INFLUENCE ON INVERSE DYNAMICS CALCULATION OF DETAILING THE HEAD-ARMS-TRUNK SEGMENT

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Abstract: The reliability of internal joint torque calculation in gait analysis during daily living activities is fundamental for clinical decisions based on joint function. This calculation, obtained by means of the inverse dynamics, depends on several modelling factors such as assumptions on the segments and on the relevant joints constituting the kinematic chain. In this study the influence of the number of the rigid segments constituting the kinematic chain was investigated. Four models with different detailing of the head-arms-trunk segment were compared during stair climbing, sitting and rising from a chair, stepping up/down and squatting motor tasks. Considering the pelvis as a separate rigid segment produced a more reliable results. While adding more details on the last segment of the kinematic chain, produced no improvements or worsening in sitting and rising from a chair and stepping. But for motor tasks where some motion between parts of the trunk could occur such as squatting a separation in more rigid segments is necessary.

Introduction

The calculation of three-dimensional internal joint torques during daily living activities by means of the inverse dynamics approach is a traditional issue in human movement analysis. Lower limb joint torques are used to evaluate the deficit resulting from pathologies or the efficacy of treatments in terms of joint function. With the inverse dynamics approach the human body is represented as a chain of rigid segments. Newton-Euler mechanics is applied to each segment of the model to calculate net internal joint torques and forces. The reliability of the model outputs is fundamental to produce relevant results for clinical decision. The calculation depends on several modelling factors such as hypotheses on the segments and on relevant joints constituting the kinematic chain, number of these segments and type of anthropometric data.

In a previous study [1], the effect of inertial parameters was investigated during the stair ascending and descending of ten young subjects. The root mean square value over the step cycle of the difference between joint torques, calculated at the lower limb with five different sets of inertial parameters expressed in percentage of their corresponding range was computed. The maximum root mean square value was 21.8% in the internal/external rotation torque at the hip [1]. In order to achieve accurate estimate of lower limb joint torques other factors should be investigated rather than optimal inertial parameter set.

In this study, the influence of the number of the rigid segments constituting the kinematic chain was investigated. The purpose was to strengthen the rigidity hypothesis on the last segment. Four models representing the human body were compared during stair climbing, sitting and rising from a chair, stepping up/down and squatting motor tasks. The models differed in detailing the head-arms-trunk segment of the kinematic chain with which the human body is represented.

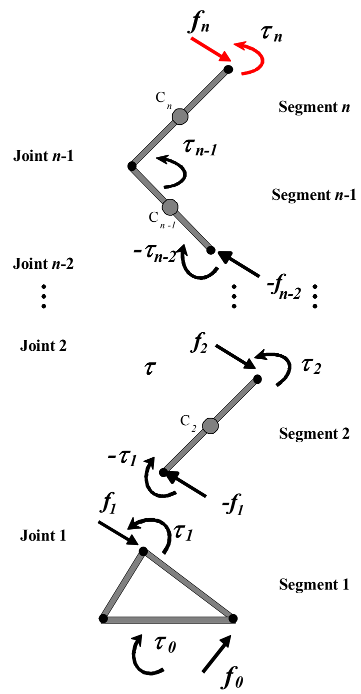

Figure 1: 3D model for the calculation of net internal joint torques and forces using the inverse dynamics approach.

Materials and Methods

Theoretical background. In order to study the dynamics of human body motion the Newton-Euler formulation is used. This formulation comprehends a set of differential equations that involves motor forces (forces and torques), kinematic data (velocities and accelerations), and inertial properties (mass, centre of mass position and moment of inertia). Using a direct dynamic approach, these equations are solved estimating the kinematic variables in order to study the strategies of the central nervous system in human motion control. Using an inverse dynamic approach, the equations are solved estimating the net internal forces and torques at the joints to find a better comprehension of the motor control mechanisms in human locomotion. The inverse dynamic approach uses the kinematic data, measured with a stereo-photogrammetric system, the inertial parameters, estimated for example with regression equations, and the ground reaction torques and forces, measured with force plates.

