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Franck Barbier, Paul Allard, Kevin Guelton, Briac Colobert, Anne-Pascale Godillon-Maquinghen.
Estimation of the 3D center of mass excursion from force plate data during standing. IEEE
Transactions on Neural Systems and Rehabilitation Engineering, 2003, 11 (1), pp.31-37.
10.1109/TNSRE.2003.810433 . hal-00793230

HAL Id: hal-00793230

<https://hal.science/hal-00793230v1>

Submitted on 21 Feb 2013

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Estimation of the 3D center of mass excursion from force plate data during standing

F. Barbier, P. Allard, K. Guelton, B. Colobert, A.P. Godillon-Maquinghen

Abstract-- Biomechanical models are used with force plate information to determine the center of mass (COM) trajectory during standing. They are usually based on simplifying assumptions and are often limited to a single plane analysis. The objectives of this study were to present a three-dimensional model to calculate the excursion of the COM of the human body, validate it and compare its performance to a video-based system during quiet standing and antero-posterior (AP) and medio-lateral (ML) self-imposed oscillations. In addition to the vertical displacements of the COM, the originality of the method lies in eliminating the accelerations terms in the model and their related assumptions. The model was able to estimate closely the COM displacements in quiet standing (RMS of 0.9 mm or less). For the self-imposed oscillations, the RMS differences were 6.6 mm in the AP and ML directions and 1.6 mm along the vertical axis. For all three testing conditions, the coefficients of correlation of the COM displacements between the model and the video methods were above 0.8 with the exception of the vertical direction where the values were more variable.

I. INTRODUCTION

The ability to maintain standing posture is usually assessed by the amount of body sway [1]. Usually, the subject is evaluated while standing quietly on a force plate while the displacement of the center of pressure (COP) is monitored using ground reaction forces and moments. The COP is indicative of body sway and reflects the neuromuscular response to correct the displacement of the center of mass [2]. Novel analytic techniques were used to model the displacement of the COP [3][4][5] to further in its interpretation in relation to pathological conditions. But, its

Manuscript received September 4, 2001. This work was supported in part by the the Région Nord-Pas de Calais (France), Direction Régionale de la Recherche et de la Technologie, the Association of Colleges and Universities of Canada, the Natural Sciences and Engineering Research Council of Canada, The Hospital for Sick Children Foundation and by a French NATO Senior Guest Scientist scholarship awarded to Paul Allard.

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displacement remains different from that of the body center of mass which can be considered as the controlled variable [6].

The trajectory of the COM cannot be measured directly but can be calculated using video-based systems which suffer from a low signal-to-noise ratio [7][8] combined with anthropometric information considered as intrinsically inaccurate [9]. Video-based systems are also a time consuming method to be applied routinely in a clinical setting and requires expensive equipment compared to the purchase of a force-plate.

Biomechanical models were developed to estimate the COM displacements during quiet standing [6][10][11][12][13][14][15] and walking [16]. During quiet standing, the human body is often modeled as an inverse pendulum representing a point mass oscillating about the ankles [6][10][17]. Three approaches have been used to determine the COM excursion from force plate data alone. Brenière [17] and Levine and Mizrahi [10] applied a low pass filter on the COP displacements. Among other limitations, this method does not account for the phase difference between the COP and the COM [18]. Another method lies in the calculation of the second integral of the acceleration since the horizontal forces are proportional to the acceleration of the COM. The difficulty with this method is to estimate the initial integration constants. Crowe [16], Levine and Mizrahi [10], Zatsiorski [18] and Shimba [12] developed curve fitting techniques, optimization methods or made some assumptions to solve with success the double integration problem. A third method consists of using inverse dynamic methods; but most of these are hampered by the linear or angular terms in the equations of motion. Levin and Mizrahi [10] assumed that the derivative of the angular momentum was negligible whereas Karlsson and Lanshammar [13] compared the accelerations given by the force plate with those from an inverse dynamic model to study postural movement strategies in the sagittal plane. All these models uniquely provide the horizontal displacements of the COM and were often the results of a planar analysis. The vertical excursions of the COM were usually considered negligible. Several simplifying assumptions were made which can be justified in part for quiet standing but not for self-imposed body oscillations.

The objectives of this work were a) to present a method to calculate the three-dimensional (3D) excursion of the center of mass of the human body from force plate data during standing, b) validate the model by comparing the accelerations of the COM with force plate and video data and c) compare the displacements of the COM calculated from the model with those obtained from a video-based system during quiet

standing and for self-imposed antero-posterior and medio-lateral oscillations.

