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A telescopic inverted-pendulum model of the musculo-skeletal system and its use for the analysis of the sit-to-stand motor task

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Abstract

For field applicability of biomechanical methodologies aiming at assessing motor ability in disabled, or at risk of disablement (e.g. elderly), subjects, measurements must be carried out using a least perceivable to the subject and essential experimental apparatus. Since data thus obtained do not necessarily lend themselves to straightforward interpretation, they should be fed to a model of the portion of the musculo-skeletal system involved that already embodies the invariant aspects of both the modelled system and the motor task. Through such a *minimum measured-input model*, richer, physiology-related, and thus easier to interpret, information may be expected. In this framework, the present study investigated the sit-to-stand motor task using information obtained only from a force plate located under seat and subject's feet, a seat uniaxial load-cell and basic anthropometric parameters. Data were collected in a sample of 12 able-bodied subjects while executing the motor task at different speeds. The musculo-skeletal system was modelled as a *telescopic inverted pendulum* (TIP) that could vary its length (shortening or elongation) by effect of a force actuator and its orientation in space by effect of two couple actuators that were looked upon as muscle equivalent effectors. The TIP model output consisted in the kinematics and dynamics of these actuators. It allowed the identification of four functional phases in which the seat-to-stand motor task could be divided, and a detailed description of the relevant mechanics in terms of balance control and centre of mass elevation. Motor strategy modifications associated with speed variation could also be identified. For a global evaluation of the motor act it showed to be no less informative than more demanding multi-segment models. Although it is true that specific musculo-articular functions can only be inferred, the more compact information yielded by the TIP model is expected to facilitate subject and/or disability classification.

Keywords: Modelling; Musculo-skeletal system; Sit-to-stand; Force plate

1. Introduction

The quantitative assessment of *physical functional reserve* (Pendergast et al., 1993) in a disabled, or at risk of disablement (e.g. elderly), person is an intriguing objective of clinical biomechanics. It entails the analysis of suitable motor tasks and the inference of information mostly related with muscular strength and balance control. The biomechanical analysis of these motor tasks, carried out using stereophotogrammetry, dynamometry, electromyography, and multi-segment human body modelling, provides thorough relevant information. However, this approach is difficult to apply for subject-specific evaluation in clinical practice by reason of the complexity of both instrumentation and experimental protocols. The so-called functional tests, currently adopted in clinical settings, are inexpensive, easy to administer, and well accepted by the test subjects. However, they rely primarily on semi-quantitative, and in some instances, subjective observations (Tinetti, 1986; Perry et al., 1995) leading to lack of reliability of the results. Therefore, the question arises as to whether methods may be devised that join objectivity with field applicability.

This study aims to answer the above general question. The basic idea is to minimise the number of variables to be measured during the execution of the selected motor task and to acquire them using an experimental apparatus least perceivable to the test subject. However, since data thus obtained do not necessarily lend themselves to straightforward interpretation in terms of physical functional reserve assessment, they should be fed to a model

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Nomenclature

CM	centre of mass
FA	frontal rotational actuator
HAT	head-arms-torso system
HS	high speed
LA	linear actuator
NS	natural speed
SA	sagittal rotational actuator
STS	sit-to-stand
TIP	telescopic inverted pendulum
TIP1	TIP model representing the HAT only, applied in the interval of time preceding the beginning of seat unloading
TIP2	TIP model representing the WB, applied in the interval of time following seat-off
WB	whole body
T	task duration, s
t_ω	instant of occurrence of maximal HAT angular velocity, % of T
t_{bu}	instant at which seat-unloading begins, % of T
t_{so}	instant of seat-off, % of T
t_v	instant of occurrence of maximal WB linear velocity, % of T
t_c	instant of occurrence of maximal HAT SA couple, % of T
θ_{bu}	HAT angle with respect to the vertical at t_{bu}, deg
v_2	maximal value of WB linear velocity, m s^{-1}
ω_1	maximal value of HAT angular velocity, rad s^{-1}
ω_{so}	value of WB angular velocity at t_{so}, rad s^{-1}
F_2	normalised maximal value of WB LA force, N kg^{-1}
C_1	normalised maximal value of HAT SA couple, N kg^{-1}
C_{so}	normalised value of WB SA couple at t_{so}, N kg^{-1}
C_2	normalised maximal value of WB SA braking couple, N kg^{-1}
LP_2	normalised maximal value of WB LA power, W kg^{-1} m^{-1}
SP_2	normalised maximal value of WB SA power, W kg^{-1} m^{-1}

of the portion of the musculo-skeletal system involved that embodies the invariant aspects of both the modelled system and the motor task. Through such a *minimum measured-input model*, richer, physiology-related, and thus easier to interpret, information may be expected.

