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MOTOR CONTROL MECHANISMS IN UNSTABLE TASKS

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Abstract – The motor control mechanisms adopted by humans when carrying out unstable tasks are investigated in relation with 2 specific paradigms: A) Stabilisation of the standing posture; B) Stabilisation of an inverted pendulum, grasped at different heights. It is shown that stiffness control is present but is insufficient and predictive control mechanisms are filling the gap.
Keywords: Human functions, Motor control; Stiffness.

1. INTRODUCTION

There has been interest in recent years in the motor control of unstable tasks, particularly in two specific paradigms: 1) stabilisation of the human inverted pendulum in quiet standing [1, 2, 3, 4]; 2) arm trajectory formation in a divergent force field [5]. In both cases the instability is associated with a potential field which has a maximum value around the reference state and feeds a repelling force field. This implies a *critical level of muscle stiffness* which may counterbalance the slope of the repelling field. In fact there are 3 possible mechanisms of stabilisation:

A) **Reflex Mechanism**, determined by a number of different sensory feedbacks;
B) **Stiffness mechanism**, related to the mechanical properties of muscles: it operates *without delay,* and can be modulated by means of coactivation of antagonistic muscles;
C) **Anticipatory feedforward mechanism**, which has an integrative central nature: it is based on some kind of *internal model* for multi-sensory fusion, thus compensating by means of prediction the transduction and propagation delays of sensory information.

The contribution of the first mechanism can be ruled out because the *significant delays* in the propagation of nervous signals are incompatible with the instability of the plant from the point of view of the stability of the control system. Thus the last two mechanism must share the burden of stabilisation and the open question is whether one or the other is predominant and under which circumstances.

2. STABILISATION OF THE STANDING POSTURE

In the case of the standing posture, it can be shown that the critical level of muscle stiffness, measured at the ankle joint, is mgh where m is the mass of the body, g is the acceleration of gravity, and h is the height of the bodily centre of gravity. Beyond that level muscle stiffness alone is sufficient to stabilise the standing body without any need of persistent active control. The opposite is true if stiffness is below that level. In the literature there is a wide range of estimates. Winter et al [1] estimated that the ankle stiffness is 108.8% the critical level, by observing unperturbed natural sway with a long time window (tens of seconds). Morasso and Sanguineti [2] argued that the estimation method was intrinsically wrong because it did not take into account the modulation of muscle activity during the measurement time. Loram and Lakie [4] measured the response to very small and very quick perturbations (0.055 deg, 70 ms) and found that ankle stiffness is under-critical: on average 91% of the critical level. In our study we used larger and slower perturbations (0.5-1 deg, 200-300 ms) which are similar to the natural sway patterns and thus are likely to be compensated in a similar way.

2.2 Methods: apparatus for the direct measurement of ankle stiffness

Figure 1 shows the prototype of the apparatus, which consists of two parts: 1) a force platform, 2) a motorised footplate. The force platform (3-components by RGM spa, with a 50 cm x 50 cm surface, 4 load cells) has a resonant frequency exceeding 200 Hz, a resolution in the computation of the COP (Center of Pressure) better than 0.2 mm and a frequency bandwidth of the overall measuring chain larger than 10 Hz. The motorised footplate has a hinge which is used as a reference point for positioning the ankle joints of the subject. The rotation is provided by a brushless DC motor and a ball screw. The motor is connected to the structure by means two gimbal rings in order to allow the self-adjustment of the motor axis which is approximately vertical. The footplate rotation is measured by means of a

