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Determination of joint efforts in the human body during maximum ramp pushing efforts

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Abstract

Determining with accuracy, the internal efforts in the human body is a great challenge in Biomechanics, particularly in Physical Therapy and Ergonomics. In this context, the present study develops a human body model that permits a non-invasive determination of the joint efforts produced by a seated subject performing maximum ramp pushing efforts. The joint interactions during these experiments are provided by a dynamic inverse model of the human body, using a symbolically generated recursive Newton–Euler formalism. The theoretical investigation is presented in two steps, with increasing complexity and relevance:

1. Quasi-static analysis: this approximates internal joint efforts, using static equations at each sample, without taking the postural chain dynamic effects into account.
2. Dynamic analysis: this takes the dynamic effects into consideration and thus presents the advantage of a more relevant description of the motion as well as a more accurate determination of the forces and torques produced at each joint during the transient effort.

The dynamic model confirms some previous studies of the effects of biomechanical factors on the performance of the task and is proposed as an accurate method for determining the joint efforts in dynamic contexts.

Finally, this application is a preliminary benchmark case that will be extended to:

- physical therapy, in order to analyse the joint and muscle efforts in various motion contexts, particularly for patients with fibromyalgia and patients with lumbar diseases;
- accidentology, in order to analyse and simulate car occupant dynamics before a crash.

Keywords: Joint efforts; Inverse dynamics; Multibody; Adjustment; Posture

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1. Introduction

The first static analysis of postural adjustments during maximum ramp pushing efforts was proposed by Gaughran and Dempster (1956), stating that postural dynamics are associated with upper-limb "*focal*" motions (Bouisset and Zattara, 1981; Zattara and Bouisset, 1988): during maximum ramp pushing efforts, the body balance is perturbed by the focal motion, and this perturbation is counteracted at each instant by the postural chain dynamics in order to efficiently perform the motion. This counter-perturbation is assumed to depend on the dynamic mobility of the postural chain (Bouisset and Le Bozec, 2002). This mobility can be expressed in terms of displacements of the chain segments and in terms of internal torques and forces at the postural joints (Bouisset and Le Bozec, 2002). More recently, it was shown that the postural chain dynamics are related to the risk of slipping on the support surface during this experiment of maximum ramp pushing efforts (Bouisset et al., 2002; Gaudez et al., 2003).

The aim of this study is to quantify the joint forces and torques associated with maximum ramp pushing efforts, by developing a dynamic model using non-invasive measurements of the system variables and of the external forces and torques applied to the body. The corresponding velocities and accelerations are determined numerically. In the context of these experiments, the seated subject is asked to produce as rapidly as possible a maximal ramp pushing effort with both hands on a horizontal freely rotating dynamometric bar. This force is referred to as the "*synthetical voluntary maximal force*", defined by Bouisset and Maton (1996) and Bouisset (2002). Once the synthetical voluntary maximal force is reached, the subject is asked to maintain this effort as constant as possible. During this experiment, the dynamic analysis will be compared to the quasi-static analysis and to the dynamic model at equilibrium described by Gaughran and Dempster (1956).

2. Material and methods

During the entire experiment, the subject is seated (Fig. 1). The subject is asked to produce a maximal as rapidly as possible, and then to maintain this force for 5 s after the transient phase. During the experiment, three distinct phases can be observed:

1. the initial phase: the seated subject is in an equilibrium state, only performing efforts to maintain his initial posture;
2. the transient phase: the subject performs a ramp pushing effort on the bar;
3. the final phase: the subject maintains his maximal pushing effort on the bar; this is the second equilibrium state of the experiment.