II. THREE-DIMENSIONAL INVERSE PENDULUM MODEL OF THE CENTER OF MASS

The standing upright human body is represented as an inverted pendulum, shown in Fig. 1. It oscillates about an origin C located at a point lying midway between the ankles with positive axes along the anterior (x), left (y) and upwards (z) directions. The origin of the inverse pendulum was arbitrarily located between ankles since the displacement of the COM is not affected by the position of the inverse pendulum's origin. Appendix A illustrates that by displacing the origin to the left ankle, the COM excursion remains the same relative to the new origin due to the transfer of moments. Consequently, setting the origin between the ankles does not exclude studies involving individuals who usually do not stand with 50% of their weight on each foot such as in lower limb amputees and cerebral palsy, strokes and Parkinson patients.

The center of mass of the body less the feet, G_{b-f} is modeled as a particle in space and of body weight \vec{W}_{b-f} having an acceleration \vec{a}_{b-f} . The distance between the origin and the center of mass of the body less the feet is assumed to be constant and of length r . The feet are modeled as a single segment of weight, \vec{W}_f and where its center of mass is located at the midpoint, G_f of a line joining the center of mass of the

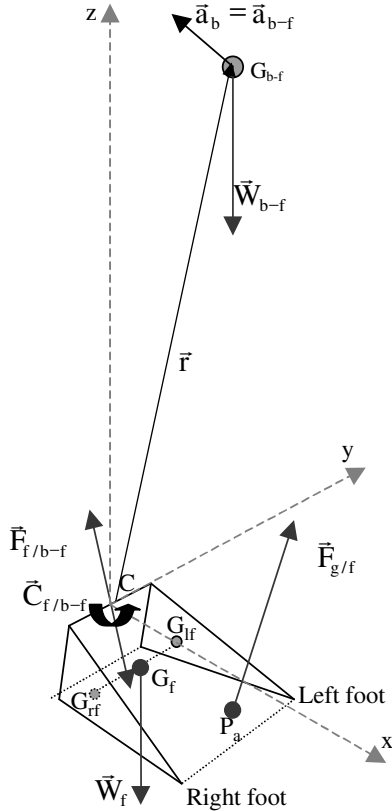


Fig. 1. Inverse pendulum representation of the center of mass of the human body oscillating about a point lying midway between the ankles. The symbols are described in the text.

right (G_{rf}) and left (G_{lf}) feet. The feet are assumed to be rigid and fixed to the force plate. The ground reaction force is acting on the feet, $\vec{F}_{g/f}$ at point P_a which lies on the force plate. It is assumed that the joint reaction force, $\vec{F}_{f/b-f}$ and moment $\vec{C}_{f/b-f}$ of the feet on the body less the feet act at the origin, C .

To solve for the COM position, the equations of motion were first applied to the feet and then to the body less the feet segment. For the feet, the second equilibrium equation about the origin, C is:

$$\vec{C}_{b-f/f} + \vec{M}_{\vec{F}_{g/f}} + \vec{M}_{\vec{W}_f} = \vec{0} \quad (1)$$

and where $\vec{M}_{\vec{F}_{g/f}}$ and $\vec{M}_{\vec{W}_f}$ are respectively the moments due to the ground reaction force and due to the weight of the feet about C .

For the body, which is modeled as a particle, the second equation of motion about the origin C is:

$$\vec{C}_{f/b-f} + \vec{M}_{\vec{W}_{b-f}} = \dot{H}_0 \quad (2)$$

and where $\vec{C}_{f/b-f} = -\vec{C}_{b-f/f}$, $\vec{M}_{\vec{W}_{b-f}}$ and \dot{H}_0 are respectively the moment produced by the weight of the body less the feet about C and the derivative of the angular momentum about C . The derivative of the angular momentum is developed into $\dot{H}_0 = \vec{CG}_{b-f} \times m_{b-f} \cdot \vec{a}_{b-f}$ with \vec{CG}_{b-f} as the moment arm of the COM of the body less the feet about the origin C , m_{b-f} as the mass of the body less the feet and \vec{a}_{b-f} its acceleration. These terms are inserted in equation 2, which becomes:

$$\vec{C}_{f/b-f} + \vec{M}_{\vec{W}_{b-f}} = \vec{CG}_{b-f} \times m_{b-f} \cdot \vec{a}_{b-f} \quad (3)$$

This substitution is the first major contribution of this work in the development of an inverse pendulum model to determine the displacement of the center of mass of the human body during standing. By replacing \dot{H}_0 in the equation 2, assumptions related to the derivative of the angular momentum are no longer required.