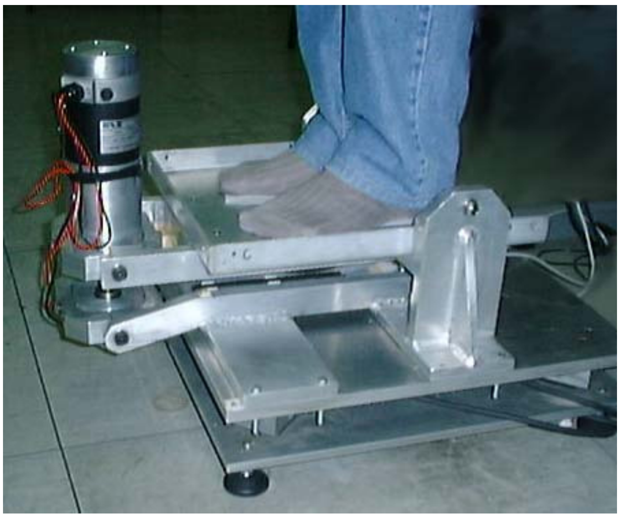


Fig. 1

LVDT (linear variable differential transformer) connected through a lever mechanism to the base in order to obtain a good angular resolution and range of measurement. The motor controller is a standard PID and has been implemented in Matlab ©.

Figure 2 shows the randomised sequence of footplate disturbances and the AP (Antero-Posterior) oscillations of the COP measured by the force platform.

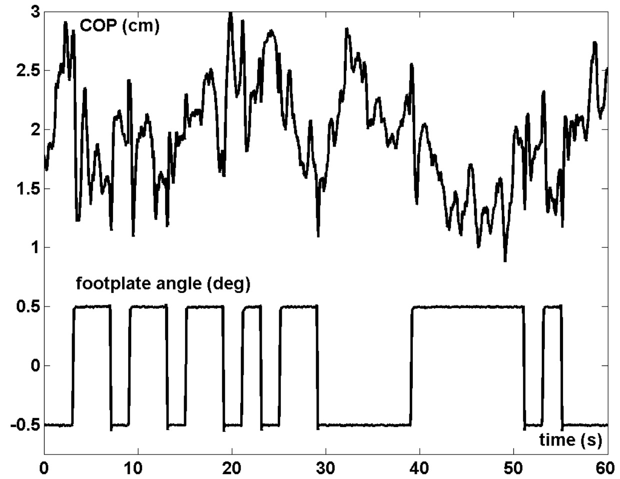


Figure 2.

The COP curve is the combination of the natural oscillations and the responses to the disturbances. In order to filter out the former component we performed a stimulus-locked average of the COP signal, on the hypothesis that this signal is independent of the stimulus.

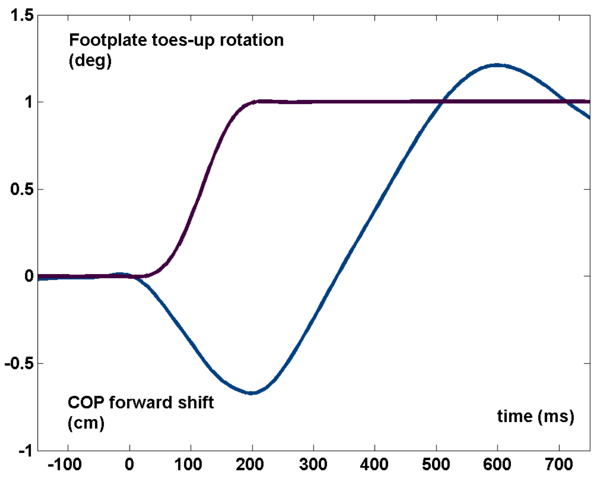


Fig. 3: ankle rotation $\theta(t)$ and COP displacement $y(t)$.

In figure 3 we show the average of the toes-up rotations in a typical subject. Inverted but equivalent patterns were found in the toes-down case. The initial part of the response is a forward shift of the COP determined by the process of "loading" the ankle spring. During that time (less than 200 ms) the COM (Center of Mass) moves by a very small amount which can be neglected as a first approximation. Then the body starts falling backward before being stabilised again by the control mechanism. By multiplying the COP position by mg we get the ankle torque τ_a and then we can fit the stimulus-response patterns with the following model

$$\tau_a = I\ddot{\vartheta} + B\dot{\vartheta} + K_a\vartheta$$

where I is the moment of inertia of the footplate + the foot, B is the viscous coefficient of the motor and the ankle, and K_a is what we are looking for: the ankle stiffness.

