Biol. Cybern. 85, 39–49 (2001)

Biological
Cybernetics

Simulating mechanical consequences of voluntary movement upon whole-body equilibrium: the arm-raising paradigm revisited

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Received: 2 December 1999 / Accepted in revised form: 8 August 2000

Abstract. Voluntary arm-raising movement performed during the upright human stance position imposes a perturbation to an already unstable bipedal posture characterised by a high body centre of mass (CoM). Inertial forces due to arm acceleration and displacement of the CoM of the arm which alters the CoM position of the whole body represent the two sources of disequilibrium. A current model of postural control explains equilibrium maintenance through the action of anticipatory postural adjustments (APAs) that would offset any destabilising effect of the voluntary movement. The purpose of this paper was to quantify, using computer simulation, the postural perturbation due to arm raising movement. The model incorporated four links, with shoulder, hip, knee and ankle joints constrained by linear viscoelastic elements. The input of the model was a torque applied at the shoulder joint. The simulation described mechanical consequences of the arm-raising movement for different initial conditions. The variables tested were arm inertia, the presence or not of gravity field, the initial standing position and arm movement direction. Simulations showed that the mechanical effect of arm-raising movement was mainly local, that is to say at the level of trunk and lower limbs and produced a slight forward displacement of the CoM (1.5 mm). Backward arm-raising movement had the same effect on the CoM displacement as the forward arm-raising movement. When the mass of the arm was increased, trunk rotation increased producing a CoM displacement in the opposite direction when compared to arm movement performed without load. Postural disturbance was minimised for an initial standing posture with the CoM vertical projection corresponding to the ankle joint axis of rotation. When the model was reduced to two degrees of freedom (ankle and shoulder joints only) the postural perturbation due to arm-raising movement increased compared to the four-joints model. On the basis of these results the classical assumption that APAs stabilise the CoM is challenged.

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1 Theoretical framework

Disturbances of postural equilibrium can result either from unexpected external forces, such as during the sudden unloading of the arm (Hugon et al. 1982) or as a consequence of one's own voluntary movement. In the latter case, when raising the arm, for instance, there are two potential sources of disequilibrium. The displacement of the mass of the arm alters the centre of mass (CoM) position of the whole body. Martin (1967) showed that when the arm lies in the horizontal position, there was a slight backward bending of the trunk which compensated for the forward CoM displacement associated with the new shoulder joint configuration. Otherwise, forward and upward arm acceleration is initiated by internal muscle contractions, which induce on the trunk reaction forces in the opposite direction. These geometrical and inertial forces can lead to postural imbalance.

A current model of the postural control during voluntary movement explains equilibrium maintenance through the action of anticipatory postural adjustments (APAs). Such adjustments and associated postural muscle activities precede and accompany arm lifting (Belenkii et al. 1967) or trunk bending (Babinski et al. 1899; Crenna et al. 1987) movements in order to offset any destabilising effects of the focal part of the task. All these studies have considered the focal movement as a source of perturbation for the equilibrium. Interestingly, similar APAs have been found as well as during locomotion initiation or before focal movement in a standing posture (Crenna and Frigo 1990). This raises a contradiction relative to the role of the APAs. On the basis of these results, the same anticipated postural synergy could curiously assume two opposite functions: the first one would initiate the CoM acceleration during tasks where the base of support is voluntarily displaced (e.g. locomotion, rising on tip-toe or sit to stand); the second one would stabilise the CoM during arm or trunk movements with a fixed base of support.

In contrast with the latter classical assumption that APAs stabilises the CoM, we have recently shown that during a whole body reaching movement with a fixed base of support the CoM was not stabilised along antero–posterior axis (Stapley et al. 1999). We also found that APAs initiated the acceleration of the CoM rather than being involved in CoM stabilisation (Stapley et al. 1998). We postulated that APAs may play a dynamic role for postural transition and contribute additional force to the generation of arm movement (Pozzo et al. 1998; Hodges et al. 1999; Stapley et al. 1999). Nevertheless, one can argue that the role of APAs is task dependent. Thus in the particular case of a whole-body reaching movement the CoM should be displaced to achieve the goal of the task even if we have demonstrated through simulation studies that the target can be reached with little changes in body limb geometry compatible with the stabilisation of the CoM (Stapley et al. 2000). One purpose of this study is to confirm our hypothesis on the role of APAs and to extend our previous approach to a more static movement. If arm-raising movement does not induce CoM displacement, this would confirm that feed-forward postural control is not devoted to stabilise the CoM.

Besides this question, there is another debate on the variable controlled by the APAs (see Hodges et al. 1999). During arm-raising movement APAs could either anticipate the global perturbation and control the resultant CoM position (Massion et al. 1997; Aruin and Latash 1995) or anticipate the local effect on adjacent limbs (inertial forces due to arm acceleration exerted on the trunk) and stabilise the trunk in the sagittal plane (Hodges et al. 1999). If the present study shows that arm movement does not induce CoM but trunk displacement, there will be strong evidence for the control of the trunk orientation by the APAs.

Therefore in order to better understand the true role of APAs, one has to know the exact mechanical passive consequences of the focal movement on the CoM stability. Hence, if we find that the CoM displacement associated with arm-raising movement is negligible, the stabilisation role of APAs must be reconsidered. This also would confirm our hypothesis and would infer that the APAs are dedicated to CoM stabilisation.

2 Previous biomechanical investigations

Interestingly, although numerous speculations around the APAs function have been formulated (see Horak and McPherson 1996, for a review), simulation and accurate quantification of the postural perturbation of the focal part of the task on equilibrium have not been extensively made. Two different kinds of studies investigating APAs during arm-raising movement can exemplify this gap.

A first set of experiments mixed experimental recording with mechanical predictions. Zattara and Bouisset's (1988) experiments primarily focused on the timing of APAs using electromyography and accelerometry. These authors proposed a modelisation which predicted mechanical consequences of arm-raising movement. Their model considered only the acceleration phase and not the whole task which consists of arm acceleration, deceleration and changes in whole-body geometry. Such a modelisation of mechanical effects supposes a sequential conception of the elaboration of the APAs where postural perturbation due to arm deceleration would be anticipated during the short period of arm acceleration. However, the acceleration phase has the opposite effect of the deceleration phase of the arm raising and can cancel the first effect. Therefore, the global mechanical effect can only be anticipated if the central nervous system (CNS) takes into account the whole task.

Another experiment conducted first by Hayes (1982) and extended by Eng et al. (1992) analysed the interaction of the reactive moments and the CoM displacement during the same task. The authors did not study the passive mechanical effect of the focal movement but both active and passive mechanisms of postural control. They recorded the resultant CoM displacements that are assumed to be a measure of the effectiveness of the postural control.

For these two studies, mechanical effects of voluntary movement were predicted on the basis of physical without including computer simulation with time dimension. Henceforth, a lot of experimental investigations have used these mechanical predictions to discuss their results.

