Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application

Maura Casadioa,b, Pietro G. Morassoa,b,*, Vittorio Sanguinetia,b

aDepartment of Informatics, Systems and Telecommunications, University of Genova, Via Opera Pia 13, 16145 Genova, Italy
bCenter of Bioengineering, Hospital La Colletta, Via Giappone 3, 16011 Arenzano, Italy

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Abstract

In this study, we describe a device for the direct measurement of intrinsic ankle stiffness in quiet standing. It consists of a motorised footplate mounted on a force platform. By generating random sequences of step-like disturbances (1° amplitude, 150 ms duration) and measuring the corresponding displacements of the center of pressure in the antero-posterior direction, we obtained torque-rotation patterns after aligning, averaging, and scaling the postural responses. Such patterns were used for estimating the value of the ankle stiffness, which was normalized as a fraction of the critical value. In order to be confident that the measurements addressed the intrinsic ankle stiffness and were not affected in a significant way by the reflex activation of the muscles in response to the test disturbances, we performed the estimates in different ways: least squares estimates with time windows of different widths and an instantaneous estimate at the time in which the angular acceleration vanishes. The statistical analysis showed that there is no significant difference among the different methods of estimate and the inspection of the electromyographic activity in response to the perturbations showed that at least two of the estimates were certainly outside the possible influence of reflex patterns. The intrinsic ankle stiffness was evaluated to be 64 ± 8% of the critical stiffness for test disturbances of the order of 1°. We argue that this figure identifies the lower bound of the range of values which characterise normal sway in quiet standing, whereas the upper bound is given by the estimates performed with much smaller test disturbances [1] which yield a higher value: 91 ± 23%. The two estimation paradigms (with very small and very large test disturbances, respectively) are complementary also because they behave in a different way as regards the sensitivity to a bias torque: it is close to zero in the Loram & Lakie’s paradigm, whereas it is significant in our paradigm. Thus, as the bias grows, it appears that the range of stiffness values is narrowed and is pushed towards the upper bound. There is a clear potential for the clinical application of these methods, in the sense that the identification of the range of stiffness values used by a patient is a measurable index of motor organisation/reorganisation.

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1. Introduction

Estimation of ankle stiffness during quiet standing is crucial in order to understand the fundamental mechanisms of motor control and is also a useful clinical tool for the analysis of the compensatory strategies, adopted by patients in different pathological conditions and adapted during rehabilitation.

Stabilisation of the upright posture is a typical example of many unstable tasks, which must be solved in everyday life and in more demanding sport or dance gestures. These situations are characterised by repulsive forces which tend to push the system away from the intended equilibrium position. Asymptotic stability of this position would be achieved if the task-dependent destabilising torque, which is typically proportional to displacement, were compensated by a stronger restoring torque, generated by the intrinsic stiffness of the muscles and other tissues carrying the load. In this case the neural drive to the muscles could be kept constant, at an appropriate tonic level. On the other hand, an anticipatory active modulation of the neural drive would be necessary if the rate of growth of the restoring torque were weaker than the destabilising torque.
In the case of quiet standing the intended equilibrium position is a slight forward tilt of the body and the instability is gravity-driven. The rate of growth of the toppling torque (i.e., the toppling torque per unit angle coefficient) sets the critical level of stiffness for avoiding the need of neural intervention. If stiffness is beyond the critical level, asymptotic stability is guaranteed without any additional control. Below this level, an active stabilisation mechanism is necessary for compensating the inadequate stiffness and restricting the residual oscillations to a small region surrounding the unstable equilibrium position.

We limit our analysis to the sway movements of the body in the antero-posterior (AP) direction, with the assumption that the body can be simulated by an inverted pendulum oscillating around the ankle with an angle \( \theta \). We set the equality of the toppling and restoring torques in order to find the critical stiffness: 

\[
K_{\text{critical}} = mgh
\]

which clarifies the fact that the critical value of the “restoring force per unit angle” \( K_{\text{critical}} \) is equal to the “toppling torque per unit angle” \( mgh \). It should be noted that both terms of the equation above imply a linearisation: the stiffness coefficient is the first order approximation of the torque-angle characteristics of the ankle muscles and associated elastic tissues; the second member of the equation uses the common approximation \( \theta \approx \sin \theta \). Both approximations are acceptable because the angular range is very small.

Many studies have been carried out over the years on the intrinsic and effective stiffness of the ankle, but only a few were performed while the subjects were standing.

In the study by Winter et al. [2] the “torque disturbance” used for the estimation is the ankle torque itself, measured by a force platform during natural sway movements: \( \tau_{\text{ankle}} = mgu \), where \( u \) is the position of the center of pressure (COP) with respect to the ankle. Sway movements of the body were observed for 10 s, collecting the evolution of \( u(t) \) (from which the ankle torque was derived) and the corresponding COM signals \( y(t) \) (from which the sway angle was derived): the ankle stiffness was then estimated by linear regression of \( \tau_{\text{ankle}} \) versus \( \theta \), and it was found to be on average 8.8% greater than the critical level. One flaw of this method, as remarked by Morasso and Sanguineti [3], is that during the observation time there is no reason to assume that descending motor commands are constant: as a consequence, this method can only provide an overall estimate of the effective ankle stiffness, which comprises the intrinsic mechanical stiffness and the neural stiffness due to short-range stretch reflexes, plus the effect of anticipatory motor commands. By definition, this estimate will be in excess of the critical level, as long as the subjects are able to stand but is unable to say anything about the intrinsic stiffness per se.

The study by Loram and Lakie [1] uses an apparatus which was designed very carefully in order to have a pure estimate of the intrinsic ankle stiffness. The apparatus is based on two footplates. One is fixed and the other is hinged around a horizontal axis, coaxial with the ankle joint; the latter footplate is rotated by means of a piezoelectric actuator which can generate very small, biphasic disturbances (0.055°, 70 ms toes-up +70 ms toes-down) which were chosen in order to perturb as little as possible the underlying sway of the standing body. In fact, the average rotation speed of the disturbance (0.78°/s) is of the same order of magnitude of the average speed of the unperturbed sway. The restoring torque, measured by means of a load cell, was fitted with a mass-spring-dashpot model after aligning and averaging the individual responses. The elastic component was multiplied by 2, to account for the two feet, yielding the following estimate of the intrinsic ankle stiffness: 91 ± 23%, as a fraction of the critical stiffness.

