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

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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.

Keywords: Posture; Ankle stiffness; Force platform; Stability

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.

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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 ϑ_b. We set the equality of the toppling and restoring torques in order to find the critical stiffness: $K_{critical}\vartheta_b = mgh\vartheta_b$.[1] From this we get the following relationship:

$$K_{critical} = mgh \quad (1)$$

which clarifies the fact that the critical value of the "restoring force per unit angle" $K_{critical}$ 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 $\vartheta \approx \sin\vartheta$. 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: $\tau_{ankle} = mgu$, 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 τ_{ankle} versus ϑ_b, 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 un 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 $\tau_{ankle} = I_b\ddot{\vartheta}_b - mgh\vartheta_b$, where I_b is the moment of inertia of the body and ϑ_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

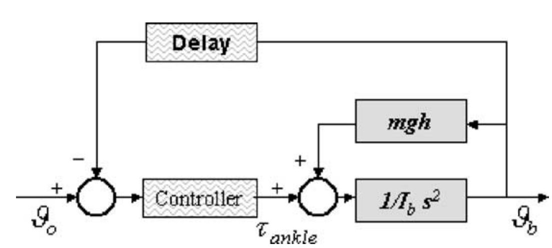

Fig. 1. Block diagram of the body inverted pendulum with a feedback controller. ϑ_b: body angle with respect to the vertical; ϑ_o: reference body angle; I_b: moment of inertia of the body with respect to the ankle; m: mass of the body; g: acceleration of gravity; h: height of the COM; τ_{ankle}: ankle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

[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.