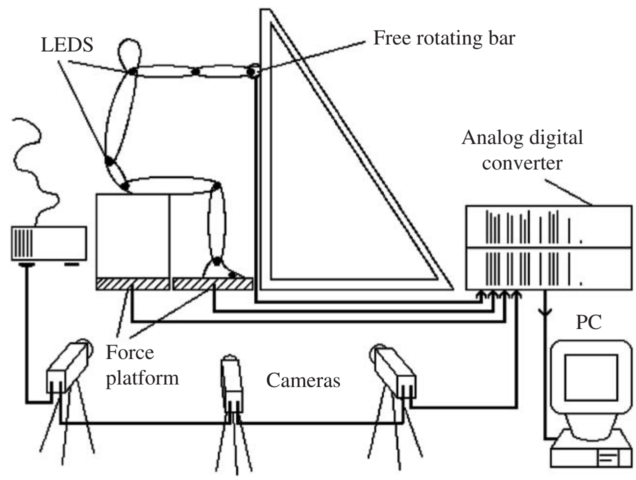

Fig. 1. The motion measurement set-up, i.e. the light emitting diodes (LEDs) and cameras, synchronized with the interaction measurement set-up, i.e. the force platforms and the freely rotating bar.

The dynamic model will be discussed for all phases in the paper.

The motion measurement set-up consists of optokinetic sensors (*Selspot*$^{\mathrm{TM}}$), i.e. light emitting diodes (LEDs) and three cameras (Fig. 1), for the estimation of the coordinates (q) of joint reference points. The interaction measurement set-up consists of two force platforms and one freely rotating bar (Fig. 1), all using piezoelectric sensors and strain gauges (Heglund, 1981) designed by our laboratory, for the determination of the horizontal and vertical interaction forces ($\mathbf{F}_{\mathrm{ext}}$) and torques ($\mathbf{M}_{\mathrm{ext}}$) between the body and its environment.

The maximal ramp pushing experiments were performed by one female and five male individuals. All the subjects were related to our laboratory and gave their informed consent to perform the experiments. The data were sampled at 250 Hz and were filtered by a 25 Hz low-pass numerical filter. These data were highly reproducible from one test to another and followed the same profiles from one subject to another.

In order to discuss the ability of the mathematical model to take into account the dynamic effects, all the results presented in the paper are taken from one test performed by one subject. Practically, the experiment presented here was performed on rough surfaces at the seat and feet (dynamic friction coefficient between clothes and surfaces: $\mu_{\mathrm{D}} = 0.35$). Furthermore, the subject was seated at 30% of his maximal possible thigh seating surface height, essentially corresponding to the ischial contact, so that the subject could develop greater thigh contact forces than with a maximal thigh seating surface height (Bouisset et al., 2002).

From these data, a model determining internal efforts can be established on the basis of the method of fictitious sections (Fig. 2). The problem is assumed to be purely two-dimensional in the sagittal plane. The device provides a total number of six force measurements in

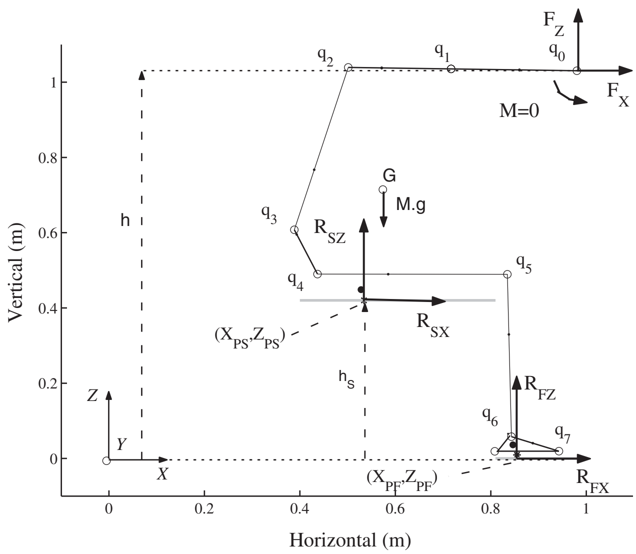

Fig. 2. Method of fictitious sections: the eight body segments articulated around the eight joints (from q_0 to q_7); the seat and feet centre of pressure coordinates ((X_{PS}, Z_{PS}) and (X_{PF}, Z_{PF}), respectively); the external forces applied to the body by the seat support (R_{SX}, R_{SZ}), the feet support (R_{FX}, R_{FZ}) and the bar (F_X, F_Z); the gravitational force $M \cdot g$ applied to the body; the heights of the seat (h_S) and the bar (h) from the ground.