The third study consisted in a simulation of a model of postural control during arm raising movement (Ramos and Stark 1990). It was the first which dissociated the passive and active mechanisms of postural control and specifically simulated the mechanical effect of the arm-raising movement. In that model, the simulation was made with a two-link inverted pendulum model. However, the body is a complex multi-joint system which can react differently if forces are transmitted through a simple (two joints) or a complex (at least four joints) modelisation of the whole body.

Our goal is to accurately quantify, using biomechanical simulation, the postural perturbation due to arm-raising movement and to discuss the results in the light of a new conception of APAs. Several simulations have been performed in order to better understand the contribution of each mechanical component to postural disturbance. To this purpose and to discuss our results in comparison with previous experimental observations obtained during active postural control, the simulation has been performed in different initial body positions, with or without load on the arm and for two different arm movement directions. Because arm-raising movement and APAs have been studied in microgravity conditions (Clément et al. 1984, 1985), the simulation has also been performed with or without gravity forces.

Arm-raising movement has been chosen because it is one movements that is the most studied in order to investigate the co-ordination between focal and postural parts of the movement. Moreover, the concept of APAs has been principally developed by using the experimental data provided by this paradigm.

3 Model

3.1 Body modelisation

A standing subject was modelled by four links in the sagittal plane. The four links were the shanks, the thighs, the trunk and the arms which were connected by a one degree of freedom joint (see Fig. 1). Three spatial angles (θ_{Trunk}, θ_{Thigh} and θ_{Ankle}) and one inter-segmental angle ($\theta_{\mathrm{Shoulder}}$) were considered (Fig. 1). The arbitrary normal standing condition was considered when the body CoM vertical projection was superimposed with the ankle joint axis of rotation.

3.2 Body dynamic

Passive and active components of the movement were both considered in the model. The passive component was viscoelastic properties acting at each joint and modelled by a linear torque

$$\mathbf{T}(\text{passive}) = K\mathbf{X}_p + B\mathbf{X}_v$$

where $\mathbf{X}_p$ represents the vector of angular displacement and $\mathbf{X}_v$ represents the vector of angular velocity. B and K are mechanical parameters that are associated with the effective viscous and elastics properties of each joint (Zangemeister et al. 1981a,b). The only active component of the simulation was the torque applied at the shoulder joint to produce arm raising (see Appendix B). This torque represented the degree of activation of agoinst and antagonist muscles expressed in terms of muscle tension level in accordance to the velocity profile recorded by Zattara and Bouisset (1988) during the same movement. Several variables have been implemented in the model: the presence or absence of gravity field, two different additional loads (0.45 and 2.2 kg), three different initial postures (normal, forward and backward tilting) and two arm-raising movement directions (forward and backward).

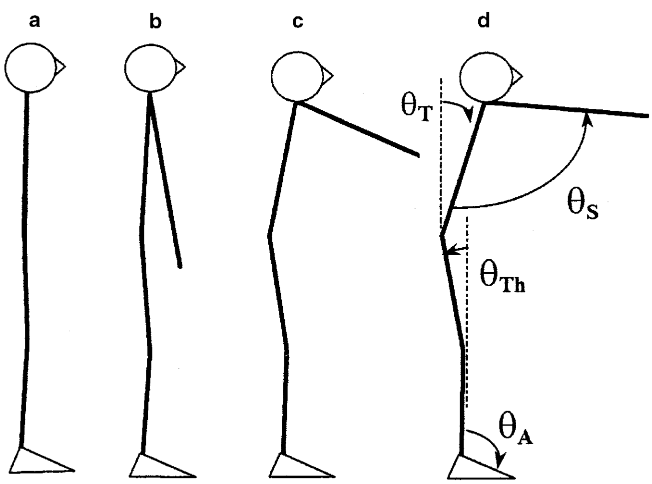

Fig. 1. Simulation of postural disturbance during a rapid upward arm raising. The body is modelled as three rigid links and the two arms together as another link. The stick diagram represents the four successive positions of the body (from left to right): at rest, at maximum acceleration of the arm, at the beginning of the deceleration and at the end of arms movement. Shown are three limb spatial angles: $\theta_{\mathrm{T}}, \theta_{\mathrm{Th}}$, and θ_{A} for trunk, thigh and ankle angular displacements with respect to the vertical and one inter-segmental angle θ_{S} for shoulder angular displacement (arm with respect to the trunk)

3.3 Parameters values

Estimation of body limb parameters, such as mass, inertia and size, were made with reference to the body parameter measurement and estimation scheme proposed by Hanavan (1964) for body modelisation and Ramos et al. (1990) and Zangemeister et al. (1981a,b) for viscous and elastic values of each joint. These biomechanical parameters are given in Table 1.

3.4 Simulation scheme

The upward arm raising followed the movements found in the literature (Zattara and Bouisset 1988) which were used to determine input torque (with or without gravity) to the shoulder joint (see Appendix B). The simulation outputs were the joints angles and derivatives, sagittal position of arm centre of mass ($\mathrm{CoM_u}$), trunk and lower limb centre of mass ($\mathrm{CoM_{tl}}$) and whole body centre of mass (CoM) were calculated using a four rigid-segment model. Figure 2 shows shoulder input torque and the resulting kinematic parameters of arm raising movement in normal gravity (1 *g*) and without gravity (0 *g*). The total movement time, as measured from arm movement velocity, was 350 ms. The total duration of the initial positive moment (flexion) was 130 ms (105 ms in 0 *g*), the peak torque value was 134 N/m (154 N/m in 0 *g*) and the occurrence of the peak moment appeared 40 ms (32 ms in 0 *g*) after the beginning of the movement. The total duration of the negative moment (extension) was 180 ms (195 ms in 0 *g*), the peak torque value was 49 N/m (64 N/m in 0 *g*) and the occurrence of the peak appeared 60 ms (70 ms in 0 *g*) after the beginning of the negative moment. Maximum values of arm velocity,

Table 1. Biomechanical parameter values as estimated from the scheme proposed by Hanavan (1964) for body modelisation (mass, width, length and height are given) and Ramos et al. (1990) and Zangemeister et al. (1981a,b) for joint viscosity (B) and elasticity (K)

Trunk link mass and size	37.5 kg; (0.20 m × 0.3 m × 0.54 m)
Thigh(s) link mass and size	14.9 kg; (0.15 m × 0.2 m × 0.45 m)
Shank(s) link mass and size	7 kg; (0.10 m × 0.13 m × 0.48 m)
Arm(s) link mass and size	6.7 kg; (0.10 m × 0.10 m × 0.6 m)
Ankle	K = 1.0 N m deg^{-1}
	B = 0.2 N m s deg^{-1}
Knee	K = 0.9 N m deg^{-1}
	B = 0.2 N m s deg^{-1}
Hip	K = 0.3 N m deg^{-1}
	B = 0.3 N m s deg^{-1}

