Joint forces and moments calculation for a 3D whole body model during complex movement

Application to the balance recovery movement following a support surface translation

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***Abstract*—The purpose of this study was to calculate the joint forces and moments for a complete human body model during 3D complex movements of balance recovery. A 40-dof body model was used. The movements were measured using an optoelectronic system and reconstructed using a semi-global optimization method. The joints forces and moments were calculated through a classical Newton-Euler recursive algorithm. The consistency of the computed loads was checked by comparing the results of different calculation strategies. The first attempts with classical hypotheses showed poor results. A sensitivity analysis was lead to determine the most critical parameters. Thus, investigations were performed to improve the estimation of the position of the centers of mass and the derivation of the kinematic parameters. The use of 3D regression equations for the estimation of the Body Segment Inertial Parameters and the use of the residual analysis to adjust the filters cut-off frequencies allowed major improvements of the quality of the solutions. The results obtained that way were found to be meaningful.**

Keywords - Inverse dynamics; 3D movements analysis; balance recovery

I. INTRODUCTION

The calculation of joint forces and moments during a movement is a classical process in the human movement analysis. Nevertheless, in many studies, the forces and moments are computed for simplified models of the human body (e.g. few segment models [1], 2D models [2]…) or for smooth and non-complex movements (e.g. gait analysis). In addition, validation of these calculated loads is rarely presented.

The aim of this study was to compute the joint forces and moments for a 3D whole body model during balance recovery movements. These results will then be used for balance recovery analysis. Therefore, the accuracy of the computed loads needed to be sufficient in order to well represent their temporal evolution and in order to differentiate the movement strategies. Nevertheless, the convergence of a direct dynamics simulation using these results was not required.

This paper presents the methods used, the difficulties encountered and the validated results.

II. MATERIALS AND METHODS

A. *Test set-up*

Ten young male volunteers, close to the 50th percentile morphology, participated in this study. They were standing on a platform suddenly translated toward their back. This represents the situation of a standing passenger in a railway public transport vehicle submitted to a real-life incident, as an emergency braking or a light collision. The perturbations were half sinusoidal impulses with a maximum acceleration between 2 and 10 m/s² and a duration of 300 ms. These perturbations induced strong disequilibrium: for the low perturbations, subjects needed at least one recovery step to restore their balance, while nearly 4 meters were needed to stop the fall for the high perturbations.

A Motion Analysis® system equipped with five Eagle® cameras recorded the 3D position of 50 reflective markers stuck on the subject with a sampling frequency of 200 Hz. Loads between the feet and the ground were collected using two Bertec® force plates sampling at 1 kHz.

B. *Kinematics*

The kinematics of the balance recovery movement was computed from the collected trajectory of the skin markers.

The complexity of the observed movements raised measurement difficulties. Thus, in order to obtain an over-constrained system and to improve the quality and the stability of the results, a kinematic model of the human body was introduced. Therefore, the virtual dummy Man3D [3] was used to represent the subjects' body. Its kinematic model was simplified to a 15 linked segments model corresponding to 40 degrees of freedom (see Fig. 2a). A morphing method was used to adapt this kinematic model to each volunteer's anthropometry.

A semi global optimization method was applied at each frame of the movement so that the dummy's posture best fits the subject measured position. The principle is to find the set of joint angles that minimizes the distance between markers fixed on the virtual dummy and their measured position. The global

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system is considered as a set of small kinematic chains hierarchically organized. The algorithm solves the problem from the most distal chain to the most proximal one. A global solution is found iteratively [4]. Fig. 1 shows an example of a reconstructed movement.

Figure 1. comparison between a video recording and a reconstructed movement

C. Inverse dynamics method

Different methods to calculate the joint forces and moments have been considered. An interesting approach is the use of a global optimization method combining inverse and forward dynamics ([5], [6]). The main interest of these methods is that they tend to obtain consistent solutions: the forces and moments calculated, applied then in direct dynamics, induce a movement that best fits the oneh reconstructed from measurements. Nevertheless, the complexity of the kinematic model used in this study was a major drawback for this approach. Indeed, the need to write the complete set of dynamic equations driving the movement for the whole system would have been time consuming. In addition, these methods, applied to the inherently unstable models of standing humans, leads to calculation difficulties and divergence problems ([5]).

As the knowledge of forces and moments patterns during the balance recovery movement was sufficient in this study, it was not necessary to ensure that the computed solutions were consistent in direct dynamics. Thus a more classical approach was used: the Newton-Euler recursive algorithm. It consists in isolating each body segment and solving their dynamic balance equation. This can be written as follows (1):

$$[\Phi_{i-1/i}]_{R0} = [A_{i/0}]_{R0} - [\Phi_{i+1/i}]_{R0} - [\Phi_{P/i}]_{R0} - [\Phi_{ext/i}]_{R0.} \qquad (1)$$

$[A_{i/0}]$ represents the generalized forces due to the segment dynamics and $[\Phi_{i-1/i}]$, $[\Phi_{i+1/i}]$, $[\Phi_{P/i}]$, $[\Phi_{ext/i}]$ the forces and moments applied on the segment i by the previous segment, the next segment, the gravity and the other external forces (i.e. contact forces) respectively. By the way, given the movement of the segments, their inertial properties and the external forces applied on the system, the inter-segmental forces and moments of a linked segment system can be recursively calculated from the most distal segment to the root. In this study, external forces and moments were directly recorded and the kinematics was determined with the method previously described. The homogeneous matrix formalism [7] has been used for the calculation because of its simplicity of use and because of the stability of its results against the input perturbations [8]. The derivation of the kinematic parameters and the estimation of the Body Segments Inertia Parameters (BSIP) will be discussed below.