order to obtain a statically determined structure: these external forces are applied to the body by the seat support (R_{SX}, R_{SZ}), the feet support (R_{FX}, R_{FZ}) and the bar (F_X, F_Z). The model considered here consists of eight body segments, labeled q_0–q_7 in Fig. 2: lower arm, upper arm, trunk and head, pelvis, thigh, shank, instep and foot sole. The segments are articulated around nine anatomical landmarks: wrist, elbow, shoulder, lumbosacral hinge, greater trochanter, lateral condyle, lateral malleolus, head of fifth metatarsal, heel. The system is fully described by a total of 8 *bodies* × 3 *variables* − 7 *rotoid joints* × 2 *link constraints* = 10 *generalized coordinates*, representing the 10 degrees of freedom of the model. The system referential frame is located at a fixed point on the laboratory floor.

For a global analysis, the dynamic equations of the mechanical model are developed on the basis of a preliminary estimation of the centres of mass (CM) and pressure coordinates.

2.1. Preliminary calculation

In the system considered, the global posture depends on the coordinates of the various CM, and the external forces are applied at the centres of pressure (CP).

The centre of mass of a system of material elements, G, is the barycentre of its elements G_i of mass m_i. Its expression related to the system of reference O is given by

$$\overrightarrow{OG} = \frac{\sum_i m_i \overrightarrow{OG_i}}{\sum_i m_i}, \tag{1}$$

where the masses m_i and the positions of the CM of the various body segments can be estimated from tables of inertia (De Leva, 1996).

The centre of pressure coordinates, related to the reference point of a body (i.e. G_i here), can be determined from force platforms, using the following definition:

$$\overrightarrow{G_iP_i} = (X_{P_i}, Z_{P_i}) = \left(-\frac{M_{Y_i}}{R_{P_{i,z}}}, \frac{M_{X_i}}{R_{P_{i,z}}}\right), \tag{2}$$

where

- M_{X_i} and M_{Y_i} are anterior–posterior and lateral components of the resulting torque related to the reference G_i, respectively;
- $R_{P_{i,z}}$ is the vertical component of the force at the body centre of pressure.

Both quasi-static and dynamic analyses can be developed using these definitions of CM and pressure.

2.2. Theoretical investigation

The equations of the model, using fictitious sections, are systematically presented for each body segment i, using the external forces applied to this segment and the generalized forces and torques produced by the adjacent segments (Fig. 3). A quasi-static analysis and a dynamic analysis are described in the following two sections, respectively.

2.2.1. Quasi-static model of internal efforts

Referring to Fig. 3, the quasi-static equations for each body segment i can be written as follows:

$$F_{i,x} = -F_{i-1,x} - R_{P_{i-1,x}}, \tag{3}$$

$$F_{i,z} = m_i \cdot g - F_{i-1,z} - R_{P_{i-1,z}}, \tag{4}$$

$$\begin{aligned} M_i = & -M_{i-1} + Z_{P_i} \cdot R_{P_{i,x}} - X_{P_i} \cdot R_{P_{i,z}} \\ & + (-l_{1,i} \cdot F_{i-1,x} + l_{2,i} \cdot F_{i,x}) \sin\theta_i \\ & + (l_{1,i} \cdot F_{i-1,z} - l_{2,i} \cdot F_{i,z}) \cos\theta_i, \end{aligned} \tag{5}$$

