulum more accurately so that it is more in balance when it comes to rest (ii) by reacting to a loss of balance more effectively so that the pendulum does not sway so far from rest given a certain initial torque imbalance. Thus sway minimisation relies on minimising inaccuracy in the detection of movement, in the estimation of the response and in the actuation of the response. I predict that this will also be true in quiet standing.

Conclusion 10.

The intrinsic mechanical ankle stiffness during quiet standing and pendulum balancing is not under neural control. It increases only slightly with ankle torque (Loram & Lakie, 2002a) and decreases slightly when there is more movement at the ankle joint (Loram & Lakie, 2002b, a). Thus the effective intrinsic, mechanical ankle stiffness is thought to reflect contributions from tendons and connective tissue surrounding the ankle joint as well as the stiffness of the foot. The largest contribution to this stiffness is thought to come from the Achilles tendon which is under tension due to the contraction of the triceps surae muscles. At 15 Nm of ankle torque the series elastic stiffness of the triceps surae is about 3.5 Nm deg^{-1} (de Zee & Voigt, 2001) which is around 70% of the stiffness values that I have measured.

Conclusion 11.

Short latency stretch reflexes are not elicited by the smaller sways in quiet standing and are unlikely to play a role in standing (Loram & Lakie, 2002a). This result is consistent with previous investigations (Kelton & Wright, 1949; Gurfinkel *et al.*, 1974; Nashner, 1976). I found that longer latency reactions are apparent in pendulum balancing and standing (Loram & Lakie, 2002a) and this is also consistent with the previous investigations.

A dynamic, bias model of the ankle mechanisms used in balance.

During pendulum balancing subjects produce a great variety of changes in ankle torque with contemporaneous change in ankle angle. This is illustrated in Figure 1 - 'Changes in torque and angle'. Most changes in torque with angle are elastic-like, (torque increases as angle increases) but are greater than the intrinsic mechanical ankle stiffness. Approximately 25% of torque changes are not elastic-like (torque decreases as angle increases). How can such a variety of changes in ankle torque be produced if the intrinsic mechanical stiffness is so low and constant? The results and conclusions gained so far have led to the formulation of a simple model of the ankle mechanisms used during pendulum balancing and during quiet standing. The model has a single contractile element connected to the heel by a compliant spring. This model is shown in Figure 2 - Dynamic bias model. The contractile element represents the triceps surae muscle and the spring represents the series elastic element of the triceps surae muscle. (Strictly the spring includes the series elastic element of the triceps surae in combination with the foot stiffness). A key feature of this spring is that it is too compliant to stabilise the real, or human, inverted pendulum. I expect this spring to have a stiffness of about 80 - 90 % relative to the gravitational spring of the CoM. The proximal end of the spring represents the myotendinous junction between the triceps surae muscles and the Achilles tendon. I hypothesise that, by shortening or controlled lengthening, the muscle controls the position of the proximal end of the spring. Thus I suggest that the CNS controls the bias of the series elastic spring in a dynamic manner.