2.2 Results

The experiments were carried out with 4 subjects. In all of them the estimated ankle stiffness was markedly under-critical. On average it was 60,29% of the critical value, also taking into account that during the footplate rotation (1 deg) the rotation of the upper body was evaluated to be about 0.04 deg. So the qualitative result of Loram and Lakie [4] is confirmed but our estimate is significantly smaller. The reason of the difference, in our opinion, is that using a quick and small disturbances, as Loram and Lakie did, may emphasise the short-range stiffness and cause an over-estimate in relation with the physiological level of stiffness.

3. MANUAL STABILISATION OF UNSTABLE PLANTS

Experiments have been performed by Burdet et al [5] in which the subjects had to perform reaching movements, from an initial point to a target, which were disturbed by a divergent force field generated by a computer-controlled robotic arm. The force field was null along the nominal trajectory and was orthogonal to the trajectory outside it, with an intensity which grew linearly with the distance from such trajectory. This experimental situation simulated a sort of unstable "ridge", characterised by the rate of growth of the field. The slope of the field was about 200-300 N/m, which is within the physiological range of hand stiffness [6]. The subjects had no difficulty to solve the task after a sufficient number of repetitions, in the sense of generating trajectories very close to the nominal one, thus minimising the interaction with the divergent force field. The analysis of the EMG patterns and of the stiffness ellipse demonstrated that the adaptation to the field goal was achieved by means of a re-arrangement of the muscle activities in such a way to orient the ellipse in the direction of the field, together with a suitable level of coactivation, thus suggesting that modulation of muscle stiffness was the chosen mechanism of stabilisation in this case.

In order to get a better insight into the interplay between stiffness stabilisation and anticipatory feedforward stabilisation, an experimental setup was developed in which the subjects were required to stabilise an inverted pendulum by grasping it at different heights.

3.2 Methods: the instrumented pendulum

The pendulum, which is hinged on a low-friction ceramic joint allowing 3 degrees of freedom, has a mass m of 10 Kg, a height h of 1.8 m and the grasping distance d can be varied between 30 and 80 cm. The small oscillations of the pendulum are detected by means a mechanical linkage and a pair of precision potentiometers. Since the gravity torque is $\tau_g = mgh\theta$ and the muscle torque is

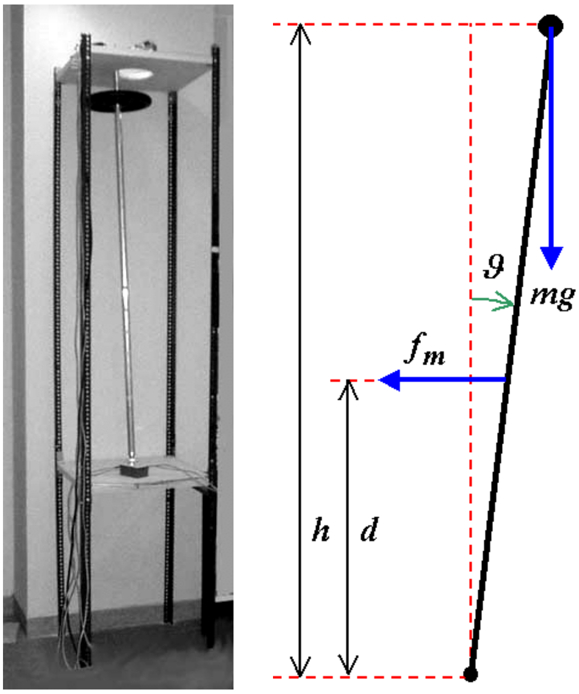


Fig. 4

$\tau_m = -f_m \cdot d = K_m \cdot (d \cdot \vartheta) \cdot d = K_m d^2 \cdot \vartheta$ then the critical level of muscle stiffness is dependent on d in a quadratic way: $K_m = mgh/d^2$. This means that the range of critical stiffness values, characteristic of the experimental setup, is 276÷1962 N/m. On the other hand, the physiological range of hand stiffness [6] is 300÷400 N/m and this means that the experiment allowed us to find under which circumstances the brain chooses stiffness control vs. anticipatory control.