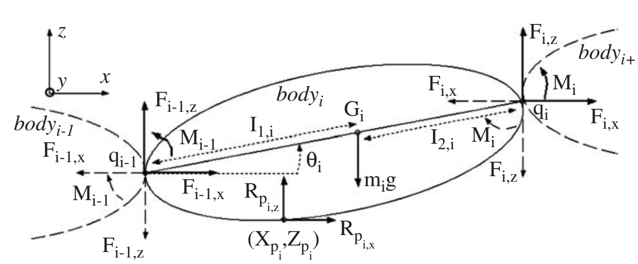

Fig. 3. Body i and adjacent bodies: external forces ($m_i \cdot g, R_{P_{i,x}}, R_{P_{i,z}}$), generalized forces ($F_{i-1,x}, F_{i-1,z}; F_{i,x}, F_{i,z}$) and torques ($M_{i-1}; M_i$) produced by the previous segment and by the following one (indexes $i-1$ and i, respectively); centre of pressure coordinates (X_{P_i}, Z_{P_i}); distances $l_{1,i}$ and $l_{2,i}$ of the centre of mass G_i from the joints q_{i-1} and q_i, respectively; angle θ_i between the $[q_{i-1}, q_i]$ segment and the inertial frame (X, Y, Z).

where $R_{P_{i,x}}$ and $R_{P_{i,z}}$ equal zero for all segments except for the thighs and the feet, lying on their support.

A more accurate determination of joint efforts in the human body requires a dynamic model of the system in order to take the body segment inertia parameters into account.

2.2.2. Dynamic model of the transient efforts

For the problem dynamics, we introduce the moments of inertia given by the tables of inertia of De Leva (1996). The corresponding dynamic equations are obtained from a Newton–Euler formalism and a Newton–Raphson iterative algorithm (Fisette et al., 2002): this algorithm provides the vector Q of internal interaction torques and forces applied at the joints for any configuration as a function of $(q, \dot{q}, \ddot{q})$, q being the vector of the human body generalized coordinates. The vector $Q(q, \dot{q}, \ddot{q})$ can be directly obtained from the equations of motion for the eight body segments:

$$\begin{aligned} Q &= f(q, \dot{q}, \ddot{q}, \mathbf{F}_{\mathrm{ext}}, \mathbf{Q}_{\mathrm{ext}}, g) \\ &= M(q)\ddot{q} + G(q)u(\mathbf{F}_{\mathrm{ext}}, \mathbf{Q}_{\mathrm{ext}}) - g(q), \end{aligned} \tag{6}$$

where

- $M(q)$ (8×8) is the positive-definite symmetric inertia matrix;
- u (8×1) is the vector of external forces and torques applied to the system;
- $G(q)$ (8×8) is the kinematic matrix indicating the corresponding body segments to which each external force and torque is applied;
- $g(q)$ (8×1) is the vector of forces and torques resulting from gravity.

In order to validate the dynamic model, the input force values at the rotating bar were recalculated using an inverse dynamic model with constraints, and were equal to the actually measured input forces at the bar, with a mean error approximately equal to 0.01%. Further, the analytical solutions of both dynamics and quasi-statics were checked to be equal when the velocities and accelerations were set to zero.

3. Results

On the basis of the convention of vector orientations defined in Fig. 3, the dynamic analysis provides the time evolution of forces and torques in the joints (Fig. 4). A comparison of dynamic and quasi-static generalized forces and torques can be obtained (Fig. 5) for the joints considered in Section 2. Further, the time evolution of the global CM and pressure is also provided by the model (Fig. 6). Let us remind that the results from Figs. 4–7 are taken from one trial performed by one subject, in order to discuss the ability of the mathematical model to take into account the dynamic effects. In Figs. 4–6, the two vertical dotted lines separate the initial and transient phases at $t = 0.16$ s, and the transient and final phases at $t = 0.82$ s, respectively. By convention, these transitions are based on the time evolution of the horizontal force produced at the wrist, which is the "focal" perturbation described in Section 1.

Furthermore, a real-time body segment animation was developed in order to show the evolution of the joint positions and the local and global CM and pressure. Fig. 7 shows the initial and final configurations of this body segment animation (see also file Animation.avi).

Finally, various body segment accelerations were measured using accelerometers during the experiments (Fig. 8), in order to observe whether the segment accelerations are negligible or not.