In this study, a minimum input model is presented, and applied for the analysis of the sit-to-stand (STS) motor task. The latter was selected as a biomechanically demanding task. In fact, significant leg muscle strength and wide ranges of joint motion are involved, associated with a considerable challenge to balance. Furthermore, being an activity of daily living, normally it is not affected by the subject's motivation and learning, and it is safe. For these reasons, STS is generally included in performance tests for certain categories of disabled and/or elderly persons (Tinetti, 1986; Berkman et al., 1993; Guralnik et al., 1994; Tinetti et al., 1995; Means, 1996).

The model uses only information obtained from a six component force plate, a seat uniaxial load-cell and basic anthropometric data. Data were collected in a sample of able-bodied subjects and the following questions addressed. (1) Does the model output provide an effective description of the STS motor task, and how does this description compare with that reported in the literature obtained using both force and multi-segmental movement measurements (Kralj et al., 1990; Pai et al., 1990a,b; Riley et al., 1991)? (2) Is it possible, using the model, to characterise in quantitative terms a motor strategy variation determined by a change in speed of execution of the motor task? Positive answers to these questions would indicate the possibility that, using this model, a subject-specific motor strategy, as associated with the relevant neuro-musculo-skeletal system functional reserve, may be disclosed.

2. The telescopic inverted-pendulum model

Biped human locomotor acts are characterised by a rotation about the base of support, accompanied by a shortening or elongation of the musculo-skeletal structure as a whole. Then the location of the base of support may be changed and the procedure repeated, thus obtaining the displacement of the body centre of mass (CM) through space. This essential way of describing locomotion inspires an equally essential model of the musculo-skeletal system based on a *telescopic inverted-pendulum* (TIP). This is composed of a telescopic massless link hinged at a point effectively approximating the base of support and joining the latter point with the CM of the moving portion of the body where the mass of the latter is supposed to be concentrated. The link may vary its length (shortening or elongation) by effect of a force (linear) actuator (LA) and its orientation in space by effect of two couple (rotational) actuators. One rotational actuator is responsible for forward and backward rotations (sagittal plane rotational actuator, SA), and the other for lateral rotations (frontal plane rotational actuator, FA). These actuators may be looked upon as muscle equivalent effectors.

The model has three degrees of freedom, corresponding to those of a particle (the CM of the relevant body portion) moving in the 3-D space, and its input is the trajectory of the CM and its parameters are the relevant mass and the stationary hinge location. Note that, by its own nature, the model does not take account of angular inertia effects.

For the analysis of STS, two TIP models were used in temporal sequence that represented two different systems. During the interval of time preceding seat unloading, only the head–arms–torso (HAT) system moves (Schenkman et al., 1990) and therefore was accounted for in the model (TIP1, Fig. 1). Following the loss of contact with the seat (seat-off) the whole body (WB) contribution was considered (TIP2, Fig. 1). During the transient phase of seat unloading neither of the two TIP models was applicable.

TIP1 is hinged at the midpoint between the hips, and TIP2 at the midpoint between the ankles. The LA is the main responsible for the elevation of the CM. In TIP1, assuming that the upper limbs and the head do not move with respect to the trunk, LA can represent the action of trunk muscles that increase or decrease lumbar lordosis, and, in TIP2, the action of trunk, hip and knee flexors and extensors. The SA of TIP1 controls the HAT rotation and can represent the action of the trunk flexors and extensors in co-operation with hip flexors and extensors. In TIP2, SA controls balance, through the control of the antero-posterior distance between the WB CM and the ankle hinge, and can be associated with the action of trunk, hip, knee and ankle flexors and extensors. The FA accounts for lateral balance and can be associated with trunk lateral bending muscles in TIP1 and with the latter muscles in addition to lower limb ab-adductors in TIP2. It is expected that, in an able-bodied population, frontal plane quantities do not carry useful information (Wheeler et al., 1985), however, the opposite may occur when disabled subjects are analysed. In this perspective frontal quantities are included in the model presented herein.