The displacement of the COM given by \vec{CG}_{b-f} is still expressed as a function of the linear acceleration in equation 3. Simplifying assumptions such as the double integration methods [18], iterative processes [6][10] or curve fitting techniques [12] can be used to solve equation 3. To avoid the use of simplifying assumptions, we proposed to eliminate the linear acceleration term in equation 3. This is first achieved by setting $\vec{a}_{b-f} = \vec{a}_b$ since the feet are fixed and where

$\vec{a}_b = \frac{\vec{W}_b + \vec{F}_{g/f}}{m_b}$ is the acceleration of the COM of the whole

body with \vec{W}_b the weight of the body. Then using the first equation of motion, the accelerated mass was replaced by the

sum of external forces, namely the weight of the body \vec{W}_b and the joint reaction force $\vec{F}_{g/f}$. Equation 3 becomes:

$$\vec{C}_{f/b-f} - \vec{CG}_{b-f} \times (1 - K) \cdot \vec{F}_{g/f} = \vec{0} \quad (4)$$

with $K = \frac{m_f}{m_b}$. The substitution of \vec{a}_b is the second major

contribution of this work. By eliminating the linear acceleration related assumptions are no longer required.

Expanding 4 yields the x, y and z the co-ordinates of \vec{CG}_{b-f}

$$\begin{cases} C'_x - F_z \cdot y + F_y \cdot z = 0 \\ C'_y + F_z \cdot x - F_x \cdot z = 0 \\ C'_z - F_y \cdot x + F_x \cdot y = 0 \end{cases} \quad (5)$$

with F_x, F_y, F_z the ground reaction force and

$$\begin{bmatrix} C'_x \\ C'_y \\ C'_z \end{bmatrix} = \frac{1}{1-K} \cdot \begin{bmatrix} y_a \cdot F_z - z_a \cdot F_y + y_f \cdot K \cdot W_b \\ z_a \cdot F_x - x_a \cdot F_z - x_f \cdot K \cdot W_b \\ x_a \cdot F_y - y_a \cdot F_x \end{bmatrix}$$

comes from the expansion of the equation 1 with x_a, y_a, z_a and x_f, y_f, z_f the co-ordinates of the point of application of the ground reaction force, P_a and of the center of mass of the feet, G_f respectively.

Equation 5 can be solved for an infinite number of z values. For a particular subject, the distance between the center of mass of the body less the feet and the origin C needs to be specified. This is the length r (Fig. 1) which is assumed to be constant and can be obtained by calculating the position of the center of mass of the body by the reaction board method and subtracting from it the height of the ankles. The constraint is the radius of a sphere r given by

$$x^2 + y^2 + z^2 - r^2 = 0 \quad (6)$$

Substituting equation 6 into 5, three-second order equations were obtained and solutions are

$$\begin{aligned} x_{1,2} &= \frac{C'_z \cdot F_y - C'_y \cdot F_z \pm \sqrt{\Delta_x}}{\left| \vec{F}_{g/f} \right|^2} \\ y_{1,2} &= \frac{C'_x \cdot F_z - C'_z \cdot F_x \pm \sqrt{\Delta_y}}{\left| \vec{F}_{g/f} \right|^2} \\ z_{1,2} &= \frac{C'_y \cdot F_x - C'_x \cdot F_y \pm \sqrt{\Delta_z}}{\left| \vec{F}_{g/f} \right|^2} \end{aligned} \quad (7)$$

with

$$\Delta_x = (C'_y \cdot F_z - C'_z \cdot F_y)^2 - \left| \vec{F}_{g/f} \right|^2 \cdot (C_y'^2 + C_z'^2 - F_x^2 \cdot r^2)$$

$$\Delta_y = (C'_z \cdot F_x - C'_x \cdot F_z)^2 - \left| \vec{F}_{g/f} \right|^2 \cdot (C_x'^2 + C_z'^2 - F_y^2 \cdot r^2)$$

$$\Delta_z = (C'_x \cdot F_y - C'_y \cdot F_x)^2 - \left| \vec{F}_{g/f} \right|^2 \cdot (C_x'^2 + C_y'^2 - F_z^2 \cdot r^2)$$

There are two solutions to equation 7 which are a mirror image of each other. The solution for a positive z value was kept. Afterwards, the center of mass of the feet G_f and that of the body less the feet G_{b-f} are transposed into the laboratory coordinate system. The COM position \vec{OG}_b is calculated by

$$\vec{OG}_b = \frac{m_f \cdot \vec{OG}_f + m_{b-f} \cdot \vec{OG}_{b-f}}{m_b} \quad (8)$$

Equation 8 enables the estimation of the displacement of the center of mass of the human body during standing using force plate information only. This is the third and most important contribution of this work. The following sections describe the comparison between the proposed inverse dynamic model and a video-based method.

III. METHOD

Eight male able-bodied subjects participated in the validation study. Their average age and weight were 24.0 ± 4.12 years and 74.86 ± 9.35 kg respectively. Their height was 1.815 ± 0.076 m. Subjects had no previous orthopedic ailment or neurological disorders which could affect their standing posture.