4. Discussion and conclusion

From the above results, we can analyse the three phases of this experiment for both dynamic and quasi-static approaches:

- During the initial phase, the configuration and efforts can be considered as being practically constant. This is an equilibrium state, and consequently the dynamic and quasi-static forces and torques in the joints are approximately equal.
- During the transient phase, the forces increase up to a maximum value. Furthermore, small postural adjustments and small relative motion of the body segments are observed. By geometry, the agonistic muscles fixed to the moving body segments decrease their length. Thus, it is assumed that these agonistic muscles are subject to anisometric contraction, producing the observed motion. This phase is analysed in Section 4.1, comparing the quasi-static and dynamic analyses.
- During the final phase, the configuration and efforts can be considered as being practically constant. In consequence, as the segments to which the muscles are fixed do not move, all the muscular contractions can be considered as isometric. In this phase, the dynamic and quasi-static values are approximately equal. In Section 4.2, we will compare these values to the dynamic model at equilibrium described by Gaughran and Dempster (1956).

Finally, Section 4.3 presents the limitations and perspectives of this model for the human body.

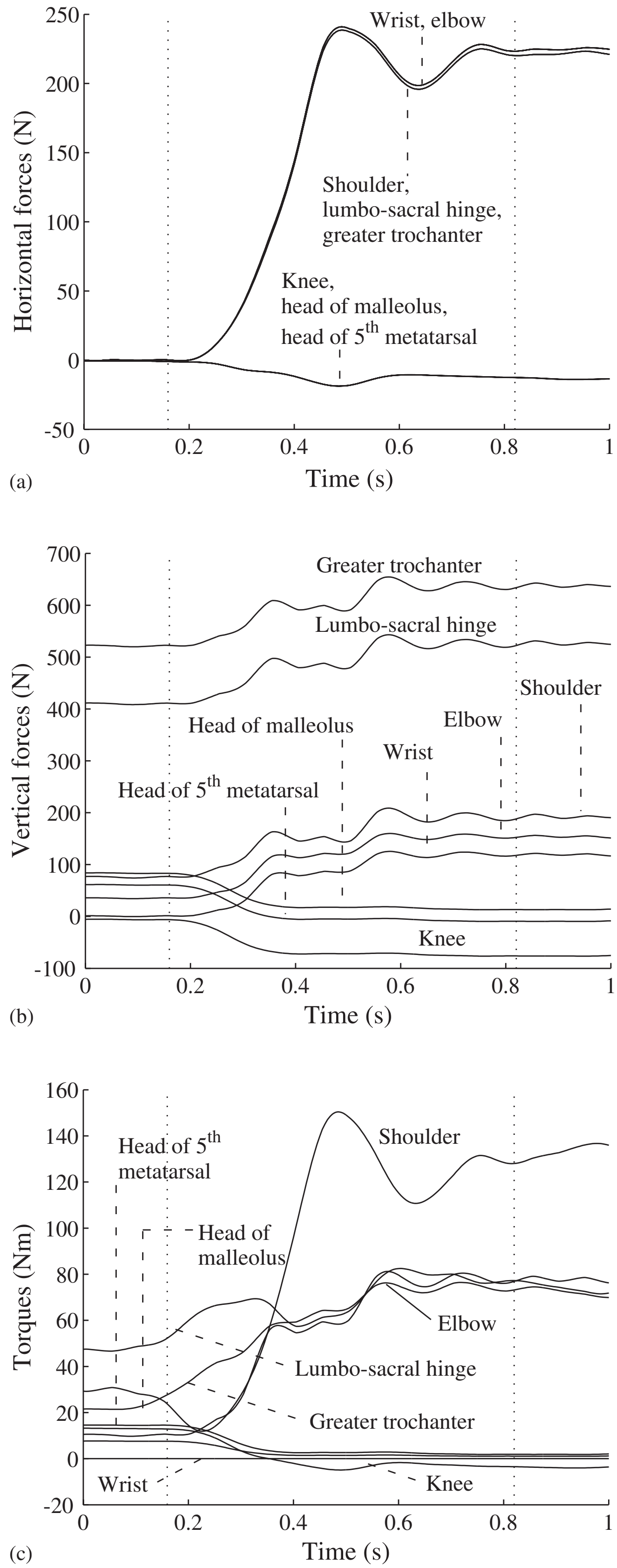

Fig. 4. Time evolution of horizontal (a) and vertical (b) forces, and torques (c) in the joints. The two vertical dotted lines separate the initial and transient phases at $t = 0.16$ s, and the transient and final phase sat $t = 0.82$ s, respectively.