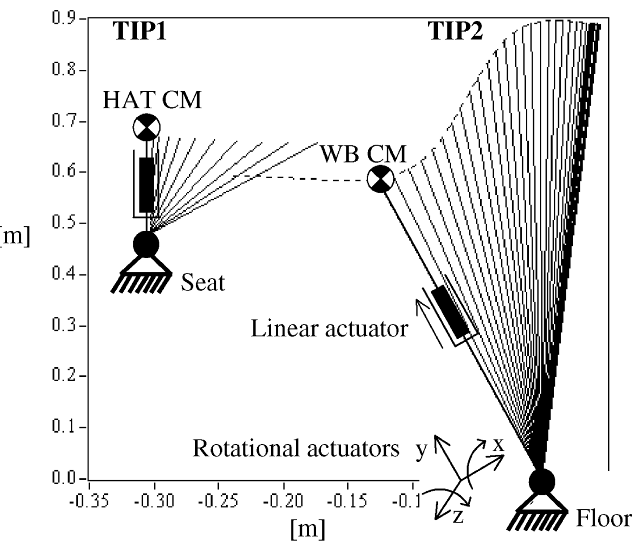

Fig. 1. Schematic representation of the two TIP models employed for the description of STS. TIP1: before seat unloading. TIP2: after seat-off. The sign convention for the actuators is also indicated.

3. Materials and methods

3.1. Subjects and experimental protocol

A sample of 12 healthy young adults (6 males and 6 females, age 22–34, body mass 48–84 kg, stature 1.58–1.78 m) was investigated, after informed consent had been obtained. The experimental apparatus adopted consisted of: (1) a modular seat without backrest; (2) a six-component Bertec force plate (0.4 m × 0.6 m), positioned under both the seat and the subject's feet; (3) a uniaxial load-cell embedded in the seat; (4) software tools for data acquisition and processing purposely developed using National Instruments LabView®. In order to make the subject feel comfortable, the base of support provided by the force plate was enlarged by means of a wooden platform (0.6 m × 0.9 m × 0.02 m). Nevertheless, during the experiments, the centre of pressure remained within the area defined by the four force transducers.

Subjects were asked to sit in a standardised posture (seat height adjusted at 80% of knee height from the ground, buttocks and proximal third of the thighs in contact with the seat, arms folded across the chest, vertical trunk, 18° ankle dorsiflexion, self-selected feet distance apart and orientation). Their anatomical planes were consistent with the force plate (laboratory) frame. During quiet sitting, the co-ordinates of points approximating the hip, knee and ankle joint centres in the laboratory frame were reconstructed using a photograph. Subjects were instructed to rise from the seat at an operator command and remain still once the upright position was reached until a further operator command was imparted. The motor task was first executed at natural (self-selected) and then at a maximum speed neither moving the feet nor lifting the shoulders. For each speed, natural [NS] and high [HS], five trials were recorded.

During each trial, the force plate and seat load-cell readings were acquired for 5 s with a sampling rate of 100 sample/s. Note that the force plate moment components were not used in the model.

3.2. Data processing

The inertial parameters of body segments were estimated using the regression equations provided by Chandler et al. (1975) and individual anthropometric measures. The initial coordinates of the HAT and WB CM were calculated using the latter parameters and the initial posture geometry. The corresponding CM trajectories were then estimated through double integration of the inertial components of the ground reaction force divided by the relevant mass. The appropriate CM 3-D kinematics and model parameters were fed to the two TIP models. The CM antero-posterior and medio-lateral co-ordinate values thus obtained during quiet sitting and quiet standing could be compared, for validation purposes, with the relevant coordinates of the centre of pressure calculated from the force plate output. The maximal difference found was 3 cm.

The instantaneous orientation in space of the telescopic link, i.e. of the relevant local frame relative to the laboratory frame, was represented through a sequence of two rotations, according to the Bryant convention (Wittenburg, 1977). The angles describing these rotations (θ_S, θ_F) and the length of the link (l) were regarded as generalised coordinates of the model.