3.2 Results

The analysis focused on the component of the hand oscillation due to the elbow rotation and it was correlated with the EMG activities of two antagonistic muscles of the

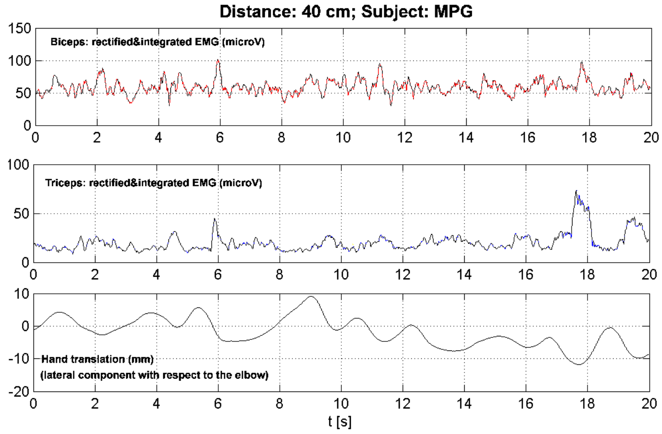
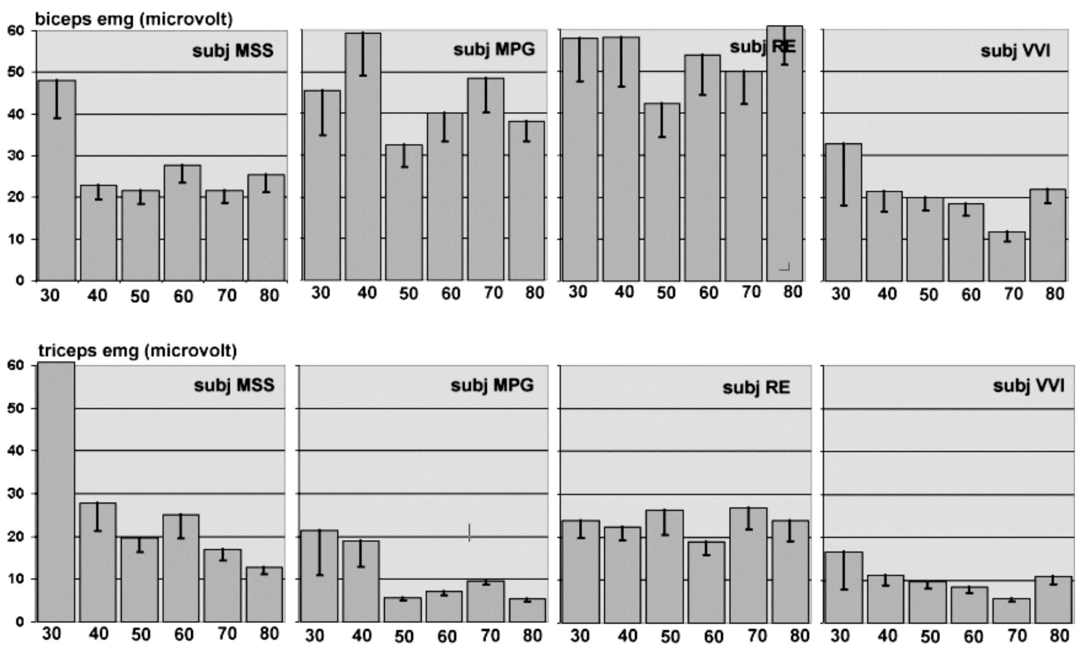