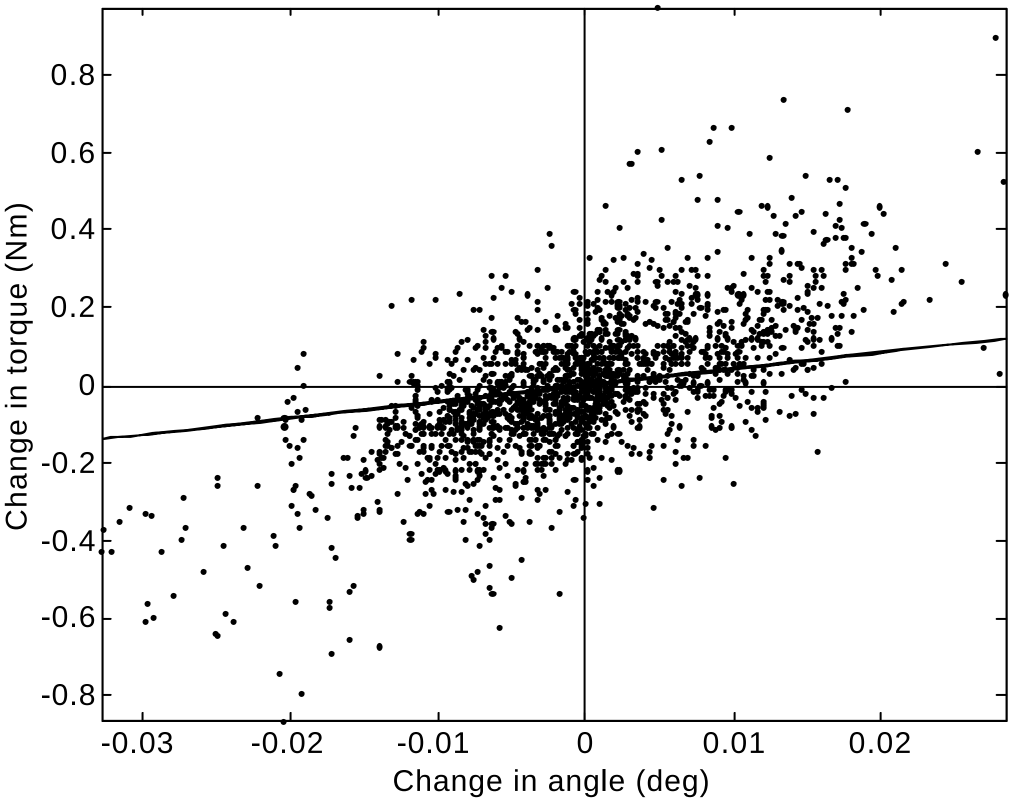

Figure 1. Changes in torque and angle.
Changes in left ankle torque at 40 ms intervals are plotted against the contemporaneous change in pendulum angle. The inclined straight line represents the directly measured, intrinsic mechanical ankle stiffness (4.2 Nm deg^{-1}) for the subject in this trial.