The model output in terms of force (F) and couple vector (**C**) supplied by the actuators was then obtained, in each sampled time interval, using the following equations, where vectors are defined in the local (principal) reference frame:

$$F = m\mathbf{a}_{\mathrm{CM}} \cdot \hat{l} - m\mathbf{g} \cdot \hat{l}, \tag{1}$$

$$\mathbf{C} = \frac{\mathrm{d}([J]\boldsymbol{\omega})}{\mathrm{d}t} + \boldsymbol{\omega} \times [J]\boldsymbol{\omega} - \hat{l} \times m\mathbf{g}, \tag{2}$$

where $\hat{l}$ indicates the link versor, m represents the mass of the relevant body portion, $\mathbf{a}_{\mathrm{CM}}$ the CM acceleration, $m\mathbf{g}$ the gravitational force and $\hat{l} \times m\mathbf{g}$ its moment with respect to the centre of the hinge,

$$[J] = \begin{bmatrix} ml^2 & 0 & 0 \\ 0 & 0 & 0 \\ 0 & 0 & ml^2 \end{bmatrix} \tag{3}$$

is the inertia matrix, and

$$\boldsymbol{\omega} = \begin{bmatrix} 0 & 1 & 0 \\ \sin\theta_F & 0 & 1 \\ \cos\theta_F & 0 & 0 \end{bmatrix} \begin{bmatrix} \dot{\theta}_S \\ \dot{\theta}_F \\ 0 \end{bmatrix}, \tag{4}$$

the angular velocity. The first two terms of the right-hand side of Eq. (2) will be referred to as inertia couple. The **C** medio-lateral and antero-posterior components were regarded as associated with the SA and FA, respectively.

The instantaneous velocities of the three actuators were calculated as

$$v = \frac{\Delta l}{\Delta t}, \quad \omega_S = \frac{\Delta\theta_S}{\Delta t}, \quad \omega_S = \frac{\Delta\theta_S}{\Delta t}, \text{ respectively,}$$

where $\Delta\theta_S$ and $\Delta\theta_F$ were obtained by applying the Bryant convention to the local frame orientation in two successive sampled instants of time, and Δt is the sampling interval. The instantaneous power of the actuators was calculated by multiplying the above velocities by the relevant force or couple component.

The CM momentum components are given by $p_l = mv$, $p_S = m\omega_S l$, and $p_F = m\omega_F l$. Thus, velocities may be looked upon as momenta normalised with respect to body mass and body mass times the link length, respectively. Note that the conclusions drawn analysing these quantities were not affected by the fact that l varied in time. Force components were normalised to body mass, and power and couples to the product body mass times stature.

In order to characterise the pattern vs. time of the aforementioned variables, selected peak values and values at specific event times were extracted (Figs. 3–6; see Nomenclature). The inertial and gravitational contributions to the actuator force and couples, as defined by the appropriate right-hand-side terms of Eqs. (1) and (2), were also analysed.

The instants of time at which the motor task began and ended, which defined task duration (T), were detected using appropriate thresholds on the SA and the LA accelerations, respectively. The time of occurrence of significant events was represented as percent of T.

A paired t-test was performed to investigate significance of differences between NS and HS parameters.

4. Results

4.1. Frontal plane rotational actuator

Variables relative to the FA did not carry significant information (peak-to-peak values of velocity and couple within 15% of those relative to the SA, low intra- and inter- individual repeatability, no significant changes with speed of execution).

4.2. Seat unloading timing

The seat load-cell permitted the accurate detection of the seat-off occurrence (t_{so}). On the contrary, the load-cell signal displayed no obvious discontinuity associated with the beginning of seat unloading (t_{bu}). By pure convention, t_{bu} was made to occur when the load-cell reading crossed the quiet sitting base line and began decreasing continuously (Fig. 2, Table 1).

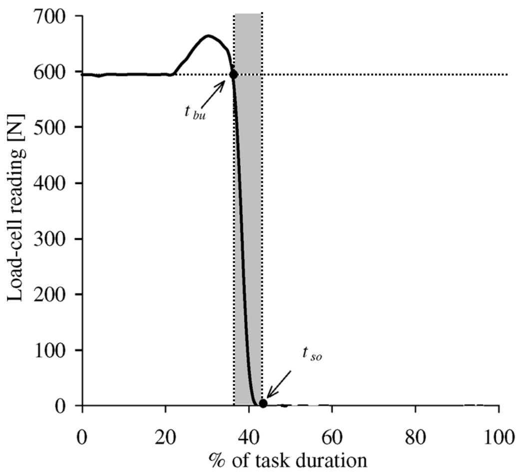

Fig. 2. Seat load-cell reading for one randomly selected trial at NS. The grey area represents the seat unloading phase.