The human body is represented as a 3D kinematic chain of rigid segments connected by ball-and-socket joints (Figure 1). Once the initial conditions are known, the Newton-Euler formulation is applied iteratively from the feet segments to the last segment of the kinematic chain (Figure 1). The iteration can be applied from the last segment to the feet segments also, however in this case the ground reaction forces and torques are not used as input variables but only as a control variables and thus the net internal joint forces and torques calculated are less reliable.

From a free-body diagram of the *i*-th rigid segment, the joint internal net torques and forces are estimated using the following equations

$$\vec{f}_{i+1} = \vec{f}_i + m_i \cdot \vec{g} - m_i \cdot \ddot{\vec{p}}_{C_i} \quad (1)$$

$$\vec{\upsilon}_{i+1} = \vec{\upsilon}_i + \vec{f}_i \times \vec{r}_{i,C_i} - \vec{f}_{i+1} \times \vec{r}_{i+1,C_i} - \vec{I}_i \cdot \dot{\vec{\omega}}_i - \vec{\omega}_i \times \left(\vec{I}_i \cdot \vec{\omega}_i\right) \quad (2)$$

where, m_i is the mass of the segment, I_i is the inertia matrix of the segment, $\vec{g}$ is the acceleration of gravity, $\vec{f}_i$ is the force applied from the *(i-1)*-th segment to the *i*-th segment, $\vec{f}_{i+1}$ is the force applied from the *i*-th segment to the *(i+1)*-th segment, $\vec{\upsilon}_i$ is the torque applied from the *(i-1)*-th segment to the *i*-th segment, $\vec{\upsilon}_{i+1}$ is the torque applied from the *i*-th segment to the *(i+1)*-th segment, $\ddot{\vec{p}}_{C_i}$ is the linear acceleration of the segment centre of mass, $\vec{\omega}_i$ is the segment angular velocity, $\dot{\vec{\omega}}_i$ is the segment angular acceleration, $\vec{r}_{i,C_i}$ is the vector that goes to the segment centre of mass from the *i*-th reference system, and $\vec{r}_{i+1,C_i}$ is the vector that goes to the segment centre of mass from the *(i+1)*-th reference system.

Experimental study. Movement analysis was performed using a stereo-photogrammetric system (e-Motion, Padova, Italy) and two force plates (Bertec Corporation, Columbus, OH, USA). Rigid plates mounting 4 markers were attached with straps to each bony segment, a number of anatomical landmarks were calibrated and segment anatomical frames were defined [2]. A-model was a chain of seven rigid segments (feet, shanks, tights and a single segment for head-arms-trunk-pelvis (HAT)); B-model was a chain of eight rigid segments (pelvis was considered separated from the trunk and L5 as a ball-and-socket); C-model was a chain of nine rigid segment (head was considered separated from the trunk and C7 as a ball-and-socket); D-model was a chain of ten rigid segment (the trunk was considered separated in two parts and T9/T10 as a hinge). In all models relevant anthropometric data were taken from Zatsiorsky after DeLeva [3]. Newtonian mechanics was applied to each segment to calculate net internal joint torques and forces, starting from the feet (Figure 1, Equations 1-2). The residual torque, τ_n (Figure 1), to the last segment of the kinematic chain (HAT for A-model, HAT without the pelvis for B-model, head for C- and D-models, figure 2) was considered as a quality parameter of the models.

Two acquisition sessions were performed. In the first one, one young male subject (height 1.73 m and weight 67 Kg) was analysed during a stair ascending motor task with only two models: A-model and B-model. In the second session, another young male subject (height 1.78 m and weight 75 Kg) was analysed during sitting and rising from a chair, stepping up/down and squatting using all four different models (Figure 3).