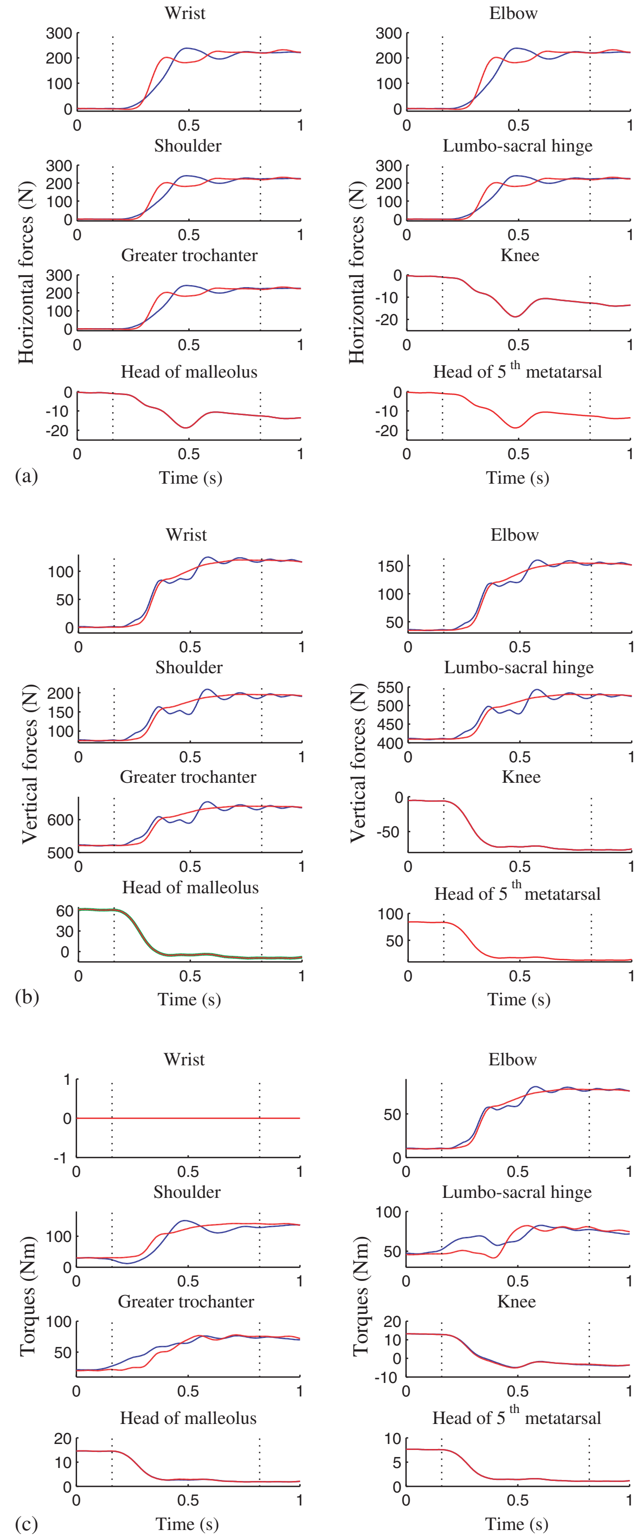

Fig. 5. (Online color). Dynamic (blue) and quasi-static (red) plots of horizontal forces (a), vertical forces (b) and torques (c) in the joints. The two vertical dotted lines separate the initial and transient phases at $t = 0.16$ s, and the transient and final phases at $t = 0.82$ s, respectively.