Table 1
Mean values (mean) and standard deviations (S.D) are reported for the selected parameters at NS and HS of execution. Δ mean indicates the difference between the parameters if this difference is significant ($p < 0.01$). Symbols are listed at the beginning of the paper

Parameter	Natural speed		High speed		Δ mean
	Mean	S.D.	Mean	S.D.	
T (s)	1.54	0.31	0.96	0.18	− 37%
t_ω (% of T)	33.0	3.4	36.2	3.6	—
t_{bu} (% of T)	39.0	4.7	42.0	4.2	—
t_{so} (% of T)	48.0	4.0	50.6	4.0	—
t_v (% of T)	60.0	4.6	62.7	3.9	—
t_c (% of T)	13.0	3.2	21.0	5.0	8
θ_{bu} (deg)	29.30	6.00	22.10	7.20	− 25%
v_2 (m s^{-1})	0.60	0.14	0.98	0.17	63%
ω_1 (rad s^{-1})	1.80	0.27	2.10	0.40	18%
ω_{so} (rad s^{-1})	0.70	0.14	0.93	0.14	34%
F_2 (N kg^{-1})	12.40	0.70	15.00	1.20	21%
C_1 (N kg^{-1})	0.10	0.04	0.21	0.09	116%
C_{so} (N kg^{-1})	0.36	0.15	0.45	0.20	—
C_2 (N kg^{-1})	—	—	0.21	0.07	—
LP_2 (W kg^{-1} m^{-1})	3.80	1.10	7.40	1.90	93%
SP_2 (W kg^{-1} m^{-1})	0.28	0.14	0.39	0.20	37%

4.3. Analysis of the motor task

Examples of patterns vs. time of velocity, i.e. normalised momentum, force, couple and power of the two actuators are depicted in Figs. 3–6, respectively, for a NS and a HS trial of the same randomly selected subject. In Table 1 the sample means and standard deviations of selected parameters are reported. When significant ($p < 0.01$), their variation with speed is also given.

Based on SA and LA maximal velocity occurrence (ω_1 and v_2 at t_ω and t_v, respectively Fig. 3), t_{bu} and t_{so}, five functional phases of the motor task were identified.

Phase 1: *HAT angular acceleration* ($0 - t_\omega$). The HAT undergoes a rotational acceleration while flexing (Fig. 3) under the action of the SA couple (Fig. 5, C_1 in Table 1). Soon after the SA couple changes sign (negative power, Figs. 5 and 6), however the HAT still accelerates under the increasing action of the gravitational force moment (Fig. 5). The LA starts moving after 10% of T. After moderate compression, it starts elongating, producing positive power (Figs. 3 and 4 and 6).

At the higher speed, the SA positive couple acts for a longer portion of T (Fig. 5; see also t_c in Table 1), mean C_1 and ω_1 increase by 116, and 18%, respectively.

Phase 2: *HAT angular deceleration* ($t_\omega - t_{bu}$). The extending action of the SA couple (negative power, Figs. 5 and 6), balances gravity and, in addition, causes angular deceleration of the HAT (Fig. 3). At HS, deceleration at t_{bu} is higher and occurs with the torso more erect (θ_{bu} in Table 1) than at NS. Note that the link elongation and consequent moment of inertia augmentation (see Eq. (2)) caused by the LA, contributes to HAT deceleration. This may be appreciated at HS by looking at the inertia couple at t_ω (zero angular acceleration) in Fig. 5b.

Phase 3: *momentum transfer* ($t_{bu} - t_{so}$). At t_{bu}, the HAT is flexing while decelerating. Immediately after, as seen in the literature (Ikeda et al., 1991; Kralj et al., 1990), its flexion ceases and the hip joint is momentarily blocked allowing for momentum transfer from the HAT to the WB through rotations about the knee and/or ankle joints. In fact, as soon as the seat becomes unloaded, i.e. at t_{so}, the WB already exhibits a forward rotational velocity (ω_{so} in Fig. 3 and Table 1). During this phase, lower limb muscles alone cannot produce momentum (no one can rise starting with the CM at zero velocity and located behind the base of support provided by the feet), i.e. movement is certainly partly ballistic, and thus momentum transfer has occurred (Kralj et al., 1990). At HS, ω_{so}, and thus WB momentum, is significantly higher than at NS. This is due to a higher HAT angular deceleration during phase 2 associated with a more erect torso. These two factors contribute to a larger inertia force acting orthogonal to the HAT and forward, and to a greater value of its moment arm relative to the knee and/or ankle joints. This makes inertia action and, thus, momentum transfer at HS more effective than at NS. The transfer of the momentum associated with elongation did not exhibit any consistent trend.