Fig. 5.

elbow: the biceps and the triceps. Fig. 5 shows a typical result for one of the 4 participating subjects: the rectified and integrated EMG activities and the lateral components of the hand oscillation with respect to the elbow. The statistical analysis (fig. 6) yielded the following results:

A) the hand oscillation is significantly correlated with EMG activity (both muscles) in all conditions and all subjects;
B) hand and EMG activities are in phase (biceps delay: 1.7 ± 8.0 ms; triceps delay: –0.02 ± 0.3 ms);
C) coactivation level is weakly dependent upon the critical level of stiffness;

Thus, the prevailing control mode is anticipatory, because on one hand we should expect a delayed muscle activation in the case of reflex control and, on the other, stiffness control should determine an increasing level of coactivation as the critical level of stiffness becomes bigger. Moreover, the same kind of control mechanism seems to be active also for the high values of d, when the critical stiffness is below the physiological level. This result is clearly in contrast with what was found by Burdet et al [5].



4. DISCUSSION

The experimental evidence presented in the two previous sections is somehow contradictory. In one case, upright standing, physiological levels of muscle stiffness are insufficient to stabilise the unstable plant and thus the only feasible solution is anticipatory control. In other cases, involving the upper extremity and with required levels of muscle stiffness which fall inside the physiological range of values, the solution adopted by the brain appears to depend upon the task: a) stiffness modulation in the case of the arm movements in a divergent force field, b) anticipatory compensation in the case of the manual stabilisation of an inverted pendulum. Which kind of circumstances might explain such difference of implementation?

Let us first consider, in general, the dynamics of "fall" which is described by the following equation

$$\ddot{y} = \frac{g}{h_e}(y-u)$$

where y is the position of the COM on the support surface, u is the corresponding position of the COP and h_e is the effective distance between the ankle and the COM. This equation can be derived in the specific case of the standing posture but, with a suitable abstraction, it is applicable to all the unstable loads characterised by an equilibrium state and a divergent force field: $\ddot{x} = \alpha x - u$, where α is a (positive) parameter depending on the structure of the unstable plant and u is the compensatory control variable.

If we consider the transfer function corresponding to the equation above, we realise that the system has two real poles: $p = \pm\sqrt{\alpha}$. The positive pole is the source of instability and, in absence of an appropriate corrective action, determines the exponential "fall" from the equilibrium state ($x = 0$) with the following time constant: $T = 1/\sqrt{\alpha}$. In the case of the standing posture $\alpha = g/h_e$ and if we assume that $h_e = 1\,\mathrm{m}$ we get $T = 320\,\mathrm{ms}$; in the case of the manually controlled inverted pendulum of fig. 4 $\alpha = g/h$, h = 1.8 m and thus $T = 430\,\mathrm{ms}$; in the case of the reaching movements in the divergent force field $\alpha = K_{field}/M$, where K_{field} is the elastic constant of the field (200 N/m) and M is the apparent mass of the hand in the direction collinear with the field (about 1 Kg), thus giving $T \approx 70\,\mathrm{ms}$.

Therefore we can say that in the two cases in which there is evidence of anticipatory compensation the characteristic time constant of the fall is much longer than in the case in which stiffness modulation is the adopted strategy. This result might be explained by considering that anticipatory compensation must have enough time to recover the intrinsic delays of the reafferent pathways (which come close to 100 ms) in order to generate functionally useful anticipatory commands. Therefore, anticipatory compensation in unstable tasks is only feasible if the critical time horizon before a catastrophic fall is significantly longer than 100 ms. As a matter of fact, if we consider a variety of stabilisation tasks such as standing on stilts, rope-walking, balancing a stick etc. it is easy to recognise the fact that the purpose of common tricks, like spreading out the arms or holding a long balancing rod is just to increase the natural "falling time constant", thus giving time to the internal model to generate an appropriate stabilisation action.

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