The reflex component of ankle stiffness and the gain of the automatic variation of muscle drive controlling human standing have been studied by Fitzpatrick et al. [4,5], using a weak continuous perturbation applied at waist level to standing subjects. These experiments show that the gain of these reflexes can be altered, thus changing the effective stiffness, but fail to provide a direct estimate of the ankle stiffness because the experimental approach is affected, as in the case of [2], by unaccounted descending motor commands. Moreover, in line of principle we can reject the hypothesis that an unstable system, like the body inverted pendulum, can be stabilised by means of a simple linear control strategy of neural origin. This point is clarified by the block diagram of Fig. 1, in which the inverted pendulum (described by the dynamic equation \( \tau_{\text{ankle}} = I_b \dot{\theta}_b - mgh \dot{h}_b \), where \( I_b \) is the moment of inertia of the body and \( \dot{h}_b \) is the sway angle) is driven by a linear feedback controller. It is easy to demonstrate, by using classical control theory, that if the controller is purely proportional it is impossible to obtain a stable control for any value of the loop gain, even if we neglect the feedback delay. If we add a derivative component

\[
\frac{1}{s^2} G(s) = \frac{K}{s^2}
\]

Fig. 1. Block diagram of the body inverted pendulum with a feedback controller. \( \theta_b \): body angle with respect to the vertical; \( \dot{\theta}_b \): reference body angle; \( I_b \): moment of inertia of the body with respect to the ankle; \( m \): mass of the body; \( g \): acceleration of gravity; \( h \): height of the COM; \( \tau_{\text{ankle}} \): ankle torque; \( s \) is the complex variable used by the Laplace transform. Grey-shaded blocks refer to biomechanics; wave-motive filled blocks refer to control.

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1. \( m \) is the mass of the person, \( g \) is the acceleration of gravity, and \( h \) is the distance of the center of mass (COM) from the ankle.
(PD controller) stability can only be achieved with unrealistically high values of the loop gain, but this result is readily lost by a modest delay.\textsuperscript{2}

The estimates of the intrinsic ankle stiffness by Loram and Lakie \cite{1} must be related to the physiological range of sway movements. It is known indeed that in the COP signals coming from a force platform we can distinguish low-frequency, large-amplitude components and high-frequency, small amplitude components. The former ones are well below 1 Hz and have an amplitude which corresponds to body rotations up to 1\textdegree. The high frequency components of the COP plots (up to 5 Hz) have a much smaller amplitude, which is of the same order of magnitude of the rotation (0.055\textdegree) used in the Loram and Lakie’s apparatus. It is also worth noting that the disturbance torque delivered by that apparatus (about 0.35 Nm for one ankle) is not far from the disturbance generated by the hemodynamics of the heart beat (about 0.2 Nm) which was evaluated by means of synchronised averaging \cite{6} and probably represents the smallest self-generated disturbance of the standing posture. Thus, the range of body sway which must be counteracted by the mechanical properties of the ankle muscles goes from a few hundredths of a degree up to 1\textdegree.

In general, estimates of effective muscle stiffness, as a response to imposed stretch, tend to decrease with the amplitude of the stretch \cite{7}. This suggests that the estimate by Loram and Lakie \cite{1} is close to the upper bound of the value of the ankle stiffness in quiet standing. But we also need a lower bound, related to the high amplitude components of the sway movements, in order to have a complete picture of the role of the mechanical properties of the muscles in the stabilisation of the standing posture. This is the reason for which we designed an apparatus for the direct measurement of intrinsic ankle stiffness which has a clear connection with the apparatus by Loram and Lakie \cite{1} but uses much larger (0.5–1\textdegree) step-like perturbations with a rise time of the same order of magnitude, although a little bit longer for implementation reasons (110–220 ms).

The proposed system was designed with clinical applications in mind, as a supplement of standard posturographic analysis. It has a clinical potential, in addition to the enrichment of the posturographic analysis, and we think that it can be used also as a rehabilitation tool (e.g., to learn corrective or anticipatory responses to predictive disturbances). It is known that in many human actions, learning or re-learning (as in rehabilitation) is accompanied by a relaxation of stiffness and thus a simple device capable to provide a precise estimate of ankle stiffness is likely to be a useful tool in clinical practice.

\begin{table}
\centering
\begin{tabular}{|c|c|}
\hline
Parameter & Value \\
\hline
m & 80 kg \\
h & 1 m \\
\hline
\end{tabular}
\caption{Parameters used in the simulations.}
\end{table}

\textsuperscript{2} Suppose, as a typical example, that \( m = 80 \text{ kg}, \ h = 1 \text{ m} \) and the proportional gain of the PD controller is 10 times bigger than the derivative gain. By using the root-locus technique it is possible to determine that in order to reach a phase margin of 45\textdegree we need a loop gain of 6300. However, the phase margin is totally lost if the feedback delay exceeds 75 ms.

2. Materials and methods

2.1. Apparatus

The apparatus (see Fig. 2) consists of two parts: (i) a force platform, and (ii) a motorised footplate. We used a three-components platform manufactured by RGM SpA, Italy. The platform has a 50 cm \( \times \) 50 cm surface, is built in aluminium, and has four load cells (DS Europe, mod. 546QD). The overall resonant frequency (platform + load cells) is greater than 200 Hz. The frequency bandwidth of the measurement chain (load cells + amplifiers + A/D converter) is greater than 10 Hz. Resolution, linearity, hysteresis, and cross-talk were evaluated and improved by a calibration system which is outlined in Appendix A. A preliminary report of the system is given in \cite{8}. The calibration system was applied to the force platform alone and to the combined system (force platform + motorised footplate). In particular, the resolution in the computation of the COP, which is the most important feature for our purpose, is better than 0.2 mm.

The motorised footplate has two parts: one is bolted onto the force platform; the other can rotate around an horizontal axis whose distance from the footplate is approximately equal to the distance between the ankle and the foot plate. In this way the hinge of the footplate is kept approximately collinear with the subject’s ankle joints. Rotation is provided by a torque motor (M642, MAE SpA, Italy) and a ball screw with a 4 mm pitch of the thread. The motor is connected to the moving part of the footplate by means of a gimbal ring; the ball screw bearing is connected to the fixed part by means of another gimbal ring. In this way the motor axis, which is vertical in the reference position, can self-adjust its orientation during the rotation of the footplate. Footplate rotation is measured by means of a linear variable differential transformer (LVDT Schaevitz mod. E 1000, \( \pm 25.4 \text{ mm} \) measuring stroke, linearity \( \pm 0.5\% \) FS) connected to the base of the footplate through a lever mechanism in order to obtain a suitable angular resolution (0.0037\textdegree) and range of measurement (\( \pm 1\textdegree \)). Calibration of the LVDT in terms of angular rotation of the footplate was carried out directly, by attaching a long and thin rod to the platform and comparing the readout of the sensor with the manually measured chord of arc. The motor is controlled in position by means of a PID module (a Proportional + Integrative + Derivative feedback control block), with a PWM output (Pulse Width Modulation with a carrier frequency of 4 kHz and a sampling rate of 10 kHz).