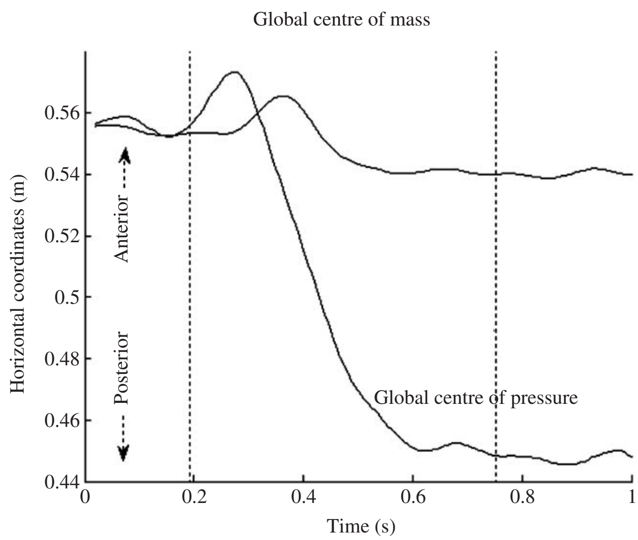

Fig. 6. Time evolution of the global centres of mass and pressure. The two vertical dotted lines separate the initial and transient phases at $t = 0.16$ s, and the transient and final phases at $t = 0.82$ s, respectively.

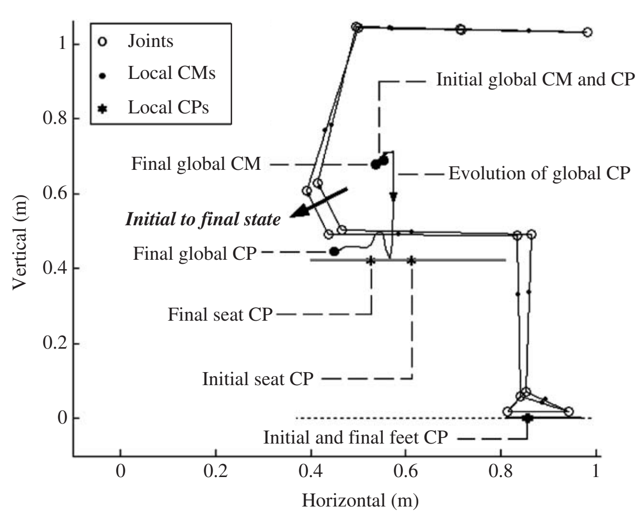

Fig. 7. Initial and final configurations of the body segments, the corresponding joints and local centres of mass (CM); trajectory of the global CM and also the local and global centres of pressure (CP).

4.1. Quasi-static and dynamic analyses during the transient phase

First, the displacement of the global centre of mass (Figs. 6 and 7) can be considered as negligible compared to the displacement of the global centre of pressure, in accordance with Bouisset et al. (2002). Secondly, the dynamic results in Fig. 5 closely match the corresponding quasi-static values. Thirdly, the quasi-static model is easier to implement than the dynamic one. For these three reasons, the quasi-static model could be considered as sufficient to describe this experiment. Nevertheless, the quasi-static model is an approximation of

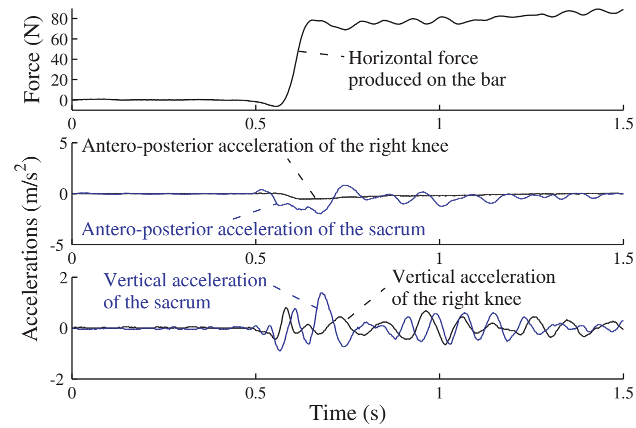

Fig. 8. Time evolution of the horizontal force produced on the bar and of four corresponding accelerations measured using accelerometers: antero-posterior and vertical accelerations of one knee and of the sacrum.