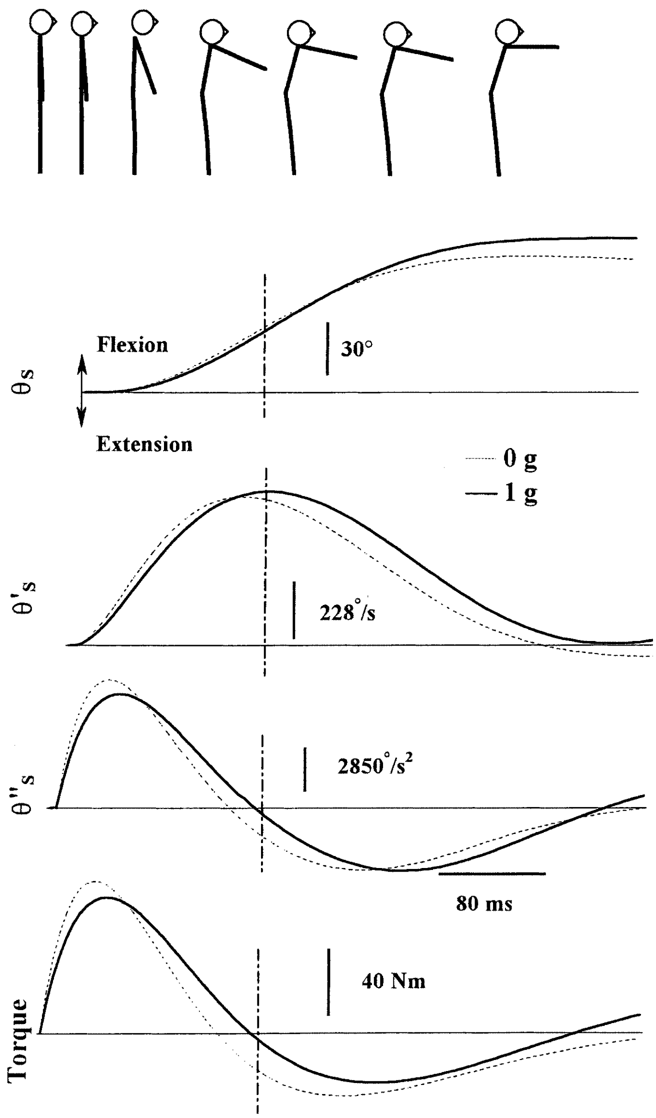

Fig. 2. Stick diagram of the arm-raising simulation with, in the lower part, the corresponding shoulder angular displacement (θ_S), velocity (θ'_S), acceleration (θ''_S) and torque applied to the shoulder to simulate agonist and antagonist muscle activity. Input values of the simulation are given for normal (1 *g*) and for the weightlessness conditions (0 *g*). *Vertical dashed lines* indicate the beginning of the arm deceleration phase

acceleration and deceleration were respectively 580° s^{-1} (same value in 0 *g*) 8156° s^{-2} (9570° s^{-2} in 0 *g*) and 4350° s^{-2} (4582° s^{-2} in 0 *g*). Vertical dashed lines of each figure, represent the occurrence of the arm deceleration phase.

3.5 Formalism

The model consisted of a standing subject performing a rapid raising movement of the two arms. The equations of motion for the inertial torque acting on this system can be derived using the Lagrangian method. For the simulation of the voluntary arm movement, the model system can be represented by a set of ordinary nonlinear differential equations of the first-order (see Appendix A):

$$\vec{\mathbf{X}}(t) = f(\vec{\mathbf{X}}(t), \vec{\mathbf{u}}(t); t) \tag{1}$$

where $\mathbf{X}(t)$ describes the state vector, $\mathbf{u}(t)$ the vector of generalised applied forces or moments. The initial conditions are specified $\mathbf{X}(t_0)$. The torque initiated arm angular acceleration which produced postural imbalance of the whole body described by the generalised co-ordinate $\mathbf{q}$. The differential equations of the multi-joint model were written as (see Schiehlen 1991):

$$\mathbf{H}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{C}(\mathbf{q}, \dot{\mathbf{q}}, t) + \mathbf{G}(\mathbf{q}, t) = \mathbf{F}(\mathbf{q}, \dot{\mathbf{q}}, t) \tag{2}$$

where $\mathbf{H}$ is the inertia matrix, $\mathbf{C}$ the vector of generalised gyroscopic terms, $\mathbf{G}$ the vector of gravitational term and $\mathbf{F}$ the vector of generalised applied joint torque.

For system state equation notation (3), the vector $\mathbf{q}$ of the generalised co-ordinate was represented by the vector of angular displacement X_p

$$\begin{aligned}\dot{\mathbf{X}}(t) &= \begin{bmatrix} \dot{\mathbf{X}}_p \\ \dot{\mathbf{X}}_v \\ \dot{\mathbf{X}}_\nu \end{bmatrix} \\ &= \begin{bmatrix} \mathbf{X}_v \\ \mathbf{H}^{-1}(X_p)(\mathbf{F}(X_p,X_v,X_\nu;t) - \mathbf{C}(X_p,X_v;t) - \mathbf{G}(X_p;t)) \\ \mathbf{\Gamma}(X_\nu) \end{bmatrix}\end{aligned} \tag{3}$$

The vector $\dot{\mathbf{q}}$ of the generalised velocities was represented by the vector of angular velocity $\mathbf{X}_v$, $\mathbf{X}_v$ represented both the activation state of the muscle (active component of shoulder rotation) and passive mechanical parameters associated with effective viscous and elastic properties of each joint.

The vector $\mathbf{\Gamma}$ contained the stimulation function (only for the shoulder joint) and passive forces (viscosity and elasticity forces) at each joint (Appendix B). Angular accelerations, displacements and velocities of the limbs were computed via a fifth-order Runge–Kutta integration routine provided by the software Solid Dynamic Simulator (SDS, Roanne, France).

4 Results

When the arm was stabilised in a horizontal position in a standing condition, the CoM of the upper limbs was located 24 cm forward and the CoM of the whole body was displaced 3.5 cm forward from the initial standing position. This value can be considered as the static effect on the CoM of a whole body geometry change.

Stick figures in the upper part of Figs. 1 and 2 show the successive positions of the whole body obtained from the simulation of an arm-raising movement and illustrate qualitatively the postural perturbation due to arm movement. The main effect was hip and knee flexion producing a collapse of the whole body which ended by a fall into a sitting position on the ground.

Angular displacements of trunk and thigh in the sagittal plane and CoM_{tl} and CoM displacements along the antero–posterior axis are shown in Fig. 3. Data obtained for the two gravity conditions (0 *g* and 1 *g*) and for three conditions of additional loads (unloaded, 0.45 kg and 2.2 kg) have been plotted on the same figure. In order to separate the effect of gravity and inertial forces and to simplify our analysis, we will describe the effect of arm raising first without gravity (0 *g*) and second with gravity force (1 *g*). In this way, in the 0 *g* condition inertial forces are only responsible for postural disturbance.