The control software is based on the Simulink/Real-Time Workshop\textsuperscript{©} packages (Mathworks), and runs under Windows\textsuperscript{©}. Control loop calculations (including the acquisition of the LVDT signal, by means of a 12 bit digital to analog converter-DAC, Advantech mod. 711B) are performed at 2 kHz. The output of the load cells is sampled at 200 Hz, with the same DAC.
2.2. Determination of the perturbation parameters

During unperturbed sway movements, the torque acting at the ankle is given by:

\[ \tau_{\text{ankle}} = mgu \]  
(2)

where \( m \) is the mass of the body and \( u \) the distance of the COP from the ankle in the AP direction (see Fig. 3). In normal subjects, the range of variation of \( u \) is less than 2 cm, and the bandwidth does not exceed 5 Hz [9].

Let us consider a toes-up rotation of the footplate, with amplitude \( \Delta \theta_f \) which causes a dorsiflexion of the ankle: because of the inherent elastic properties of ankle muscles (in particular, the Achilles’ tendon and the gastrocnemius are stretched) and of the feet (the connective tissue of the plantar surface is compressed), the ankle torque is increased. A measurable biomechanical consequence of such an increase, which has nothing to do with neuromuscular control, is the forward shift of the COP, and we can predict its order of magnitude by considering that the change in ankle torque is proportional to ankle rotation, with a proportionality coefficient which is the ankle stiffness, \( K_a \):  

\[ \Delta \tau_{\text{ankle}} = mg \Delta u = K_a \Delta \theta_f \]  
(3)

and therefore:

\[ \Delta u = \frac{K_a \Delta \theta_f}{mg} \]  
(4)

Thus, if we assume that the ankle stiffness is at the critical level, the forward shift of the COP is given by the following expression:  

\[ \Delta u = K_a \frac{\Delta \theta_f}{mg} = mg \Delta \theta_f/mg = h \Delta \theta_f. \]  

For a typical subject (\( h = 90 \) cm) the shift is 1.57 cm/deg. This suggests that rotations greater than 1° with a rise time of 150 ms or less are likely to be unsettling, because they would cause shifts of the COP that exceed the normal range before the brain has a chance to intervene. The suggestion was indeed confirmed during preliminary experiments in which we used perturbations of different sizes and durations. Up to 1.5° the perturbations were not perceived as unsettling, although the average speed of the perturbations was about
10 times bigger than average speed of natural sways, and the subjects had no difficulty, either with open or closed eyes, to keep their balance; beyond 1.5°, with a duration of 150 ms, the situation changed and most subjects had to recover their balance with additional gestures. Therefore, we chose a perturbation size of 1° (from $-0.5^\circ$ to $+0.5^\circ$ with respect to the horizontal and vice versa) and a rise time of 150 ms because this is an upper bound of the disturbances which are tolerated by the postural control system, allowing us to estimate the lower bound of the intrinsic ankle stiffness occurring during postural sway.

Pursuing the analysis of the toes-up disturbance at the end of the rotation, we can say that the predicted forward shift of the COP will automatically induce a backward push of the COM and thus a backward shift of the COP. This means that ankle rotation might be slightly smaller than the imposed rotation of the footplate at the end of the disturbance; the amount depends on the dynamics of the body inverted pendulum, which is characterised by the following equation:

$$I_b\ddot{\theta}_b + K_a(\dot{\theta}_b - \dot{\theta}_f) = mg\sin(\dot{\theta}_b)$$  (5)

If we consider a nominal equilibrium state ($\theta_0 = \theta_f = 0$) and give a toes-up footplate perturbation, the body will start falling backward with a rather small angular acceleration. We are interested in the order of magnitude of the backward rotation at the end of the perturbation and thus we simulated the dynamics of the above system for different values of the relative stiffness (60%, 90%, 120%). Fig. 4 shows that the body rotation at the end of the perturbation is indeed quite small: for the three cases above the rotation, relative to that of the footplate, has the following values: 3.97%, 4.92%, 6.19%. Such small figures were confirmed by a spot check, in which the COM displacement during the perturbation (amplitude 1.5°) was directly measured by using a LVDT (the same model mentioned above): the case of the LVDT was connected horizontally to a vertical wooden board on the back of the subject, at the level of the COM; the moving magnetic core was connected with a thin rod to the waist of the subjects (two: a male and a female). The measured displacement, in the range 2–3 mm, corresponds to 0.11°–0.17° of rotation which is somewhat larger than the expected range (0.059°–0.092°) but of the same order of magnitude.

Summing up, we can say that the expected response to a toes-up disturbance can be characterised by the following phases: (1) a forward shift of the COP synchronous with the footplate rotation, mainly determined by the ankle stiffness, (2) a backward shift of the COP, which is related to the incipient backward fall and thus mainly expresses a biomechanical phenomenon, (3) a rebound of the COP towards a new reference value, which is related to the intervention of reactive motor commands, produced in anticipation of the forthcoming fall. If the test disturbance has an opposite sign, i.e. it is a toes-down rotation, we expect a symmetric response pattern: a backward shift of the COP synchronous with the footplate rotation, followed by a forward shift and then a stabilisation around a new reference value. For the purpose of this paper we shall focus our analysis on the first phase of the response patterns, taking into account that the expected patterns are immersed in the background postural activity, which is a source of “noise” for our measurement and thus needs to be filtered out.

Fig. 5 gives an overview of the experimental situation. The top part recalls the geometry of the system and the

![Fig. 4. Simulation of the body movements (body angle and COP shift) as a consequence of the footplate angular rotation, for different values of the ankle stiffness, as a percentage of the critical value: 60%, 90%, 120%.](image-url)
bottom part the block diagram which includes 3 basic components: (a) the biomechanical part, directly derived from Fig. 1, (b) the measurement system (servomotor + footplate + force platform), (c) the neuromuscular system, which includes (in a simplified manner) stiffness control, segmental reflexes, and descending motor commands (the reference angle $\theta_o$ and the anticipatory “active” torque). The overall dynamics is quite complicated but, if we consider only the first part of the response to the servo-generated footplate disturbance, we can see that the measurement given by the force platform in response to $\theta_f$ is just the sum of the following torque elements: (a) the torque for accelerating the motor and the foot, (b) the torque provided by the block of transfer function $K_a + B_a s$, which expresses that viscous ($B_a$) and elastic ($K_a$) properties of the ankle, (c) the torque determined by the friction of the footplate. The hypothetical intervention of the reflex circuit and the biomechanical feedback due to the incipient fall, as well as the descending active torque only come later on.