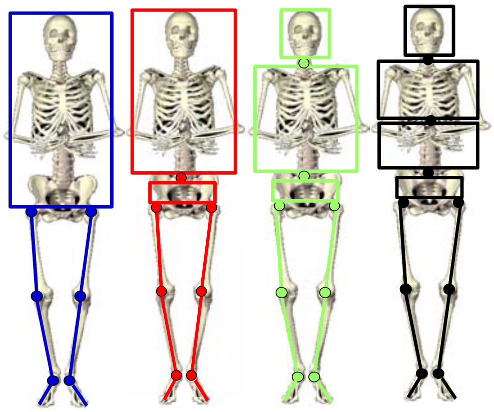

Figure 2. The four models with which the human body was represented: A-model (blu, seven rigid segments), B-model (red, eight rigid segments), C-model (green, nine rigid segments), and D-model (black, ten rigid segments).

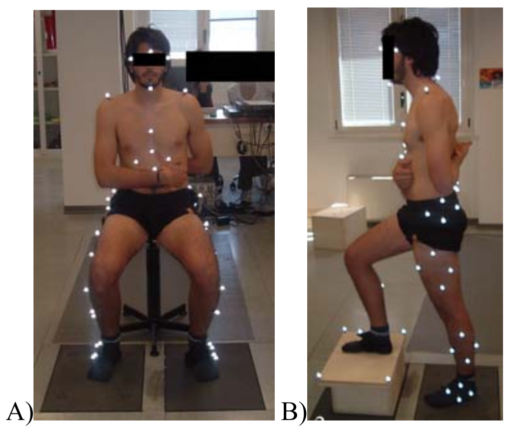

Figure 3. Experimental set-up of the second acquisition session for sitting and rising from a chair (A) and stepping up/down (B).

Results

In all four models and in both acquisition sessions, the maximum value of the residual torque was of the same order of magnitude as net internal joints torques in the lower limb.

Table 1: First study. Mean (minimum÷maximum) values of the three components of the residual torque normalized in percentage of the body weight [N] multiplied by the height [m].

	Antero-posterior	Vertical	Medio-lateral
A- model	0.26 (-4.35÷3.91)	-0.16 (-1.33÷0.83)	-3.00 (-11.28÷7.99)
B-model	0.8 (-2.72÷5.18)	0.17 (-0.66÷0.97)	0.58 (-4.89÷10.20)

Table 2: Second study. Mean (range) values of the three components of the residual torque normalized in percentage of the body weight [N] multiplied by the height [m] for sitting and rising from a chair motor task.

	Antero-posterior	Vertical	Medio-lateral
A- model	-0.22 (0.85)	-1.28 (3.05)	-0.62 (4.03)
B-model	-0.26 (0.96)	-1.28 (2.43)	-0.66 (3.46)
C-model	-0.27 (0.98)	-1.30 (2.53)	-0.62 (3.60)
D- model	-0.27 (1.81)	-1.27 (3.92)	1.23 (10.33)

Table 3: Second study. Mean (range) values of the three components of the residual torque normalized in percentage of the body weight [N] multiplied by the height [m] for stepping up/down motor task.

	Antero-posterior	Vertical	Medio-lateral
A- model	-0.15 (1.47)	-1.39 (5.54)	-0.56 (4.17)
B-model	-0.15 (1.37)	-1.29 (5.63)	-0.66 (4.03)
C-model	-0.16 (1.39)	-1.41 (5.89)	-0.56 (4.33)
D- model	-0.22 (1.70)	-1.48 (7.65)	0.87 (9.55)

Table 4: Second study. Mean (range) values of the three components of the residual torque normalized in percentage of the body weight [N] multiplied by the height [m] for squatting motor task.