Without gravity (left part of Fig. 3), the successive arm-raising acceleration and deceleration produced successive trunk and thigh flexion and extension. It may be noted that trunk and thigh movements (upper left part of Fig. 3) reversed about 30 ms after the beginning of the deceleration phase illustrating the inertia of the model given by the term $\mathbf{C}(q, \dot{q}, t)$ in (2). The resulting trunk and lower limb CoM (CoM_{tl}) final position was a backward displacement (2 cm). Due to the forward displacement of CoM of the arm (CoM_{tl}) the final outcome of the whole body was a forward and upward displacement of the CoM (respectively 1.3 cm and 2.5 cm along horizontal and vertical axes) (lower left part of Fig. 3).

In contrast to 0 *g*, in 1 *g* (right part of Fig. 3) trunk and thigh displacement did not reverse during the arm deceleration phase. Consequently, the initial backward displacement of the CoM_{tl} continued and the total amplitude (4.5 cm) increased compared to 0 *g* condition. This result can be explained if we consider that in 1 *g* the effect of gravity on the trunk was maximal at the end of the acceleration phase because of the maximum forward tilting of the trunk with respect to the vertical. Thus arm deceleration, which tends to oppose the forward rotation of the trunk, was not sufficient to bring the trunk back to its initial vertical position. Consequently, and in addition to forward arm CoM displacement, the final CoM position was slightly displaced forward and upward from its initial position (respectively 0.15 cm and 0.85 cm along horizontal and vertical axes). In other words, the gravity tends to minimise the CoM displacement with the consequence that in 1 *g* postural disturbance decreased compared to the 0 *g* condition.

Arm raising performed with additional loads at the level of the wrist (0.45 kg and 2.2 kg) produced similar trunk and thigh displacement curves. Without gravity and with 2.2 kg (see lower left part of Fig. 3), arm raising produced successive backward and forward CoM displacements (1.4 cm), the final CoM position being located slightly forward, not far from its initial position. In 1 *g*, the effect of the loads was an increase of initial

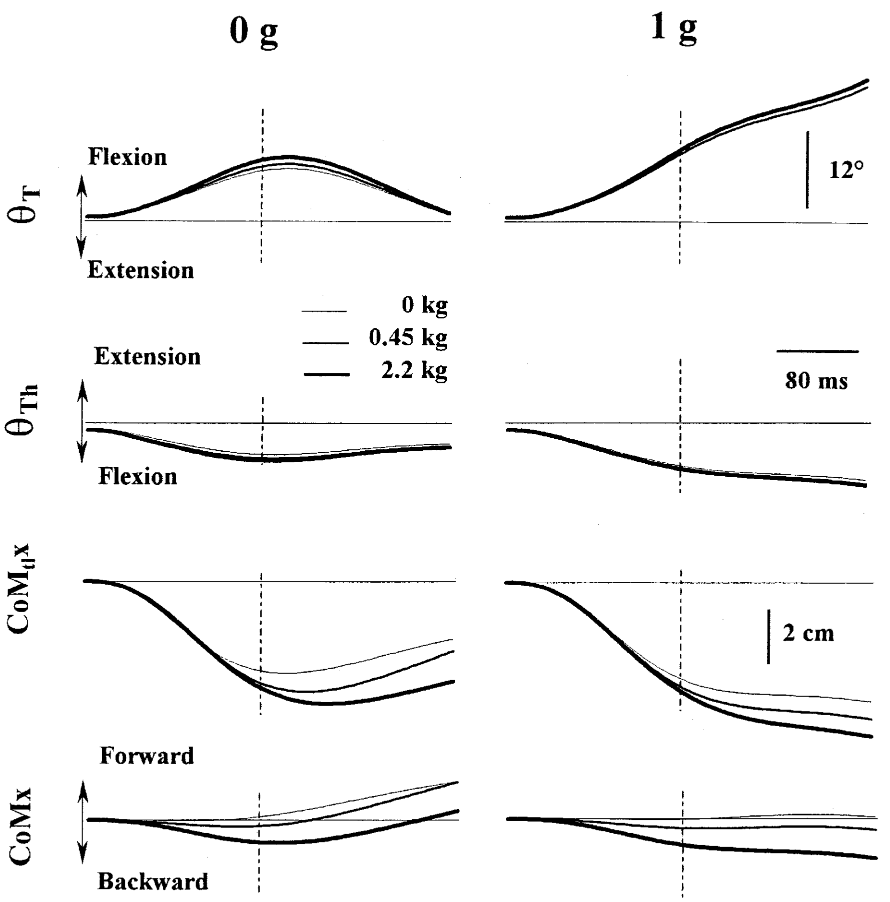

Fig. 3. Effect of arm-raising movement on trunk and thigh angular displacement (θ_{T} and θ_{Th}) and on centre of mass position along antero–posterior axis of the trunk and lower limb ($CoM_{tl}x$) and of the whole body (CoMx). The data is presented for the three conditions of arm loading (0 kg, 0.45 kg and 2.2 kg) with and without gravity field (*left column* 0 *g* and *right column* 1 *g*). *Vertical dashed lines* indicate the beginning of arm deceleration phase

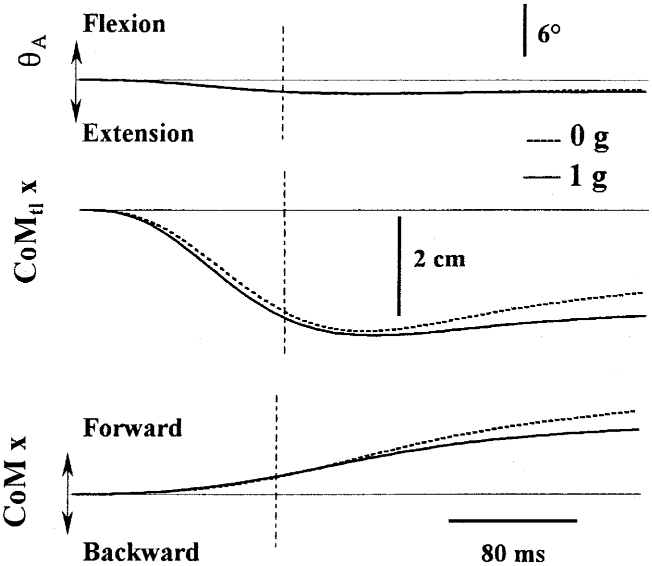

Fig. 4. Postural perturbation due to arm-raising movement on ankle angular displacement (θ_A and on centre of mass position along antero–posterior axis of the trunk and lower limb ($CoM_{tl}x$) and of the whole body (CoMx) for a simple two-link body modelisation (ankle and shoulder joints only). *Vertical dashed lines* indicate the beginning of arm deceleration phase. *Solid* and *dotted lines* represent the results obtained respectively in 1 *g* and in 0 *g* conditions