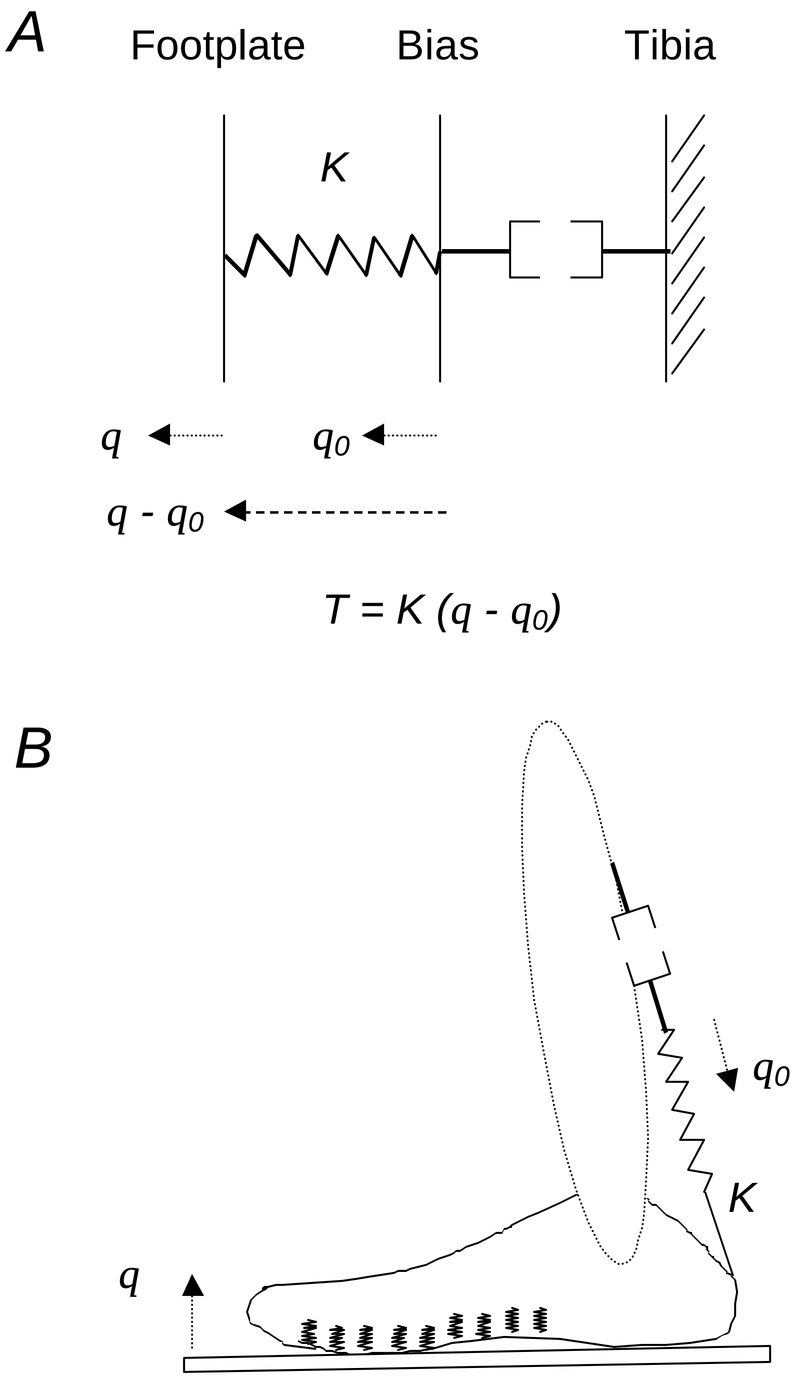

Figure 2. Dynamic bias model

Panel ***A*** shows a spring like element of intrinsic stiffness K in series with a contractile element. The spring like element acts between the footplate and the bias position. The contractile element acts between the bias and the tibia. The angular length of the spring is given by the angle of the footplate relative to the tibia, q, minus the position of the bias q_0, such that the ankle torque is given by $T = K(q - q_0)$. Panel ***B***. We identify the bias with the myotendinous junction between triceps surae and the Achilles tendon. The contractile element represents the length of the triceps-surae muscle fibres. The spring represents the series elastic element of the triceps surae, additional parallel ankle stiffness and the foot.

This theory has some similarities with the 'equilibrium point - invariant' or λ-model' hypothesis of Feldman (1986) who also suggested that the bias and not the stiffness of the spring is controlled by the CNS. There are two main differences between our theories. (i) In this theory the bias is controlled dynamically whereas in Feldman's theory the bias is a set which is maintained for a period of time. (ii) My spring element is purely mechanical whereas in Feldman's formulation the spring element includes a spinal reflex element.

This theory postulates that the inverted pendulum is controlled using a series spring whose relative stiffness is less than 100%. It might be questioned whether this is actually possible. Recently, this question has been tested practically by balancing the pendulum using just such a spring. Results to be published shortly demonstrate that this task is perfectly possible.

Tension in the series spring is determined by the length of the spring. The length of the spring is altered by mechanical stretching at the distal end as the footplate moves relative to the tibia and by stretching at the proximal end as the muscle shortens or lengthens.

Sway of the CoM away from the vertical will stretch the spring distally between the footplate and the tibia. If the relative stiffness of the series spring is less than 100%, then the changes in torque produced will not be enough to compensate for the changes in gravitational torque. The length of the spring will have to be increased further by a synchronous movement of the proximal end of the spring. Thus relative to the tibia, the myotendinous junction will on average have to move in anti-phase with the footplate. On average, the muscle contracts as the CoM sways away from the vertical. Thus during balancing, the muscle fibre length changes 180 degrees out of phase with the angle of the CoM. It is perhaps surprising to think that the muscle shortens rather than stretches when I lean forwards.

In this theory, the position of the proximal end of the series spring is controlled by shortening and lengthening of the muscle fibres. The length of the muscle fibres is not controlled by the tension in the spring. This means that the collective intrinsic stiffness of the muscle fibres must be much greater than the tendon. Effectively, the length of the fibres is determined by neural control which makes the fibres contract or which allows them to be lengthened by the tension in the spring. If the proximal end of the spring is not controlled neurally, then the spring cannot be made to lengthen sufficiently and produce sufficient changes in ankle torque to compensate for the gravitational spring.

There is an electromechanical delay between neural modulation of muscle activation and the changes in length of the muscle fibres. On average the muscle fibres need to change length in synchrony with the sway of the CoM and so the neural signals must change in an anticipatory manner to bring about this synchronised change in muscle length. Thus balance of the inverted pendulum by a compliant spring of relative stiffness less than 100% requires anticipatory neural control.