Phase 4: *WB linear acceleration* ($t_{so} - t_v$). The WB undergoes a linear acceleration under the action of the LA elongating force (positive power, Figs. 4 and 6, and LP_2 in Table 1), the maximum value of which (F_2) at HS is 21% larger than at NS. This force builds up a momentum in elongation that, at t_v (v_2), is 63% larger at HS than at NS. The WB SA, at t_{so}, is endowed with a forward velocity (ω_{so}) while decelerating. At NS, the SA couple, aided by the initial momentum, controls backward acceleration by counterbalancing the action of gravity. At HS,

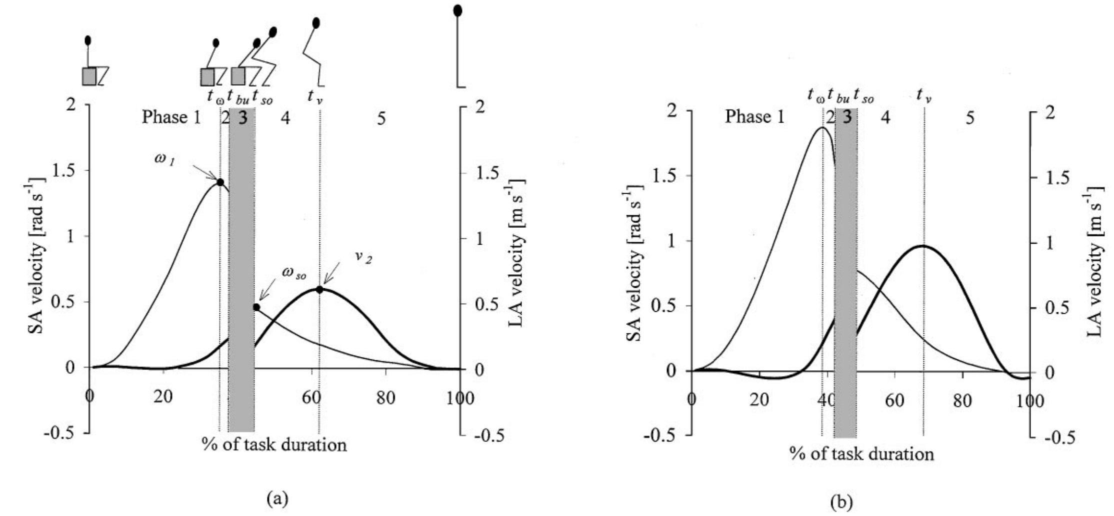

Fig. 3. Linear (thick line) and angular (normal line) velocity of the rotational (SA) and linear (LA) actuator at NS (a) and HS (b) of execution of STS for trials of the same subject. See Fig. 1 for sign convention.

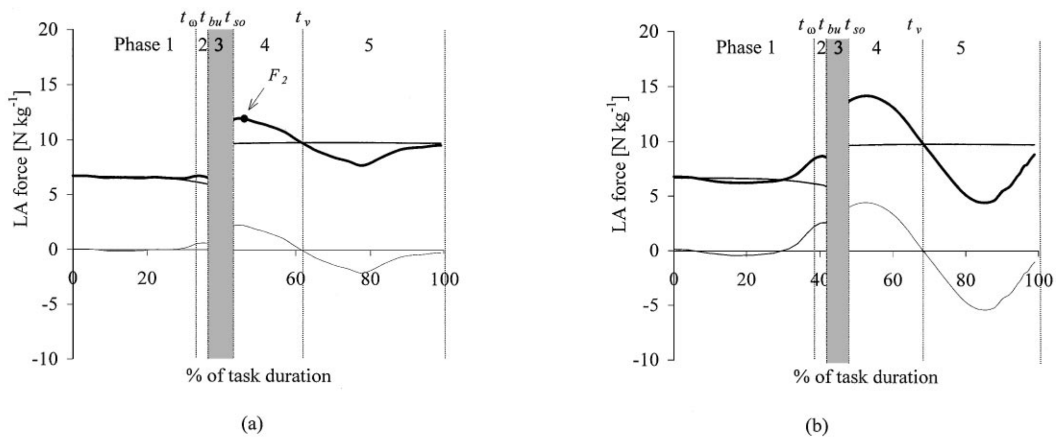

Fig. 4. Inertial (thin line) and gravitational (normal line) contribution to the LA normalised force (thick line) at NS (a) and HS (b) of execution of STS for trials of the same subject. See Fig. 1 for sign convention.