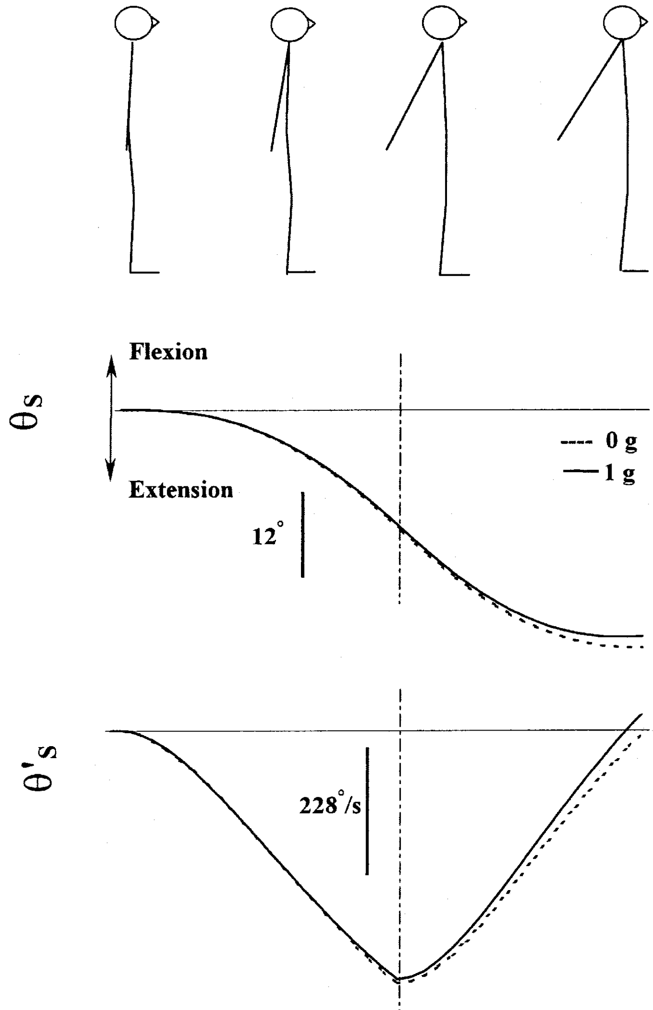

Fig. 5. Stick diagram (upper part) of the backward arm-raising simulation with the corresponding shoulder angular displacement (θ_S) and velocity (θ'_S). *Dotted lines* indicate input values for the 0 *g* condition. *Vertical dashed lines* indicate the beginning of arm deceleration phase

backward CoM displacement (0.5 cm with 0.45 kg and 1.5 cm with 2.2 kg), leading to a backward final position, i.e. in an opposite direction compared to the perturbation obtained without load.

Most interestingly, when the model was reduced to two degrees of freedom (ankle and shoulder joints only) (see Fig. 4) the postural perturbation due to arm-raising movement increased compared to the four-joints model, producing 1.5 cm of forward CoM displacement. This value is similar to the CoM displacement recorded in experimental studies where active postural control is available (Eng et al. 1992; Aruin and Latash 1995).

The simulation of a backward arm raising (shoulder extension) was deliberately limited to 40° due to anatomical constraints (Fig. 5). Figure 6 describes the effect of shoulder extension which resulted in a backward rotation of the trunk (hip extension) and a forward rotation of the thigh (knee extension). In the second part of the movement, thigh rotation reversed due to the blocking of the knee joint in extension, which was at its anatomical limit. This resulted in a forward CoM_{tl} displacement (about 2 cm) and a negligible forward (in 1 *g*) or backward (in 0 *g*) CoM displacements (less than 0.4 mm)

In fact, in 1g, if 40° shoulder extension produced reactive moment to the upper trunk opposite to this that is produced by a 90° shoulder flexion, the resulting postural disturbance was a forward CoM displacement, i.e. in a similar direction to the forward arm raising movement. The similar effect of shoulder flexion and extension on CoM displacement can be explained by the different amplitudes of shoulder angular displacements in the two directions. Thus, simulation of shoulder flexion and extension of the same amplitude (40°) revealed CoM displacements in opposite directions (not shown here).

Figure 7 shows the CoM_{tl} and CoM displacements which follow arm-raising movement from three different initial standing postures, with and without gravity, and without additional load. The results of the simulation indicated that the smaller postural perturbation was obtained when the CoM vertical projection was aligned with the ankle joint axis of rotation.

Greater postural perturbations were obtained when the subject adopted a forward or a backward (±1.8°) tilting standing posture. For these conditions, the arm-raising movement moved the CoM (about 2 cm) in the direction of the initial standing posture. When the CoM was initially located backward (lower part of Fig. 7), arm acceleration and gravity, which acted in the same direction, shifted the CoM_{tl} backward (6 cm). The forward displacement of the CoM of the arm did

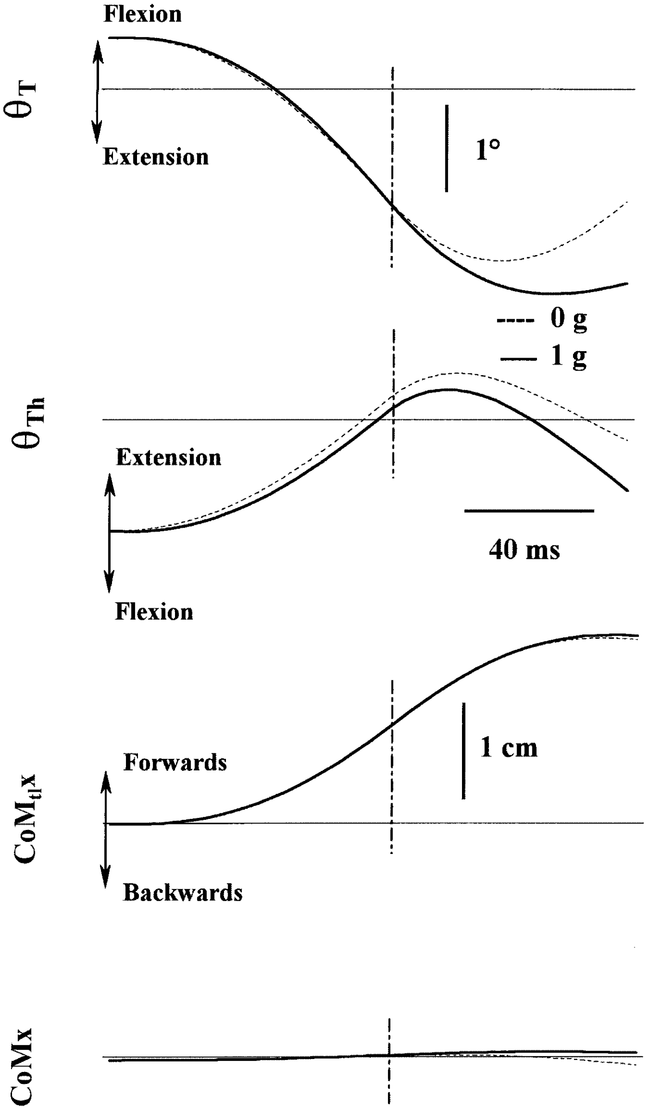

Fig. 6. Postural perturbation due to backward arm-raising movement on trunk and thigh angular displacement (θ_T and θ_{Th}) and on centre of mass position along antero–posterior axis of the trunk and lower limb ($CoM_{tl}x$) and of the whole body (CoMx). The data is presented for the two conditions of gravity (*solid line* 1 *g* and *dotted line* 0 *g*). *Vertical dashed lines* indicate the beginning of arm deceleration phase