If the proximal end of the spring is moved in a synchronised anti-phase manner compared with the distal end of the spring, then the rate of change of ankle torque with ankle angle will be constant and will be greater than the stiffness of the spring. The speed, of the proximal anti-phase movements relative to the distal movements will determine the rate at which ankle torque changes with respect to ankle angle. (See Figure 3 - 'Effect of dynamic bias on changes of torque with angle'). Faster anti-phase movements of the proximal end will result in a greater rate of change of torque (Δtorque/Δangle). Provided the rate of change of torque with angle is greater than the gravitational spring, the CoM will oscillate at a frequency determined by the effective ankle stiffness (Δtorque/Δangle) and the moment of inertia of the centre of mass. In practice, it is difficult for the neural controller to move the proximal end exactly in synchronous anti-phase to the distal end. If anti-phase movement of the proximal end is slowed down relative to the distal end, then the rate of change of ankle torque with ankle angle will decrease. If the proximal end moves in phase with the distal end then the effective stiffness can pass through zero to become negative. If anti-phase movement of the proximal end is speeded up relative to the distal end, then the rate of change of ankle torque with ankle angle will increase. The exact phasing and timing of the proximal movements will have the effect of throwing the CoM into movement or catching the CoM and bringing it to rest depending on the exact circumstances at the time. These throwing and catching operations can be either intentional or unintentional. These throwing and catching operations appear to be discrete events which are either reactively triggered or intentionally planned. The process would thus be intermittent rather than continuous. The intermittent, throwing effect resulting from the timings of these proximal movements is why I call this a ballistic bias mechanism.