Perturbations were generated by filtering a random binary process with a second-order transfer function, in such a way to have a rise time (interval between 10% and 90% of step amplitude) of 150 ms and an amplitude of $1^\circ$.

2.3. Experimental protocol

Subjects were positioned on the footplate, set in the horizontal position, in the typical Romberg position with eyes open, fixating a target at the eyes level about 90 cm in front of them. The servocontroller delivered alternated toes-up and toes-down perturbations at random intervals. Each session included two trials, lasting 200 s each; after the first trial the subjects were asked to step down from the platform and relax a little while. In this way, we could get about 20 artifact-free stimulus–response patterns for both types of stimuli (toes-up and toes-down). Eighteen subjects participated in this experiment.

Two more subjects participated in a pilot experiment, characterised by a similar protocol with the difference that the subjects were required to lean forward during the recording time. The amount of forward shift of the COP was autonomously chosen by the subjects with the requirement that it could be kept “comfortably” for the duration of the trial (200 s) and did not require the heels to loose contact with the support surface.

All the subjects had no record of pathological conditions which might affect their postural stability. They were informed of the purpose of the experiments and gave their
consent according to the regulations of the department. Half the subjects were females and half were males; the age ranged between 21 years and 31 years; weight between 47 kg and 88 kg; height between 1.55 m and 1.88 m; height of the COM (measured manually with respect to anatomical landmarks) between 0.73 m and 0.98 m; critical value of ankle stiffness between 351 Nm/rad and 749 Nm/rad. Table 1 shows the anthropometric data of all the subjects.

### Table 1

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age (years)</th>
<th>Weight m (kg)</th>
<th>Height h (m)</th>
<th>$K_{critical}$ (Nm/rad)</th>
</tr>
</thead>
<tbody>
<tr>
<td>LC</td>
<td>F</td>
<td>25</td>
<td>65</td>
<td>1.70</td>
<td>0.88</td>
</tr>
<tr>
<td>LB</td>
<td>F</td>
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<td>51</td>
<td>1.68</td>
<td>0.87</td>
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<tr>
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<td>60</td>
<td>1.68</td>
<td>0.87</td>
</tr>
<tr>
<td>ES</td>
<td>F</td>
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<td>47</td>
<td>1.58</td>
<td>0.77</td>
</tr>
<tr>
<td>VS</td>
<td>F</td>
<td>22</td>
<td>60</td>
<td>1.68</td>
<td>0.86</td>
</tr>
<tr>
<td>EC</td>
<td>F</td>
<td>24</td>
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<td>1.70</td>
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<tr>
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<td>1.58</td>
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<td>49</td>
<td>1.55</td>
<td>0.73</td>
</tr>
<tr>
<td>VG</td>
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<td>29</td>
<td>77</td>
<td>1.76</td>
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<tr>
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<td>M</td>
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<td>1.81</td>
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<td>AP</td>
<td>M</td>
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<td>73</td>
<td>1.78</td>
<td>0.91</td>
</tr>
<tr>
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<td>M</td>
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<td>74</td>
<td>1.82</td>
<td>0.93</td>
</tr>
<tr>
<td>AS</td>
<td>M</td>
<td>32</td>
<td>72</td>
<td>1.76</td>
<td>0.88</td>
</tr>
<tr>
<td>MG</td>
<td>M</td>
<td>26</td>
<td>77</td>
<td>1.78</td>
<td>0.87</td>
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<tr>
<td>GG</td>
<td>M</td>
<td>25</td>
<td>67</td>
<td>1.73</td>
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<tr>
<td>PR</td>
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<tr>
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<td>M</td>
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<td>81</td>
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<td>0.87</td>
</tr>
<tr>
<td>SR</td>
<td>M</td>
<td>23</td>
<td>78</td>
<td>1.83</td>
<td>0.98</td>
</tr>
</tbody>
</table>

2.4. Data analysis

Load cell recordings were used to estimate the antero-posterior (AP) component of the COP $u(t)$; the LVDT provided the angle of rotation of the footplate $\theta_f(t)$. These data were low-pass filtered by means of a fourth-order Butterworth filter, with a cut-off frequency of 10 Hz. Fig. 6 shows a portion of a typical experimental session (85 s), in which oscillations of the COP are displayed together with the randomised sequence of footplate rotations ($1^\circ$, 150 ms rise-time). It is quite apparent that the disturbance introduced by the platform does not grossly interfere with the underlying sway pattern, which is quickly recovered at the end of each disturbance.

The background sway patterns need to be filtered out in order to estimate the postural response to the test disturbance. On this purpose we carried out a stimulus-locked average of the response signals, under the hypothesis that sway patterns and disturbances are uncorrelated. Fig. 7 gives an example of averaging, over the whole set of 20 perturbations of a trial. As expected, the response pattern appears to be characterised by three subsequent phases: (1) the immediate response of the mechanical ankle impedance to the footplate disturbance, (2) the incipient fall in the opposite direction, (3) the recovery of balance by means of descending commands. In order to estimate the order of magnitude of the residual “background noise”, after averaging, we averaged in the same manner as explained above 20 segments of sway immediately preceding the postural disturbance. The standard deviation of the averaged curve, which can be considered as a residual noise affecting the response pattern, varied between 0.2 mm and 0.4 mm. This measure of uncertainty should be compared with the amplitude of the response which is about 10 mm.

2.5. Estimate of the mechanical impedance

After averaging, we stored stimulus-response pairs: $\delta r(t) = mg[u(t)-u_o]$, $\delta \theta(t) = \theta_f(t) - \theta_o$, where $u_o$ and $\theta_o$ are, respectively, the position of the COP and the orientation of the footplate at the beginning of the perturbation. We then fitted the stimulus–response profiles with a second-order model which represents the response of the mechanical impedance of the ankle:

$$\delta r = I_{tot} \ddot{\theta}_f + B_{tot} \dot{\theta}_f + K_o \delta \theta$$ (6)

![Fig. 6. Illustrative segment (85 s) of the randomised sequence of footplate rotations with the corresponding oscillations of the COP. (Top curve) Footplate rotations angles (amplitude = $1^\circ$; positive values = toes-up). (Bottom curve) Oscillations of the COP in the AP direction (positive values = forward shifts).](image)
where $I_{tot}$ is the total inertia (it includes the footplate, the feet and the motor), and $B_{tot}$ is the viscous friction of the feet and the motor + ball screw; $\dot{\theta}_t$ and $\ddot{\theta}_t$ were computed by an algorithm that approximates the time series of platform angles by means of a third-order polynomial in relation with a balanced moving window of 30 ms which is shifted along the time axis. The identification of the impedance parameters was carried out by means of the standard least squares estimation (LSE) method, with a time window which corresponds to the rise-time of the footplate rotation (150 ms). Fig. 8 shows an example of this evaluation procedure. The measured torque is well approximated by the model response in the first phase of the response, as it should be. They differ afterwards for the intervention of descending motor commands that re-stabilise balance. The first two terms of the impedance ($I_{tot}$, $B_{tot}$) have no physiological meaning and we are only interested in the last one ($K_a$), which is an approximation of the ankle stiffness.