not compensate for the CoM_{tl} displacement and the total CoM displacement was 2 cm backward. In contrast, when the body was initially tilted forward (middle part of Fig. 7), inertial and gravity forces acted in an opposite way on the trunk. Consequently the backward CoM_{tl} decreased by a factor of two (3 cm) compared to the initial backward tilted position and the resulting CoM displacement was 2 cm forward. Therefore, the direction of the CoM displacement was dictated by the initial standing position, which determined the subsequent gravity force applied on the whole body. This predominant action of gravity is illustrated on the left part of Fig. 7, which describes the results of the simulation in 0 *g* condition. In this condition, the effect of arm raising on the CoM was always forward regardless of the initial standing posture. The same simulation was made when initial tilted body postures were stabilised. This was achieved by increasing the stiffness (in extension or in flexion respectively for backward and forward tilting postures) of ankle, knee and hip joints. The direction of CoM displacements was not affected, but the amplitude decreased when compared to the condition where the arm-raising movement began from an unstable initial standing posture.

5 Discussion

The results of the simulation investigating mechanical effects that accompany an upward arm movement from a standing posture can be summarised as follows:

1. The mechanical effect of the arm-raising movement is mainly local, i.e. at the level of trunk and lower limbs (CoM_{tl}) in contrast to the negligible perturbation of CoM, which represents the global effect of the arm-raising movement on the whole body.
2. The so-called 'local effect' consisted of trunk rotation in the sagittal plane, inducing a backward CoM_{tl} displacement. Trunk and thigh rotations described here were in agreement with the mechanical predictions of Eng et al. (1992). In fact, our results show that CoM stability is the consequence of compensated effects of the trunk displacement (CoM_{tl}) and of the displacement of the centre of mass of the arm (CoM_{tl}). In other words, inertial forces (i.e. the dynamical aspect of the task including successive arm acceleration and deceleration) are compensated by geometrical body changes (i.e. the static aspect of the task).
3. With additional load (2.2 kg), trunk rotation increased producing a CoM displacement in an opposite direction when compared to arm movement performed without load.
4. Postural disturbance is minimised in the case of an initial standing posture that has a CoM vertical projection aligned with the ankle joint axis of rotation.
5. When the model was reduced to two degrees of freedom (ankle and shoulder joints only) the CoM perturbation due to the arm-raising movement increased compared to the four-joints model.

Finally, no significant effect of the direction of arm movement on CoM displacement was found. With regards to the results of the simulation based on a two- and four-link body modelisation, the following remarks can be made.

5.1 Complex vs. simple body modelisation

The result of the simulation of the postural perturbation obtained with the two-link model was similar to the result of Ramos and Stark (1990). This demonstrates the validity of our mathematical modelisation. However, CoM displacement was greater with the two-joint

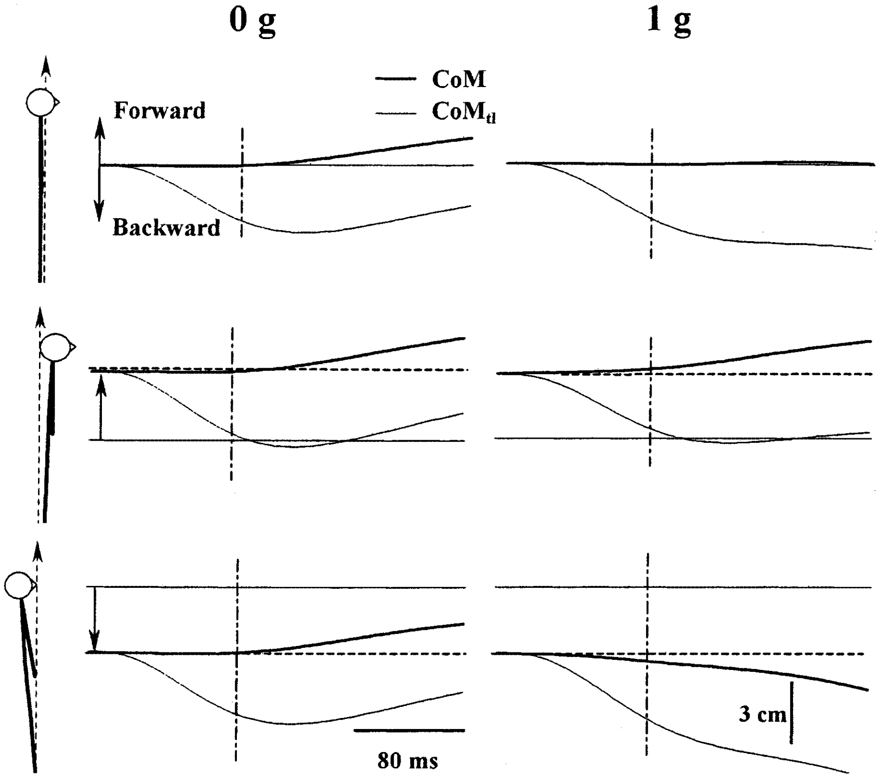

Fig. 7. Postural perturbation due to forward arm-raising movement performed from vertical (upper), forward tilted (middle) and backward tilted (lower) initial standing posture on the centre of mass position along antero–posterior axis of the trunk and lower limb (CoM_{tl}x, *thin line*) and of the whole body (CoMx, *thick line*). The data is presented for the two conditions of gravity (*right column* 1 *g* and *left column* 0 *g*). *Vertical dashed lines* indicate the beginning of arm deceleration phase

compared to the four-joint modelisation. In addition, whilst the long-term effects of arm forward movement on the standing posture was a fall forward, face down, in the two-link model, the four-link model fell down in a sitting position on the ground. This demonstrates the advantages of computer simulations with a model as close as possible to the human body.

Otherwise, experimental recordings have demonstrated important APAs at the level of the hip which consist in reproducible anticipated trunk muscle patterns (Erector spinae, Semitendinous, Biceps femoris) (Hodges et al. 1999; Aruin and Latash 1995) before arm-raising movement. Consequently, a simulation using a model without a hip joint is not appropriate to analyse the postural function of muscles which cross the hip joint (see Ramos and Stark 1990) and a realistic simulation of mechanical perturbations and concomitant APAs can only be made with a model including this joint.