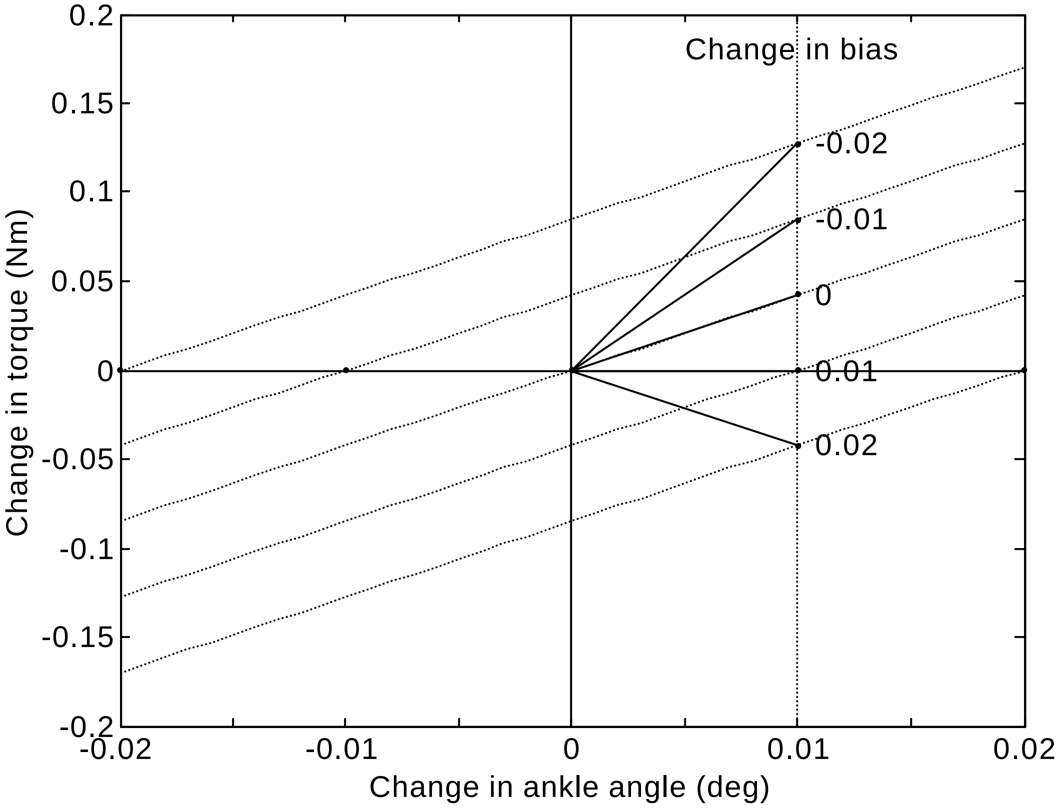

Figure 3. Effect of dynamic bias on changes of torque with angle.
The changes in left ankle torque are shown for a change in ankle angle of 0.01 degrees in a short time interval (e.g. 40 ms). The changes in ankle torque are shown for five different changes in bias position varying from -0.02 to +0.02 degrees. The dotted line through the origin, with zero change in bias position, gives the intrinsic ankle stiffness of 4.2 Nm deg^{-1} which was the value used in Figure 1 - 'Changes in torque and angle'.

Some of the above points can be made more succinctly using mathematics. In this model, the ankle torque $T(t)$ is related to the ankle angle $q(t)$ by the equation

$$T(t) = K\,(q(t) - q_0(t)) \qquad (1)$$

where $q_0(t)$ is the spring bias and K is the constant spring stiffness. By differentiation it can be seen that the rate of change of torque with angle dT/dq is given by

$$dT/dq = K\,(\,1 - \dot{q}_0/\dot{q})$$

where $\dot{q}_0$ is the angular velocity of the bias (proximal end of the spring) and $\dot{q}$ is the angular velocity of the ankle (distal end of the spring). (Movements are defined relative to the tibia). Some results follow:

(1) If the proximal end moves in phase with, and at the same speed as the distal end, $\dot{q}_0 = \dot{q}$, there is no change of torque with angle.

(2) If the proximal end moves in phase with, and faster than the distal end, $\dot{q}_0 > \dot{q}$, then the change in torque with angle are not spring like and the effective stiffness is negative.

(3) If the proximal end moves with sufficient speed in the opposite direction to the distal end the rate of change of torque with angle is greater than the minimum required for marginal stability. The limiting case is given by

$$-\dot{q}_0 \geq \dot{q}\,(mgh/K - 1),$$

where 'mgh' is the minimum required for marginal stability and where K is less than mgh.

(4) If the contractile element shortens while the ankle angle is not changing, $\dot{q} = 0$, the rate of change of torque with angle will be infinite, although the rate of

change of torque with time will be limited by the maximum velocity at which the contractile element can shorten.

From the preceding argument it is clear that the CNS needs to know the velocity of the distal end of the spring or the angular velocity of the ankle in order to plan the movements of the proximal end accurately and effectively. Muscle spindles are extremely sensitive to changes in muscle length. It is perhaps often assumed that they provide accurate information concerning the angle of the ankle joint. In reality, muscle spindles provide information about the length of the muscle fibres and thus the position of the proximal end of the series spring relative to the calf. Given the compliant spring that I am considering, the CNS cannot know the position of the distal end of the spring from proximal position alone. The nervous system needs other information to know the position of the footplate at the distal end of the spring. That other information might come from position sensitive receptors in the ligaments of the ankle joint. Or, the nervous system can estimate the position of the footplate at the distal end of the spring if it knows the stiffness of that spring and the tension in that spring. Thus in combination with past experience (knowledge of the spring stiffness) and golgi tendon organ information (spring tension) the position of the heel can be estimated. The accuracy with which this is estimated will be important in determining the accuracy of the dynamic bias mechanism.

Final conclusions.

For balancing the inverted pendulum and for quiet standing I find that intrinsic, mechanical properties of the ankle joint are on average insufficient to provide marginal stability but they do largely cancel out the destabilising effect of gravity (Loram & Lakie, 2002b, a). I found that short latency stretch reflexes are not elicited in pendulum balancing or quiet standing (Loram & Lakie, 2002a) and this result is consistent with previous investigations (Kelton & Wright, 1949; Gurfinkel *et al.*, 1974; Nashner, 1976). I found that longer latency reactions are apparent in pendulum balancing and standing (Loram & Lakie, 2002a) and that the process of pendulum balancing is neurally modulated (Loram & Lakie, 2002b). I suggest that the inverted pendulum is controlled by dynamic, ballistic-like biasing of a compliant series elastic linkage and that this process requires fine control using anticipation and prediction. I hypothesise that the same process is used in quiet standing.

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