We also considered a second order effect which might affect the estimate. In principle, for the correct estimate of $K_a$ the $d\theta$ angle of Eq. (6) should be computed using the foot rotation angle with respect to the body ($\vartheta_{foot}$), not the footplate rotation angle with respect to the vertical ($\vartheta_{foot}$), because the elastic force elicited from the stretched muscle is determined by the relative not the absolute rotation of the foot. The bias due to this effect is likely to be small because,
as shown in Fig. 4, the two angles are expected to be quite close in the initial part of the response. In any case we can recover the relative foot rotation, given the absolute foot-plate rotation, by considering the following relationships:

\[ \theta_{\text{foot}} = (\theta_f - \theta_b) \]
\[ I_b \ddot{\theta}_b + K_a (\theta_b - \theta_f) = mg \sin(\theta_b), \]

which is the previously considered Eq. (5).

Therefore, it is possible to define the following iterative estimation scheme:

1. Put \( \Delta \theta_{\text{foot}}(t) = \Delta \theta_f(t) \).
2. Estimate \( K_a \) by using Eq. (6).
3. Integrate Eq. (5) and compute body rotation, \( \Delta \theta_b(t) \).
4. Put \( \Delta \theta_{\text{foot}}(t) = \Delta \theta_f(t) - \Delta \theta_b(t) \); go to 2.
5. Stop when two subsequent estimates of \( K_a \) are sufficiently close (1%).

In practice one or two iterations were sufficient to meet the criterion.

2.6. Verification of the equivalence of ankle estimates in the toes-up vs. toes-down situations

For each subject we obtained two estimates of the ankle stiffness, one in the toes-up and the other in the toes-down situation. Since the muscle reflexes at the human ankle joint are known to be quite asymmetric [10] we checked if such asymmetry affected our measurement. This was done by means of a two-way ANOVA, taking into accounts two factors: direction of the perturbation and gender. The latter factor was considered because the anthropometric parameters of males and females in our population were significantly different.

2.7. Alternative estimates of the ankle stiffness

In addition to the estimation procedure described above, we also used two alternative procedures which tend to minimise the potential influence of short-latency reflexes:

1. A least squares estimate of the impedance on a smaller time windows (70 ms instead of 150 ms).
2. An instantaneous estimate of the ankle stiffness at the time in which the angular acceleration crosses the zero line. At that time the inertial component of the ankle torque vanishes and Eq. (6) is reduced to the following form:

\[ \delta \tau = B_{\text{tot}} \dot{\theta}_f + K_a \delta \theta. \]

Therefore, \( K_a \) can be estimated, at the zero-crossing time, by subtracting the viscous term from the ankle torque and dividing the result by the angular rotation. It should be noted that at that time the angular speed is at its peak value by definition.

2.8. Direct verification of the stretch reflex

It has been previously explained why segmental reflexes are likely to contribute little to the stabilisation
of unperturbed posture and why we expect them to be scarcely important in determining the response to postural perturbations which have an amplitude similar to normal sway. As a verification of this assumption, we collected the electromyographic activity of three muscles (tibialis anterior, medial and lateral gastrocnemius) in two subjects before, during and after the perturbation. We used the StepPC system, by DEM SAS. A variable geometry preamplifier was used for each muscle (gain: 1050; lower cut-off frequency: 30 Hz; higher cutoff frequency: 5 kHz; input impedance: 10 GΩ/5pF; input noise: 12 μV; CMRR: 96dB) and was connected to the 16 channels patient’s portable unit (low-pass filter: 500 Hz; variable gain: 1–50; A/D converter: 12 bit; sampling rate: 2 kHz). After sampling, the signals were filtered (band pass between 35 Hz and 350 Hz, with a notch filter at 50 Hz), rectified and integrated (RMS filter with a 60 ms time window), aligned with respect to the disturbances and averaged over a trial.

2.9. Influence of bias torque on the estimate of ankle stiffness

In our experimental set-up it is not possible to mechanically impose a bias torque, as is done in different types of ergometers. However, it is possible to ask the subjects to voluntarily shift forward or backward the average value of the COP with respect to the spontaneously chosen position, which is usually about 2.5 cm–4.5 cm in front of the ankle. Thus, shifting forward the COP by 4 cm–5 cm is equivalent to multiply by 2–3 the natural bias torque. Two subjects were trained to shift their COP forward and keep it stable for the duration of the experiment (200 s): a male (age = 58 years, weight = 74 kg, height = 1.78 m, $K_{\text{critical}} = 651 \text{ Nm/rad}$) and a female (age = 28 years, weight = 60 kg, height = 1.68 m, $K_{\text{critical}} = 512 \text{ Nm/rad}$). Instead of using a direct biofeedback of the COP position we simply asked the two subjects to choose a “comfortable” forward shift, without loosing the heel contact with the ground. It takes a few trials before this can be achieved in a satisfactory manner, both in the unperturbed and perturbed situations.

2.10. Validation of the measurement system

A validation of the overall measurement system was carried out by using an inverted pendulum bolted onto to the footplate and stabilised by means of two strong springs (Fig. 9). In a preliminary experiment we estimated the angular stiffness of the pendulum from basic physics. The pendulum was manually put in motion while the footplate was fixed. From the observed damped oscillations (Fig. 10(top)) we computed the damping coefficient $\zeta$ and the damped angular frequency $\omega_d$. The oscillations are described by the following equation: $I \ddot{\varphi} + B \dot{\varphi} + (K_{\text{spring}} - mgh) \varphi = 0$ from which we can derive the following expression for the damped frequency: $\omega_d = \sqrt{(1 - \zeta^2)/(K_{\text{spring}} - mgh)/I}$. In the experimental setup $m = 12.61 \text{ kg}, h = 0.83 \text{ m}, I = 7.23 \text{ kg m}^2$. Thus, we can derive from the previous expression an estimate of the “true” value of the angular stiffness: $K_a = 185.17 \text{ Nm/rad}$.