the experiment because it does not take into account any body postural chain dynamic factors such as moments of inertia. The various body segment accelerations collected by accelerometers during the experiments (Fig. 8) confirm this assumption: these data show non-negligible horizontal accelerations of the knees and the sacrum during rapid pushing efforts, and specifically in the transient phase. Indeed, the corresponding maximal $\|m_i \cdot \ddot{q}\|$ terms for the sacrum and the knees are equal to 22.4 and 9.7N, respectively: consequently, these terms can contribute up to approximately 28.3% and 12.2%, respectively, of the horizontal force jump produced on the bar, which is non-negligible.

In conclusion, even if the experiment is considered as quasi-static (Bouisset et al., 2002), a representation of the anisometric effort during the transient phase of increasing force requires defining a dynamic model taking the various body segment accelerations into consideration. Typically, the differences between quasi-static and dynamic approaches in the transient phase (Fig. 5) are due to these accelerations. Consequently, the dynamic analysis is more accurate than the quasi-static analysis: since the dynamic approach takes the dynamic effects into consideration, it permits a more relevant description of posture and motion as well as an accurate determination of the forces produced in a specific joint; this supports the assumption proposed by Bouisset and Le Bozec (2002) that the counter-perturbation of the ramp pushing effort depends on the dynamic mobility of the postural chain, associated with muscular torque in the postural joints.

4.2. Comparison with the dynamic model in the final phase

The dynamic model at equilibrium of the final phase describes the postural adjustments during the synthetical voluntary maximal force (redefined in Section 1). This synthetical voluntary maximal force can be estimated by maximizing the absolute value of the horizontal pushing force F_X, in terms of the other external forces applied to the system. The expression of $\|F_X\|$ is adapted from Gaughran and Dempster (1956) and refers to Fig. 2:

$$\|F_X\| = \frac{d \cdot Mg - d' \cdot R_{FZ} + h_S \cdot R_{SX} - l_W \cdot F_Z}{h}, \qquad (7)$$

where we define:

- d, the horizontal distance $X_g - X_{PS}$ between the global centre of mass and the seat centre of pressure;
- d', the horizontal distance $X_{PF} - X_{PS}$ between the feet and seat CP;
- l_W, the horizontal distance between the bar and the seat CP X_{PS}.

By maximizing Eq. (7), it can be concluded that both dynamic and static values confirm the joint efforts predicted qualitatively (Bouisset et al., 2002) to optimize the performance of the synthetical voluntary maximal force:

1. the pelvis turns ($\max d$) favouring the motion of the seat centre of pressure backwards;
2. the feet centre of pressure moves forward with a tendency to push the feet up ($\max(-d' \cdot R_{FZ})$);
3. the ischio-femoral contact moves backwards ($\max R_{SX}$);
4. the vertical force step applied to the bar increases ($\max(-F_Z) = \max(\|F_Z\|)$).

4.3. Limitations of the model and perspectives

The dynamic model is more accurate than the static one in determining joint efforts in dynamic contexts. So it is currently being extended to include mass parameter estimation and more involved joint models.

Another further perspective is to estimate muscle efforts from these joint efforts and from the geometrical structure of the body, in order to compare these calculated muscle efforts to values obtained from electromyography.

The final objective is to develop a three-dimensional model determining joint and muscle efforts of subjects in various dynamic contexts, applied to:

- physical therapy, in order to analyse joint and muscle efforts of subjects, particularly patients with fibromyalgia and patients with lumbar diseases;
- accidentology, in order to quantify the joint forces and torques during road traffic accidents, in particular to analyse and simulate car occupant dynamics before a crash.

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