	Antero-posterior	Vertical	Medio-lateral
A- model	-0.51 (1.91)	-2.00 (4.67)	-0.91 (3.91)
B-model	-0.53 (1.66)	-1.94 (3.32)	-1.03 (4.34)
C-model	-0.52 (1.60)	-1.97 (3.57)	-0.93 (4.54)
D- model	-0.40 (1.27)	-1.86 (3.40)	0.58 (6.37)

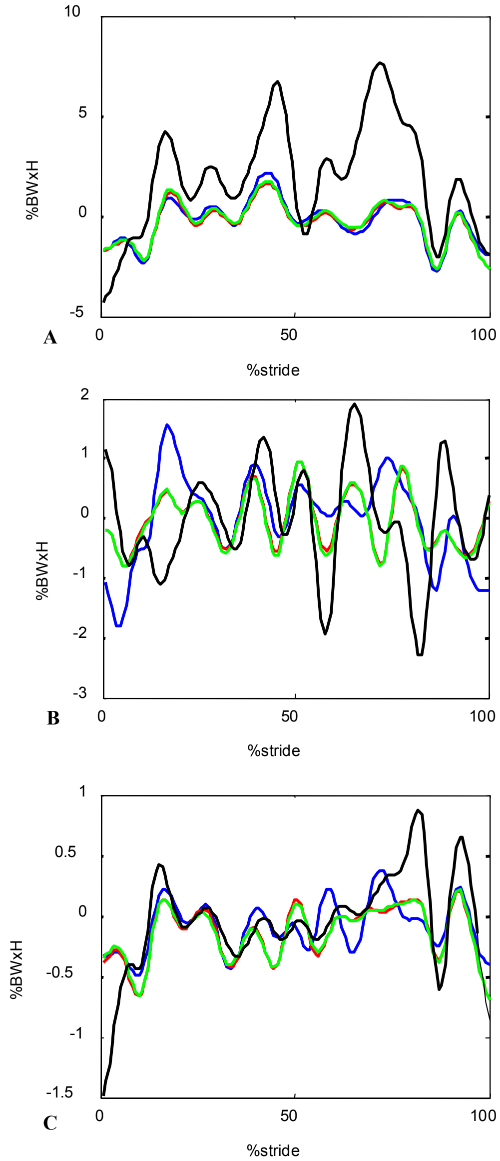

Figure 4. Residual torque to the last segment of the kinematic chain for A-model (blu), B-model (red), C-model (green), and D-model (black) during the sitting and rising from a chair motor task. The three components were reported: medio-lateral (A), antero-posterior (B) and vertical (C). The torque is normalized in percentage of the body weight [N] multiplied by the height [m].

The mean, minimum and maximum values of the medio-lateral, antero-posterior and vertical components of the residual torque obtained with the two models estimated in the first experimental study are reported in Table 1. A slightly reduced mean value can be observed in B-model with respect A- model. The mean value and the range of the three-components of the residual torque were calculated for the second experimental study also (Figure 4 and 5), the results for all four models are reported in Table 2 for sitting and rising from a chair, in Table 3 for stepping up/down and in Table 4 for

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squatting motor tasks. A slight reduction of the residual torque of B-model with respect A- model was observed, particularly for the antero-posterior and vertical components (Tables 2, 3 and 4, and Figure 4). The residual torque obtained with C-model was almost comparable with that obtained with B-model, in all three components and motor tasks. The residual torque obtained considering the trunk as separated in two rigid segment (D-model), showed a higher mean and range values for sitting and rising from a chair and for stepping. Only during squatting motor task the residual torque obtained with D-model showed a smaller mean and range values in all three components (Figure 5 and Table 4).