In a second type of experiment the inverted pendulum was perturbed according to the same protocol used for the human subjects. Fig. 10(bottom) shows the average response curve after a trial (each trial includes eight repetitions of the footplate disturbance). Twenty trials were carried out in order to evaluate the stability of the estimate. The average value is: $K_a = 183.4 \pm 5.85 \text{ Nm/rad}$. Thus, we can say that the intrinsic error of the measurement system of ankle stiffness is less than 5%, which is quite acceptable.

3. Results

In all the subjects the disturbance introduced by the platform, although clearly perceived, did not disrupt the background sway patterns. The response is stereotypically characterised by three distinct phases: (i) an immediate
mechanical response in the same direction of the disturbance (forward shift of the COP for a toes-up disturbance and backward shift of the COP for a toes-down disturbance) related to the mechanical impedance of the ankle; (ii) a biomechanical response in the opposite direction which corresponds to the incipient backward fall of the body; and (iii) a voluntary neural response which recovers the dynamic stability characteristic of quiet standing. The EMG patterns recorded for two subjects (Fig. 11 shows the averaged response of one of them for the toes-up disturbance) confirm that the stretched muscles exhibit a hardly noticeable variation of activity and the peak of activity of the antagonist muscle (the tibialis anterior in the case of the figure) occurs in the final part of the disturbance and in any case after the time of zero acceleration as well as outside the smaller time window (70 ms) in which the LSE was carried out.

“mechanical” response in the same direction of the disturbance (forward shift of the COP for a toes-up disturbance and backward shift of the COP for a toes-down disturbance) related to the mechanical impedance of the ankle; (ii) a biomechanical response in the opposite direction which corresponds to the incipient backward fall of the body; and (iii) a voluntary neural response which recovers the dynamic stability characteristic of quiet standing. The EMG patterns recorded for two subjects (Fig. 11 shows the averaged response of one of them for the toes-up disturbance) confirm that the stretched muscles exhibit a hardly noticeable variation of activity and the peak of activity of the antagonist muscle (the tibialis anterior in the case of the figure) occurs in the final part of the disturbance and in any case after the time of zero acceleration as well as outside the smaller time window (70 ms) in which the LSE was carried out.

The average error of the reconstructed torque (RMS value over the estimation window) was very small: better than 0.015 Nm for all the subjects, disturbance directions and widths of the time window. The mechanical response of the ankle to the imposed disturbance is dominated by the elastic component: within the frequency range typical of sway movements the elastic torque is 5–10 times greater than the viscous and inertial torques, in spite of the fact that the inertia of the footplate + motor + screw is added to the inertia of the foot. The inertial parameter of the impedance is the most stable one (0.45 ± 0.07 kg m², with a coefficient of variation of 15%) as one might expect because the part of the combined moment of inertia which depends on the foot and thus changes in the population of subjects is clearly minor. The viscous parameter is the least stable one: 5.81 ± 3.28 Nm/(rad s), with a coefficient of variation of 56%. Remarkably, our estimate is quite close to the range of

Fig. 10. (Top curve) Unforced oscillations of the inverted pendulum. (Middle curves) Angular rotation, velocity and acceleration of the footplate imposed by the servomotor. (Bottom curves) Measured torque (obtained from the force platform measurement by scaling with a coefficient equal to mg) and reconstructed torque (dotted line), obtained from the estimated mechanical impedance. The vertical lines identify the time window used by the estimation procedure.
values found by Loram and Lakie [1] with a completely different setup (between 3.7 Nm/(rad s) and 4.8 Nm/(rad s)). The elastic parameter or stiffness, which was the main subject of this paper, was always below the critical level: 366.74 ± 99.58 Nm/rad, with a coefficient of variation of 27% which is very similar to the coefficient of variation of the critical values of ankle stiffness for the population of subjects analysed in this study (22%). Therefore, the relative ankle stiffness (the ratio between the measured ankle stiffness and the critical stiffness) was always less than 1 (0.64 ± 0.078). The impedance parameters are listed in Table 2.

The first statistical test involved the stiffness estimates in the two directions (up = dorsal flexion versus down = plantar flexion of the ankle), also taking into account the possible influence of gender. On this purpose we ran a two-way ANOVA (factors: gender, direction) over the estimated stiffness, normalized to its critical value. We found no significant effects of gender (female: 0.608, male: 0.672, \(F(1, 16) = 3.43, P = 0.082\)) and direction (UP: 0.649, DOWN: 0.63, \(F(1, 16) = 1.96, P = 0.18\)):

<table>
<thead>
<tr>
<th>Subject</th>
<th>(I_{tot}) (kg.m²)</th>
<th>(B_{tot}) (Nm/(rad.s))</th>
<th>(K_3) (Nm/rad)</th>
<th>(K_{rel}^*)</th>
<th>(K_{rel}^{**})</th>
<th>(K_{rel}^{***})</th>
</tr>
</thead>
<tbody>
<tr>
<td>LC</td>
<td>0.52</td>
<td>3.79</td>
<td>318.40</td>
<td>0.567</td>
<td>0.583</td>
<td>0.559</td>
</tr>
<tr>
<td>LB</td>
<td>0.47</td>
<td>4.53</td>
<td>281.45</td>
<td>0.647</td>
<td>0.673</td>
<td>0.678</td>
</tr>
<tr>
<td>MC</td>
<td>0.47</td>
<td>4.79</td>
<td>231.67</td>
<td>0.452</td>
<td>0.476</td>
<td>0.482</td>
</tr>
<tr>
<td>ES</td>
<td>0.49</td>
<td>0.78</td>
<td>196.74</td>
<td>0.554</td>
<td>0.578</td>
<td>0.619</td>
</tr>
<tr>
<td>VS</td>
<td>0.47</td>
<td>2.89</td>
<td>291.04</td>
<td>0.575</td>
<td>0.605</td>
<td>0.568</td>
</tr>
<tr>
<td>EC</td>
<td>0.52</td>
<td>0.65</td>
<td>312.02</td>
<td>0.638</td>
<td>0.693</td>
<td>0.634</td>
</tr>
<tr>
<td>LT</td>
<td>0.44</td>
<td>2.45</td>
<td>310.56</td>
<td>0.716</td>
<td>0.741</td>
<td>0.726</td>
</tr>
<tr>
<td>GO</td>
<td>0.30</td>
<td>4.26</td>
<td>241.97</td>
<td>0.690</td>
<td>0.759</td>
<td>0.680</td>
</tr>
<tr>
<td>VG</td>
<td>0.45</td>
<td>5.80</td>
<td>416.17</td>
<td>0.633</td>
<td>0.636</td>
<td>0.656</td>
</tr>
<tr>
<td>TA</td>
<td>0.43</td>
<td>10.27</td>
<td>473.44</td>
<td>0.590</td>
<td>0.630</td>
<td>0.598</td>
</tr>
<tr>
<td>AP</td>
<td>0.46</td>
<td>4.18</td>
<td>413.22</td>
<td>0.634</td>
<td>0.633</td>
<td>0.664</td>
</tr>
<tr>
<td>EB</td>
<td>0.39</td>
<td>10.01</td>
<td>411.45</td>
<td>0.609</td>
<td>0.641</td>
<td>0.615</td>
</tr>
<tr>
<td>AS</td>
<td>0.47</td>
<td>10.66</td>
<td>428.80</td>
<td>0.690</td>
<td>0.705</td>
<td>0.616</td>
</tr>
<tr>
<td>MG</td>
<td>0.37</td>
<td>9.67</td>
<td>444.61</td>
<td>0.677</td>
<td>0.613</td>
<td>0.691</td>
</tr>
<tr>
<td>GG</td>
<td>0.43</td>
<td>7.21</td>
<td>434.65</td>
<td>0.769</td>
<td>0.805</td>
<td>0.747</td>
</tr>
<tr>
<td>PR</td>
<td>0.58</td>
<td>6.27</td>
<td>341.76</td>
<td>0.616</td>
<td>0.522</td>
<td>0.659</td>
</tr>
<tr>
<td>AC</td>
<td>0.55</td>
<td>5.99</td>
<td>532.72</td>
<td>0.771</td>
<td>0.764</td>
<td>0.771</td>
</tr>
<tr>
<td>SR</td>
<td>0.40</td>
<td>10.44</td>
<td>520.62</td>
<td>0.694</td>
<td>0.725</td>
<td>0.693</td>
</tr>
</tbody>
</table>