D. Calculation strategies and validation

As this method solves the problem locally for each body segment, it does not ensure the consistency of the global solution. Nevertheless, this is not a drastic problem, as a simple method exists to estimate the global consistency of the results. It consists in using the extra degrees of freedom introduced by the measurement of the loads between the subject and its environment [9]. This is sometimes mentioned in the literature by the presence of residual virtual loads acting on the top most body segment ([5] or [10]).

The subjects' body was considered as a tree structure made of eleven segments – the inertia of the hands and the feet has been classically neglected and thus fused with the forearms and legs. As recursive calculation can only be applied to convergent chains, two strategies had to be used to calculate the loads from the most distal segment to the root. An estimation of the consistency of the results was then directly given by the comparison of the loads between the pelvis and the thorax calculated with the two strategies (see Fig. 2b).

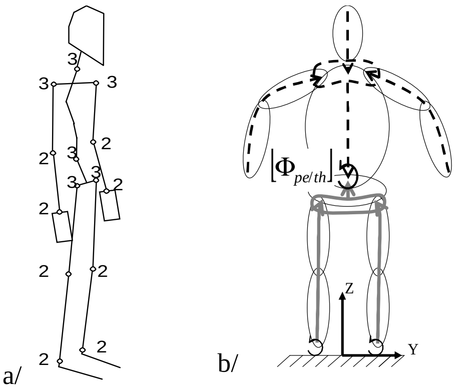

Figure 2. a/ kinematic model (number of dof indicated near the joints) b/ dynamic model with the two calculation strategies: going up (thin grey lines) and going down (dashed black lines).

III. JOINT FORCES AND MOMENTS CALCULATIONS

A. 1st step: classical hypotheses are unsatisfying

At first, classical hypotheses were assumed: the inertia parameters of the 11 body segments were estimated using De Leva's regressions [11] and a classical gradient method was used for the double derivation of the kinematic parameters, with filtering (Butterworth 2nd order) at each step. For each movement, the cut-off frequency was estimated from the data inspection and applied to every degree of freedom.

Comparison of the forces and moments between the pelvis and the thorax showed poor quality results, especially concerning the joint moments.

A sensitivity analysis was performed to find the probable causes of the bad quality of the results. As Silva [12] did, small perturbations were introduced in the input data set. The sensitivity K_{RI} of the result R to the input I was calculated for the 3 forces and 3 moments of each joint at each frame of the movement by $K_{RI} = (R_p-R_{np}) / (I_p-I_{np}) = \Delta R/\Delta I$.

These sensitivity coefficients were found to be independent of the magnitude of the perturbation. Thus, by fixing the maximum allowed difference for the outputs (ΔR_{max}) the corresponding maximum input perturbations (ΔI_{max}) were estimated: $\Delta I_{max}=f(K_{RI}, \Delta R_{max})$. From the comparison between these allowed input perturbations with the estimated precision of our data, the critical inputs were then highlighted.

Two critical sources of error were identified: the accelerations of the heavy segments and the position of the centre of mass of these segments. Transversal and longitudinal ground reaction forces were influent enough to induce large error if additional inertial loads due to the movement of the force-plates are not taken into account [13]. Masses and inertias of the segments, as angles for the distal joints, have shown little influence.

B. 2nd step: improvements

From these results, the improvements to be carried out concerned the two critical inputs: the position of the centre of mass and the derivative of the kinematic parameters.

Despite their simplicity of use, the drawback of the regressions proposed by De Leva is that the BSIP are related to the longitudinal axis of the segments. This axis is clearly defined for long segments (eg: the arms) but is not for segments such as the pelvis or the thorax.

Dumas [14] recently proposes regressions based on data from Young *et al.* [15] and McConville *et al.* [16]. These regressions were adjusted in order to provide 3D BSIP directly applicable in the conventional segment coordinate systems. In order to compare the 3D regressions with those of De Leva, the centre of mass (CoM) of subjects' whole body was estimated from a reconstructed posture with the two sets of regressions. The projection of this point onto the floor was compared with the location of the centre of pressure (CoP) measured with the force-plates. In a resting standing posture, these two points should be merged. Table 1 below clearly indicates that for this study, the 3D regressions gave better results.

TABLE I. AVERAGE |COM-COP| (MM)

	longitudinal	**transversal**
De Leva	35	7
Dumas	9	1

Alternative methods for the derivation of kinematic parameters have been considered, such as interpolation with 5th order cubic splines and formal derivation [17]. Nevertheless, the problem of compromise between smooth and accurate results remained the same. Therefore, the classical gradient method with filtering was conserved but the determination of the cut-off frequencies has been improved.