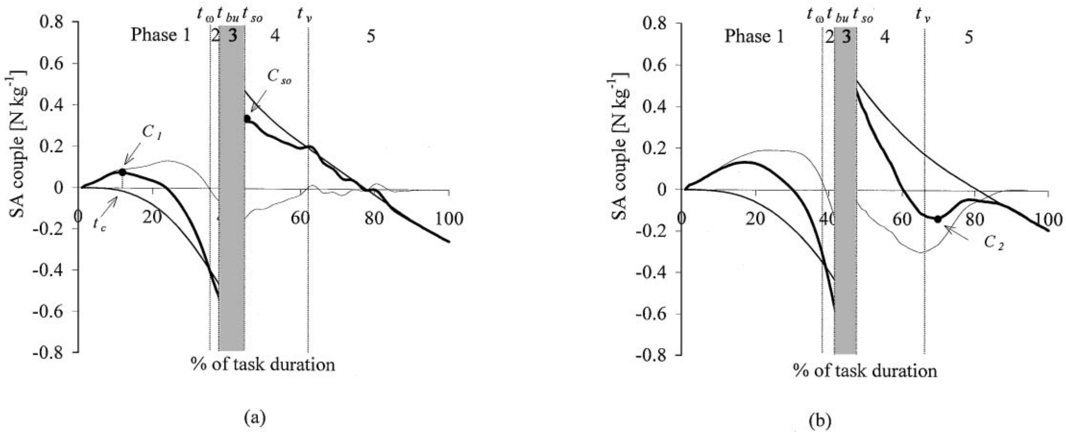

Fig. 5. Inertial (thin line) and gravitational (normal line) contribution to the normalised SA couple (thick line) at NS (a) and HS (b) of execution of STS for trials of the same subject. The inertial contribution equals the inertia couple, the gravitational contribution is opposite to the gravity force moment (see Eq. (2)). See Fig. 1 for sign convention.

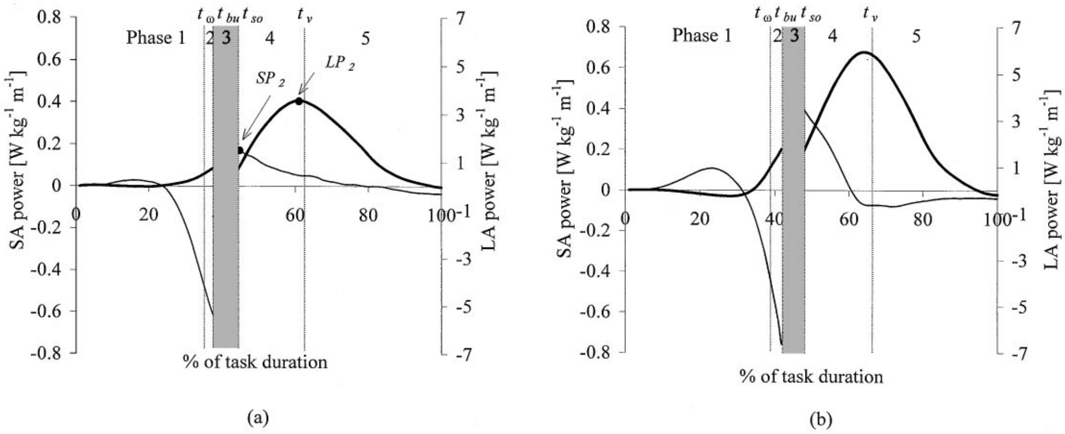

Fig. 6. LA (thick line) and SA (normal line) normalised power at NS (a) and HS (b) of execution of STS for trials of the same subject.

the SA couple, which exhibits a peak value (C_{so}) not significantly different from that at NS (positive power, SP_2 in Table 1), decreases rapidly to zero (around 60% of T), and prevents the immediate onset of a high backward acceleration. This couple action helps a higher speed of execution of the motor task and it is not necessarily required for gaining balance in the vertical position. For this latter aim the WB momentum at seat-off might be sufficient. This consideration is reinforced by the fact that, during the following phase, the SA has to perform a braking action in order to avoid an excess in forward angular velocity when the vertical position is reached (see below). Note that, at both speeds, C_{so} is smaller than the gravity force moment owing to a synergetic inertia couple. Since lift-off would be impossible at zero initial momentum, the gravitational moment must be greater than maximal muscular moment. This further emphasises that lift off is made possible by momentum transfer, irrespective of speed.