The present result indicates a slight forward displacement of the CoM. This result is not in agreement with the mechanical description of Bouisset and Zattara (1987) who predicted a backward CoM disturbance consecutive to the arm-raising movement. On the basis of their mechanical analysis the authors attributed a stabilising role of the CoM to the APAs which would act in the forward direction in order to compensate for a hypothetical backward CoM disturbance. The discrepancy with the results of our simulation can be explained by the fact that these authors only considered the arm acceleration phase and ignored deceleration effects. In addition, body geometry changes were not introduced in their analysis.

5.2 APAs and CoM stabilisation

The negligible displacement of CoM demonstrated in the present study raises the question of the exact function of postural adjustments and associated feed-forward muscle activities. Comparison of several previous experimental data with our simulation can give potential answers to this question. When active postural control is present, a greater CoM displacement is recorded (Eng et al. 1992; Hodges et al. 1999) compared to our results obtained in the passive condition. This can suggest (1) inefficiency of APAs in the active condition or (2) that APAs are not dedicated to stabilisation of the CoM. The first alternative seems difficult to accept because it supposes that the CNS would achieve a better CoM stabilisation without neural control of posture (as in the present simulation) than after active postural control.

Moreover, we found that the postural disturbance decreased when the initial posture was such that the CoM vertical projection was aligned with the ankle joint axis of rotation. In contrast, sizeable forward and backward initial standing postures increased CoM displacements respectively in the forward and backward directions. However, a slightly forward leading posture can be considered as the normal human standing posture (Schieppati et al. 1994; Gatev et al. 1999). Consequently, if the role of APAs is to stabilise the CoM, we should observe a small backward body till before forward arm-raising movement in order to align the CoM vertical projection with the ankle joint axis of rotation. From the natural forward leaning posture, another solution will consist of activating ankle extensors (soleus or gastrocnemius) before focal movement in order to compensate for the forward postural disturbance revealed by the present simulation. Up to now, anticipated backward postural movement or increase in ankle extensor activity have never been recorded except during rising on tip-toe where Nardone and Schieppati (1988) have demonstrated the role of changing activity of ankle muscles.

5.3 Local (trunk orientation) vs. global (CoM) postural control

Evidence has been presented attributing anticipatory trunk muscle activity to the control of the CoM (Aruin and Latash 1995; Massion et al. 1997). For instance, Aruin and Latash (1995) proposed that feed-forward trunk muscle activation (flexor and extensor) opposed CoM motion and that activation of ankle muscles (mainly the flexor tibialis anterior) counteracts inertial forces acting on the trunk and the lower limb. However, an important result of the simulation is that the main perturbation due to voluntary arm movement was not a CoM displacement, but a hip and knee flexion. Consequently, anticipated trunk muscle activation is more likely dedicated to counteract hip and knee flexion due to arm movement.

The results obtained with the two-link model also support this hypothesis. In simple terms, the two-link model can be assimilated to the active control of a four-link model which consists of freezing hip and knee joints by increasing agonist and antagonist muscle stiffness. Interestingly, with the two-link model, we found a CoM displacement similar in amplitude to the values given by experimental studies where active control is present (Eng et al. 1992; Aruin and Latash 1995). This suggests that APAs controlled trunk and thigh positions rather than CoM stability.

Another result suggests a local function of APAs in contrast to the classical CoM stabilisation role attributed to APAs. We found the same weak forward CoM displacement for both forward and backward arm movements. If APAs were devoted to minimise CoM displacement, experimental data should indicate the same anticipatory pattern of trunk muscle activity for shoulder flexion and extension. Recorded electromyogram (EMG) during forward and backward arm movements, which show opposite trunk muscle patterns for the two directions of arm movement (see Hodges et al. 1999; Aruin and Latash 1995), do not confirm this prediction. In return, the opposite trunk muscle patterns are suitable to compensate for the opposite patterns of trunk rotation and CoM_{tl} displacement found for the two directions of arm movement in the present simulation.

Finally, the results of the simulation obtained in weightlessness support the idea that anticipatory trunk muscle activity mainly controls trunk orientation in space (Friedli et al. 1988; Cresswell et al. 1994; Hodges et al. 1999). Thus, in the 0 g condition we found that the final trunk position was vertical. Otherwise, Clément et al. (1984, 1985) reported that anticipatory bursts of biceps femoris activity in astronauts was extremely reduced by day 7 of a space-flight mission. Such adaptation of anticipatory postural activity can be explained in view of our own results which show that the trunk remains in its initial position after an arm raising movement and consequently does not need to be stabilised through the activity of the biceps femoris.

6 Conclusion

The present study is the first one which simulates, with a four-link body modelisation, postural perturbations due to a rapid arm-raising movement. The simulation shows that the CoM was not significantly displaced and consequently suggests that for this task the variable controlled by the APAs is the trunk position and not the CoM. This does not mean that the CoM is not a controlled variable by the CNS because previous results demonstrated that during a whole-body-reaching task the APAs controlled the CoM displacement inside the base of support (Stapley et al. 1999). We also found that the mechanical analysis of the effect of arm movement depends on the degree of freedom of the modelisation of the human body. In the future, we intend to include more than four joints in the model in order to better reproduce the mechanical constraints exerted on the spine. Finally, one of the limits of the present study is that we used an indirect method to determine the function of EMG activities recorded before the beginning of the focal movement. A complementary approach would be to directly simulate the action of APAs and associated muscle activities by applying torques at each joint.

Acknowledgements. This work was supported by the CNES (Centre National d'Etudes Spatiales). We gratefully acknowledge Prof. M. Schieppati, Dr. P. Stapley and Dr. A. Grishin for their helpful comments during the preparation of the manuscript.

Appendix A

The model of the human body consists of four linked segments (shanks, thigh, trunk and arm). The segments are connected by a one degree of freedom joint. The model is bi-dimensional and can only be used for antero–posterior movements of the body. The equations of motion for the inertial torques acting on this system can be derived using the Lagrangian method.

The equation can be written in compact form

$$\mathbf{J}(\mathbf{\Theta})\ddot{\mathbf{\Theta}} + \mathbf{C}(\mathbf{\Theta})\dot{\mathbf{\Theta}}^2 + \mathbf{G}(\mathbf{\Theta}) = \mathbf{F}(\mathbf{\Theta}, \dot{\mathbf{\Theta}}, t) \qquad \text{(A1)}$$

where $\mathbf{\Theta}$ is a vector $[\theta_{\mathrm{Arm}}, \theta_{\mathrm{Trunk}}, \theta_{\mathrm{Thigh}}, \theta_{\mathrm{Shanks}}]$ whose components are segmental angles relative to the vertical axis; $\mathbf{J}(\mathbf{\Theta})$ is a symmetrical inertia matrix (4×4), $J(i,j) = J(j,i) \sim \cos(\theta_i - \theta_j)$; $\mathbf{C}(\mathbf{\Theta})$ is an antisymmetrical matrix (4×4), $C(i,j) = -C(j,i) \sim \sin(\theta_i - \theta_j)$; $\mathbf{G}(\mathbf{\Theta})$ is the gravitation vector (4×1), $G(i) \sim \sin(\theta_i)$; $\mathbf{F}(\mathbf{\Theta}, \dot{\mathbf{\Theta}}, t)$ is the joint torque vector

(4×1); passive viscosity and elasticity, and active torque for each joint: $F(i) = -K_i\theta_i - B_i\dot{\theta}_i + F_{i\mathrm{Active}}$. The only active action of the model is the torque applied at the shoulder joint.