The choice of the best method to automatically filter biomechanical data is not completely settled [18]. Some time-frequency methods adapting the filter to the signal frequency content all along the movement are more efficient in case of high non-stationarities (e.g. impacts due to landing [19] or ball kicking [20]). But, as in this study the frequency content of the kinematic signals (joint angles) did not vary drastically, a simpler method was used to determine constant cut-off frequencies all along the movement: the residual analysis proposed by Winter [10] was applied for each degree of freedom in order to automatically determine the cut-off frequency that best remove the noise without deforming the signal.

The problem of the loss of consistency for the velocity and acceleration matrices, due to term to term derivation of the position matrices, has been also investigated. It has been solved by reintroducing the Denavit-Hartenberg parameterization used for the virtual dummy. Nevertheless, this had no real impact on the results' improvement.

The results obtained by applying the residual analysis method and the 3D regressions are more consistent. The shape of the curves is respected but the results are still really noisy. Fig. 3a and 3b below show an example of the comparison of the joint forces and moments between the pelvis and the thorax obtained from the 2 calculation strategies (going up and going down).

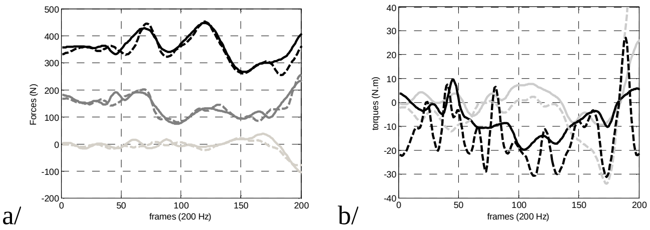

Figure 3. comparison of forces (a) and moments (b) between the pelvis and the thorax calculated with two different strategies (doted or thick lines)

C. 3rd step: validation

The origin of these oscillations has been investigated. The lack of filtering seemed to be the most logical cause. Nevertheless, an analysis of the signals frequency components showed that the movements contained frequency components of the same order of magnitude as the noise. Thus, filtering with too low cut-off frequencies could have removed not only the noise but also intrinsic components of the movement and led to incorrect calculated joint forces and moments. One solution to this problem is to perform inverse dynamics calculation with cut-off frequencies given by the residual analysis method in order to obtain consistent but noisy results. Then the low frequency components can be filtered *a posteriori* with cut-off frequencies determined from the power spectrum density of the signal. The figure 4 below illustrates this for a highly perturbed movement.

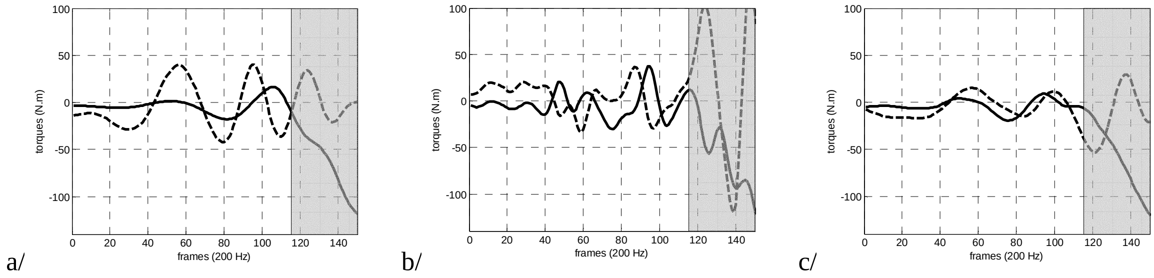

Figure 4. Hip transversal moment for a highly perturbed movement for two calculation strategies. Grey zones must not be considered as subjects step then outside of the forceplates. a/ derivation with fixed Fc (6Hz) b/ derivation with adjuted Fc (residual analysis) c/ derivation with adjuted Fc and moment refiltered (6Hz)

Results obtained so far are consistent and meaningful for the analysis of balance recovery movements. For example, the movements of the subjects for a given experimental situation have been clustered into strategies. These kinematic based clusters can also be observed for dynamic data.

IV. CONCLUSION

The aim of this study was to calculate the joint forces and moments for a whole human body model during 3D complex movements of balance recovery. It has been shown that classical hypotheses could lead to poor quality of the calculated forces and moments. This was highlighted with a simple verification method. A sensitivity analysis has shown that the most critical parameters were the accelerations and the estimation of the position of the center of mass of the heavy segments. The use of a 3D set of regressions and of the residual analysis to automatically adjust the filters with the cut-off frequencies improved the quality of the results: the joint forces and moments calculated were consistent but noisy. A frequency analysis has shown that higher filtering during the double derivation phase would remove movements' components. Consistent and meaningful results were obtained by filtering *a posteriori* the calculated forces and moments.

The two underlying conclusions of this paper are: (1) the importance of evaluating as soon as possible the consistency of the computed forces and moments; (2) the possibility to calculate meaningful joint forces and moments for a 3D whole body model with simple methods, even in case of complex movements.

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