Phase 5: *WB linear deceleration* ($t_v - T$). The link approaches the vertical orientation that, as seen from the zero-crossing of the gravitational component of the SA couple (Fig. 5), occurs around 80% of T irrespective of speed. Before this instant of time, at NS the SA couple counterbalances gravity in a quasi-static fashion (angular acceleration is close to zero). On the contrary, at HS, in order to achieve rotational deceleration, the SA is required to act with a braking couple (C_2) synergetic with the gravitational backward action. The remaining 20% of T is characterised, at both speeds, by a quasi-static leaning forward of the model under the control of a backward SA couple while elongation comes to an end.

5. Discussion and conclusion

5.1. TIP model vs. multi-segment models

In phases 1 and 2 (indicated in the literature as a single phase: "flexion momentum" in Schenkman et al., 1990; Ikeda et al., 1991 and Riley et al., 1991 or "forward momentum generation" in Kralj et al., 1990), the HAT SA couple action is consistent with the hip muscle moment reported in Kelley et al. (1976) and Pai and Rogers (1991). The gravitational force action is described by Kelley et al. (1976) in the same terms as described here. The HAT SA kinematics, inclusive of the values of θ_{bu} and t_ω (Table 1), corresponds well with the torso movement illustrated in Schenkman et al. (1990) and Ikeda et al. (1991). The LA elongation, occurring in the final part of phase 1 and during phase 2, is most likely associated with an increase of lumbar lordosis (trunk to pelvis movement referred to by Schenkman et al., 1990). Little information is provided in the literature concerning the specific muscles involved during these phases. Only the erector spinae onset is reported to occur at the end of phase 2 and partially participating in braking the HAT flexion (Millington et al., 1992; Vander Linden et al., 1994; Coghlin and McFadyen, 1994).

The existence of a momentum transfer, from HAT to WB during phase 3, has been reported by Riley et al. (1991) in terms of horizontal to vertical momentum transfer. The TIP model as well highlights this transfer, however it indicates more clearly that it is aimed at forward WB rotation, that is at gaining balance, and not at body elongation. In addition, the TIP model indicates that elongation is due to the LA action that peaks from 0.12 to 0.18 s after seat-off (Table 1). The latter actuator, which is the major responsible of CM ascent, does not benefit from any significant momentum transfer during phase 3.

The SA couple and LA force in phase 4 are consistent with knee extension and ankle plantar-flexion, and hip and knee extension muscular moments, respectively, as reported by Kelley et al. (1976) and Pai and Rogers (1991). Joint kinematic data produced by Schenkman et al. (1990) indicate that v_2 corresponds to maximal hip and knee extensional velocity.

As far as phase 5 is concerned, comparison with the literature does not lead to any particular observation.

5.2. Motor strategy variation with speed

As seen in the previous sections, the TIP model reveals a motor strategy variation associated with the augmentation in speed of execution of STS. In summary it shows that:

1. Elongation velocity increases more than angular velocity (v_2 and ω_1 in Table 1, respectively). This indicates an overall kinematic pattern variation.
2. Prior to seat-off, at HS the torso can be kept more erect than at NS and this reduces the angle spanned by the joints involved and, thus, contributes to the faster speed.
3. At HS, as opposed to NS, the SA couple exhibits a braking peak largely before upright posture is reached. In fact, HS imposes a forward angular velocity immediately after seat-off that requires deceleration to avoid excessive forward leaning.

5.3. Conclusion

The TIP model has allowed a detailed description of the STS mechanics and of the relevant motor strategy modifications associated with speed variation. This entails a positive answer to both questions posed in the introduction. For a global evaluation of the motor act, as required in the application context previously referred to, the TIP model has shown to be no less informative than the more demanding multi-segment models reported in the literature. In addition, the fact that the model did not embed the inertia effects of the involved body portion mass distribution did not hinder the achievement of this result. Although it is true that specific musculo-articular function can only be inferred, the more compact information yielded by the TIP model, is expected to facilitate subject and/or disability classification.

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