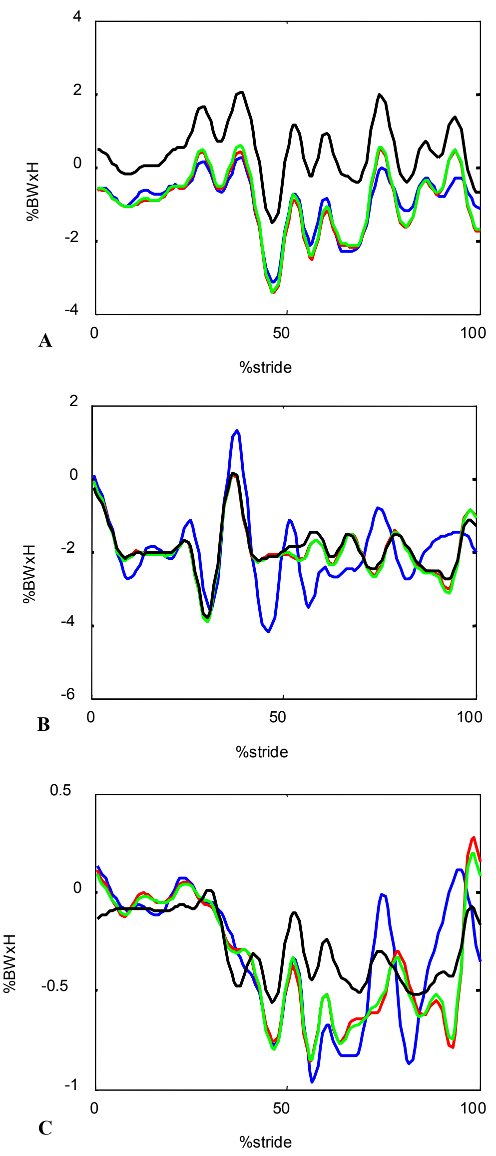

Figure 5. Residual torque to the last segment of the kinematic chain for A-model (blu), B-model (red), C-model (green), and D-model (black) during the squatting motor task. The three components were reported: medio-lateral (A), antero-posterior (B) and vertical (C). The torque is normalized in percentage of the body weight [N] multiplied by the height [m].

Discussion

The influence of the number of the rigid segments constituting the kinematic chain on joint torques calculation was investigated. Zatsiorsky anthropometric data [3] allowed to distinguish between pelvis, head, and different part of the trunk. Four models with different detailing of the HAT segment were compared during stair climbing, sitting and rising from a chair, stepping up/down and squatting motor tasks. The purpose was to strengthen the rigidity hypothesis on the last segment and thus to obtain a more reliable model of the human body trough a reduction of the residual torque. The reliability of the model outputs is fundamental to produce relevant results for clinical decision.

In the first experimental study, the pelvis was considered as a separated rigid segment from the trunk (B-model) and the residual torque was compared with a model that considered the HAT and the pelvis as a unique rigid segment (A-model). The first model showed a slightly reduction of the residual torque (Table 1). This result was confirmed in the second study where other motor tasks were analysed with the same two models (Tables 2, 3 and 4, Figure 4 and 5). Furthermore, in the second study other two models were used: one considered the head as a separate rigid segment (C-model) and another one separated the trunk in two segments (D-model). The C-model produced a residual torque comparable with the one obtained with B-model (Tables 2, 3 and 4, Figure 4 and 5). Thus considering the head as a separate rigid segment produced no improvements in the human body model. The D-model produced a residual torque higher than the other models, in all motor task with the exception of the squatting (Tables 2, 3 and 4, Figure 4 and 5). This could be explained with the difficulty of tracking of the skin markers of the lower part of the trunk. In other words the inputs of the model, the kinematic data of this rigid segment, were not reliable and thus introduced errors in sitting and rising from a chair and in stepping. While, in the squatting motor tasks where the hypothesis of rigidity of the trunk is more critical and some motion between the upper part of the trunk and the lower part could occur, the separation produce a lower residual torque and a better estimate of the human body dynamics (Figure 5).

Conclusions

The calculation of three-dimensional internal joint torques during daily living activities by means of the inverse dynamics approach is a traditional issue in human movement analysis. A reliable model of the human body is fundamental to produce relevant results for clinical decision. Considering the pelvis as a separate rigid segment produced a more reliable results. While adding more details on the last segment of the kinematic chain, produced no improvements or

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worsening in sitting and rising from a chair and stepping. But for motor tasks where some motion between parts of the trunk could occur such as squatting a separation in more rigid segments is necessary.

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