The SDS software computes at each time sample, via a fifth-order Runge–Kutta integration routine, angular displacement, velocity and acceleration of the segments.

The elements of vectors and matrixes represented in equation (A1) are given below. $I_i (i = 1 \ldots 4)$ is the moment of inertia of the ith segment relative to the centre of the corresponding segment, $m_i (i = 1 \ldots 4)$ is its mass and $L_i (i = 1 \ldots 4)$ is its length.

$$J(1,1) = I_{\mathrm{Shank}} + ((m_{\mathrm{Shank}}/4) + m_{\mathrm{Thigh}} + m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L^2_{\mathrm{Shank}}$$

$$J(1,2) = J(2,1) = (1/2m_{\mathrm{Thigh}} + m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L_{\mathrm{Shank}} \times L_{\mathrm{Thigh}} \cos(\theta_{\mathrm{Shank}} - \theta_{\mathrm{Thigh}})$$

$$J(1,3) = J(3,1) = (1/2m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L_{\mathrm{Shank}}L_{\mathrm{Trunk}} \times \cos(\theta_{\mathrm{Shank}} - \theta_{\mathrm{Trunk}})$$

$$J(1,4) = J(4,1) = (1/2m_{\mathrm{Arm}})L_{\mathrm{Shank}}L_{\mathrm{Arm}} \times \cos(\theta_{\mathrm{Shank}} - \theta_{\mathrm{Arm}})$$

$$J(2,2) = I_{\mathrm{Thigh}} + ((m_{\mathrm{Thigh}}/4) + m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L^2_{\mathrm{Thigh}}$$

$$J(2,3) = J(3,2) = (1/2m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L_{\mathrm{Thigh}}L_{\mathrm{Trunk}} \times \cos(\theta_{\mathrm{Thigh}} - \theta_{\mathrm{Trunk}})$$

$$J(2,4) = J(4,2) = (1/2m_{\mathrm{Arm}})L_{\mathrm{Thigh}}L_{\mathrm{Arm}} \times \cos(\theta_{\mathrm{Thigh}} - \theta_{\mathrm{Arm}})$$

$$J(3,3) = I_{\mathrm{Trunk}} + ((m_{\mathrm{Trunk}}/4) + m_{\mathrm{Arm}})L^2_{\mathrm{Trunk}}$$

$$J(3,4) = J(4,3) = (1/2m_{\mathrm{Arm}})L_{\mathrm{Trunk}}L_{\mathrm{Arm}} \times \cos(\theta_{\mathrm{Trunk}} - \theta_{\mathrm{Arm}})$$

$$J(4,4) = I_{\mathrm{Arm}} + 1/4m_{\mathrm{Arm}}L^2_{\mathrm{Arm}}$$

$$C(1,2) = -C(2,1) = -(1/2m_{\mathrm{Thigh}} + m_{\mathrm{Trunk}} + m_{\mathrm{Arm}}) \times L_{\mathrm{Shank}}L_{\mathrm{Thigh}} \sin(\theta_{\mathrm{Shank}} - \theta_{\mathrm{Thigh}})$$

$$C(1,3) = -C(3,1) = -(1/2m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L_{\mathrm{Shank}} \times L_{\mathrm{Trunk}} \sin(\theta_{\mathrm{Shank}} - \theta_{\mathrm{Trunk}})$$

$$C(1,4) = -C(4,1) = -(1/2m_{\mathrm{Arm}})L_{\mathrm{Shank}}L_{\mathrm{Arm}} \times \sin(\theta_{\mathrm{Shank}} - \theta_{\mathrm{Arm}})$$

$$C(2,3) = -C(3,2) = -(1/2m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})L_{\mathrm{Thigh}} \times L_{\mathrm{Trunk}} \sin(\theta_{\mathrm{Thigh}} - \theta_{\mathrm{Trunk}})$$

$$C(2,4) = -C(4,2) = -(1/2m_{\mathrm{Arm}})L_{\mathrm{Thigh}}L_{\mathrm{Arm}} \times \sin(\theta_{\mathrm{Thigh}} - \theta_{\mathrm{Arm}})$$

$$C(3,4) = -C(4,3) = -(1/2m_{\mathrm{Arm}})L_{\mathrm{Trunk}}L_{\mathrm{Arm}} \times \sin(\theta_{\mathrm{Trunk}} - \theta_{\mathrm{Arm}})$$

$$G(1) = -(1/2m_{\mathrm{Shank}} + m_{\mathrm{Thigh}} + m_{\mathrm{Trunk}} + m_{\mathrm{Arm}}) \times gL_{\mathrm{Shank}} \sin(\theta_{\mathrm{Shank}})$$

$$G(2) = -(1/2m_{\mathrm{Thigh}} + m_{\mathrm{Trunk}} + m_{\mathrm{Arm}}) \times gL_{\mathrm{Thigh}} \sin(\theta_{\mathrm{Thigh}})$$

$$G(3) = -(1/2m_{\mathrm{Trunk}} + m_{\mathrm{Arm}})gL_{\mathrm{Trunk}} \sin(\theta_{\mathrm{Trunk}})$$

$$G(4) = -(1/2m_{\mathrm{Arm}})gL_{\mathrm{Arm}} \sin(\theta_{\mathrm{Arm}})$$

Appendix B

The torque applied to the shoulder is represented in (A2) in order to show the adjustments to the torque value of arm kinematic values found in the literature (amplitude, acceleration duration, deceleration duration and attenuation):

$$F_{\mathrm{Shoulder}}(t) = Ae^{-\alpha t}(t - t_0)(t - t_1)(t - t_2) \qquad \text{(A2)}$$

Amplitude and attenuation corresponding to the torque envelope

$$Ae^{-\alpha t}$$

where A is amplitude, α is attenuation (which is a constant and inverse of the time). Acceleration and deceleration phases are determined by a third-order polynomial:

$$(t - t_0)(t - t_1)(t - t_2)$$

where t_1 and t_2 correspond respectively to the end of acceleration (this one beginning at $t_0 = 0$) and deceleration phases – t_1 and t_2 values are based on experimental recordings. A and α are obtained from maximum velocity given by experimental recordings (when $t = t_1$) and from limit conditions $\mathbf{V}(t_2) = 0$.

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