\(K_{rel}^*\): relative stiffness, as a fraction of the critical stiffness (see Table 1), computed in three different modes; (1) \(K_{rel}^*\) is computed over a 150 ms time window; (2) \(K_{rel}^{**}\) is computed over a 70 ms time window; (3) \(K_{rel}^{***}\) is an instantaneous estimate computed by taking into account only the measurement of the torque and the angular velocity at the time of zero angular acceleration. The displayed values are the averages of the toes-up and toes-down experiments.

This result should be interpreted in the framework of the study by Stein and Kearney [10] who analysed the nonlinear behaviour of the ankle reflexes evoked by sharp disturbances (0.3–5°, 50 ms). Two main results from that study are relevant for our purposes: (1) the reflex is strongly asymmetric, in the sense that it is much larger for dorsal flexion than plantar flexion (for a 1° disturbance the reflex torque in dorsal flexion is about 5 Nm and close to 0 in plantar flexion); (2) the peak of the reflex torque occurs about 100 ms after the peak of the EMG. In view of these findings, the equivalence of the stiffness estimates in the two directions suggests that short latency reflexes are not likely to affect the estimates in a significant way.

In order to further support the hypothesis that what we estimated was indeed the intrinsic ankle stiffness we performed two more estimates. One estimate used the LSE method with a smaller time window (70 ms instead of 150 ms) which was located symmetrically with respect to the time of zero acceleration. This time instant occurred 66.2 ± 2.7 ms after the initiation of the disturbance. The average value of the estimate was quite close to the other estimate with a larger time window and the corresponding standard deviation was only slightly bigger: 0.655 ± 0.087 (relative stiffness) instead of 0.640 ± 0.078. The third stiffness estimate was instantaneous and was carried out at the time of zero acceleration, by subtracting from the total torque (which on average was 4.33 Nm) the viscous torque (which on average was 1.15 Nm) and dividing by the angular rotation at that time instant (0.45°, on average). At the time of the estimate the viscous torque is a significant portion of the total torque, in spite of the small size of the \(B\) coefficient, because the computation is performed at the time of peak angular velocity (on average 10.78°/s). Also in this case we
found relative stiffness estimates that were quite close to the previous ones: 0.648 ± 0.070. In order to test the effects of the three methods of calculating the stiffness we ran a two-way ANOVA (factors: method, gender) over the last three columns of Table 2. We found no significant effect of gender \((F(1, 16) = 2.03, P = 0.17)\) or method \((F(2, 32) = 1.07, P = 0.35)\). Post hoc analysis (Newman–Keuls test) revealed no statistically significant differences between each pair of estimation methods.

Summing up, we think that 0.64 ± 0.078 is a good estimate of the intrinsic relative ankle stiffness during standing for a disturbance size of the order of 1°. Fig. 12 shows the distribution of the relative stiffness estimates as a function of the critical stiffness values.

The influence of a bias torque on the stiffness estimate was tested by asking two subjects (a male and a female) to voluntarily shift their COP forward and keep it stable during the duration of the experiment:

1. In the male subject the average forward shift of the COP was 5.4 cm; this determined an increase of the bias torque from 34.9 Nm to 74.1 Nm and an increase of the ankle stiffness from 392.2 Nm/rad to 562.7 Nm/rad, which is still smaller than the critical value (651 Nm/rad).

2. In the female subject the average forward shift of the COP was 4.5 cm; this determined an increase of the bias torque from 26.8 to 53.3 Nm and an increase of the ankle stiffness from 234.3 Nm/rad to 446.6 Nm/rad, which again is smaller than the critical value (512 Nm/rad).

These results should be compared with the measurement of the series elastic stiffness in the human plantar flexors performed by de Zee and Voigt [11] by means of the quick release method. At values of the bias torque similar to our subjects in the normal standing posture, it is possible to evaluate from those data that the series elastic stiffness, relative to the critical stiffness, is about 0.9. This is significantly larger than our estimate of intrinsic stiffness (0.64) but very close to the estimate of Loram and Lakie [1]. If the bias torque is increased as described above the series elastic stiffness, according to de Zee and Voigt [11], increases by about 40%. An increase of the same order of magnitude is detected by our method. On the contrary, the method described by Loram and Lakie [1] is rather insensitive to the same increase of bias torque.

### 4. Discussion

In agreement with the results of Loram and Lakie [1] we found that the intrinsic ankle stiffness during quiet standing is consistently below the critical value and thus the stabilisation of the human inverted pendulum requires active neural control. However, the two estimates differ as regards the relative stiffness values and the dependence on bias torque:

1. In Loram and Lakies’s study the stiffness is close to the critical level (0.91 ± 0.23) and is rather independent of the bias torque.

2. In our study the stiffness is smaller (0.64 ± 0.078) and is dependent on the bias torque.

We attribute the difference of the estimates to the fact that they use test disturbances with quite different amplitudes \((0.055° \text{ versus } 1°)\) and this is consistent with the fact in general the measured ankle stiffness tends to decrease as the amplitude of the test disturbance increases [7].

The Loram and Lakie’s estimate can be considered as an upper bound because the amplitude of their test disturbance is of the same order of magnitude of the smallest sways. Probably it mainly reflects the stiffness of the series elastic element (SEE) of the plantar flexors. The fact that this estimate method is very weakly dependent on bias torque, in spite of the known increase of the SEE stiffness with bias, might be due to nonlinear effects in the muscle mechanics which tend to saturate the rate of increase of the elastic force. In any case, this further confirms that even a
substantial amount of bias force, compatible with the range of values which characterise the standing posture, is unable to elicit intrinsic stiffness values that go beyond critical stiffness.

The estimate of ankle stiffness obtained by means of our apparatus is a lower bound of the physiological range of stiffness values that characterise upright standing because test disturbances with larger amplitudes and equal or smaller durations destabilise the sway pattern and force the subjects to use emergency actions for recovering balance. Although the test disturbance is fast enough to evoke short-latency reflex activation of the muscles, which occurs towards the end of the disturbance, its influence on the estimated stiffness value is likely to be minimal: this conclusion is supported by the absence of statistically significant differences in three estimation methods that attribute a different weight to the part of the response and by the fact that the estimates are symmetric in relation with the disturbance direction (dorsiflexion versus plantar flexion) in spite of the known asymmetry of the muscle reflexes at the human ankle joint. The fact that the estimated stiffness is smaller than the SEE stiffness is a consequence of the fact that the SEE is mechanically connected in series with the contractile element (CE) and this means that total stiffness $K_a$, which is given by the following relationship $1/K_a = 1/K_{\text{SEE}} + 1/K_{\text{CE}}$, is smaller that the smallest of the two stiffness values.

In our experiments we showed that the stiffness increased as the subjects leaned forward. We attribute this increase to an increase of either $K_{\text{SEE}}$ or $K_{\text{CE}}$ or both. Our measurements cannot discriminate among these possibilities. The stiffness of muscle and tendon are both known to increase with tension, so this provides a plausible mechanism. The strategy of increasing tension and stiffness by leaning forward is one that is quite unlikely to be used by the nervous system to promote stability but we suggest that the same effect may be produced by coactivation of muscles acting around the joint, what may be called a “stiffness strategy”. It is well known that the nervous system cannot change the resistance to stretch independent of the level of activation and thus a stiffness strategy requires coactivation to increase the level of muscle activity-and therefore stiffness-without increasing the net torque around the ankle. For example, Carpenter et al. [12] showed an enhancement of ankle stiffness in the case of “postural threat” (a manipulation of environmental conditions). Ho and Bendrups [13] found that increased ankle stiffness characterises elderly people with a tendency to fall, and Shadmehr [14] argued for a general role of stiffness in the control of posture. In summary, the estimate of ankle stiffness provided by the present paper complements the estimate provided by Loram and Lakie [1] suggesting that during normal sways the intrinsic ankle stiffness varies in the range 0.64–0.91, relative to the critical value, and this range is narrowed towards the upper bound if there is a torque bias.

A consequence of the insufficient value of the ankle stiffness is that the neural control has an anticipatory, integrative nature which may be characterised by a discontinuous, “saccadic-like” pattern [2,15,16], supplementing what is provided “for free” by the mechanical properties of muscles. However, anticipatory postural stabilisation is a rather sophisticated computational task whose efficiency depends critically upon a number of central and peripheral factors. Thus, as soon as the efficiency of the anticipatory control mechanism starts deteriorating for any reason, e.g. as an effect of aging, the possibility of increasing the role of stiffness control, by means of coactivation of the ankle muscles, becomes a viable alternative. Our data suggest that the margin of ankle stiffness modulation is substantial, although short of the critical level. In that case, a synergetic action is provided by “unlocking” the hip joint and translating the focus of attention from a partially locked ankle to the hip joint.

Acknowledgments

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Appendix A. Calibration system of the force platform

The calibration system consist of a ballast of mass $M$ (it is a perspex cylinder filled with small lead balls, with a total weight of 35 kg) and a smaller eccentric mass $m$ of 2 kg, rotating with respect to a vertical axis centered on the axis of symmetry of the ballast, with a ball bearing which allows low-friction rotation.

The center of pressure of the device depends on the mass $M$ of the ballast, the mass $m$ of the eccentric element, the corresponding instantaneous angular velocity $\omega(t)$, and geometric parameters of the set-up. The COP trajectory can be determined with great accuracy because the uncertainty about the radial position of the eccentric mass, which may be less than 1 mm, is further reduced by the scale factor $m/(M + m)$: this parameter allows to obtain a resolution of the order of 0.1 mm even with rather crude construction techniques.

It is easy to demonstrate that if we set manually in motion the rotating mass, the theoretical trajectory of the COP is a spiral, asymptotically approaching the circle of radius $R(m/(m + M))$ where $R$ is the distance of the eccentric mass from the rotation axis. This spiral, measured by the force platform, is the training set used by the verification/recalibration software. The variability induced by the human action can be
eliminated by specifying a standard rotation speed $\omega$ from which to trigger the storage of the "training data". We chose $\omega = 0.51$ Hz because in this case (considering that the height of the eccentric mass from the platform is 38 cm) it turns out that the area occupied by the spiral is contained inside a circle of radius 40.14 mm: this area covers the range of the posturographic traces plus a plausible area of variation in the positioning of the subjects. With such data it is possible to verify the linearity, the cross-talk among the AP and ML channels, the spatial resolution and other interesting indicators. In particular, the cross-talk can be estimated by computing the covariance matrix of the training data and extracting the corresponding eigenvalues. In the ideal situation they should be equal and thus the cross-talk can be evaluated by the following index: $\gamma = 1 - \frac{\lambda_{\text{min}}}{\lambda_{\text{max}}}$. An iterative procedure has been implemented which computes a calibration matrix, to be applied to the training set, with the purpose of minimising $\gamma$. The final value of the calibration matrix is stored and applied to the collected platform data. At the end of the calibration procedure it is possible to estimate different indicators. The most important one, for our purpose, is the resolution, defined as the standard deviation of the discrepancy between the computed and expected COP position, over the training set. It turns out that the resolution of the COP estimate with the used platform is better than 0.2 mm. This is acceptable if we consider that the amplitude of the COP response is about 5 mm.

References