Quiet standing balance was evaluated by means of a Saga 3 video-based system¹ and a Logabex force plate² synchronized together. Before the experimentation, six cameras were located around the force plate at a distance of 3 m from its center. These were calibrated using 16 markers distributed in a calibration volume 2 m high, 1 m deep and 1.2 m in width. The three-dimensional reconstruction error was 0.6 mm [19]. To determine the zero off-set, data were collected from an unloaded platform. The error in estimating the center of pressure by the force plate was 1.1 mm.

Each subject was fitted with 2,5cm reflective markers to define eleven body segments. Markers were put over the second metatarsophalangeal joints, lateral maleoli, knees and hips to delimit the segments of each lower limb and over the acromion, elbow and wrist of each upper limbs as well to identify the upper extremities. The head and trunk were grouped as a single segment using the acromion and hip markers as references.

For each subject, the position of the feet was standardized by means of a wooden jig lying flat on the force plate according to McIlroy and Maki [20]. With the subject standing barefooted, the heels were aligned and spaced by about 23 cm and the midline of the feet pointing externally by 15°. Once the subject was in position, the rig was removed. Then, he was asked to stand quietly, with the arms at the sides and focus on a target placed at eye level, 1,2 m in front.

Video cameras were synchronized with the force-plate system. During an acquisition, video and force-plate data were collected simultaneously at 50 Hz for a period of 20 s. In quiet

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standing, 90% of the power spectral density of the COM excursion is contained between zero and 5Hz [21]. Winter et al. [22] calculated acceleration values from a video-based system working at a frequency of 20Hz. A 50Hz sampling frequency satisfies both the video and force-plate requirements and is often found in the literature [15][23]. Afterwards, video and force plate data were filtered with a fourth order zero-phase lag Butterworth filter having a cut-off frequency of 5Hz to reduce the noise [24].

To test the model, each subject was evaluated for three experimental conditions. The first was quiet static standing. This was followed by a self-imposed antero-posterior (AP) oscillation and a self-imposed medio-lateral (ML) oscillation. There was a single trial for each condition. For the self-imposed antero-posterior and medio-lateral conditions, subjects were asked to remain rigid and oscillate back and forth as much as possible in a given direction. Particular care was given to ensure that the trunk and lower limbs oscillated in unison to avoid double pendulum motions where the hips and shoulders move in opposite directions. The maximal body oscillations could vary from one subject to another; but, this does not affect the data analysis since the displacements of the COM calculated from the video information were compared to the corresponding results obtained from the inverse dynamic model for each subject, each trial and each condition.

To validate the results, the accelerations the body center of mass obtained by the second derivative of the linear displacements of video and model data using the central difference technique [24] were compared in each direction with those obtained from force-plate data after removing the mass and the action of the gravity. For each testing condition and for each axis, RMS differences were calculated between the accelerations given by the force plate and those obtained by the video and model methods.

The force plate measured body weight while the position of the center of mass of the body with respect to the feet was determined by the reaction board method. Anthropometric tables were also used to estimate the weight of each body segments [24]. The displacements of the center of mass were calculated from video data combined with anthropometric information and using equation 8.

The performance of the model was given by the RMS differences between the antero-posterior, medio-lateral and vertical trajectories of the COM calculated from the video data which acted as reference values and those of the proposed model. The Pearson coefficients of correlation were also calculated to determine the goodness of fit between the displacement of the COM given by the video system and those estimated by the model. The RMS displacement differences and the coefficients of correlation were calculated for each test conditions and for each axis. Student t tests for dependant samples were performed to determined statistical differences ($p < 0.05$).

IV. RESULTS

The first objective was to develop a method to calculate the three-dimensional excursion of the center of mass of the human body from force plate data during standing. The

trajectories of the COM obtained from video information and calculated by the model are given in Fig. 2 along the antero-posterior, medio-lateral and vertical axes for a subject during quiet standing. Though, the curves generally followed a similar pattern as those illustrated in Fig. 2, the difference between the vertical displacements (Fig. 2c) estimated by the model and those obtained from the video-based system are within the reported range. For this particular subject, the horizontal displacements were more closely matched.

Fig. 3 show the RMS differences between the COM accelerations given by the force plate and those obtained from the video and the model respectively for the three test conditions and for each direction. The average RMS values were relatively low and averaged 59.8 mm/s^2 ranging from 1.7 to 142 mm/s^2 . The lowest average RMS values were observed in the static condition where the mean RMS error was $19.9 \pm 6.6 \text{ mm/s}^2$. For a given direction, the lowest mean RMS value was $32.4 \pm 18.9 \text{ mm/s}^2$ in the vertical axis while the highest RMS error was in the medio-lateral direction with a value of $85.5 \pm 39.0 \text{ mm/s}^2$.

In the static and the self-imposed antero-posterior oscillations, there was no statistical difference between the video and the model RMS acceleration. The only statistically significant differences were found for the self-imposed medio-lateral oscillations. The RMS acceleration of the model were less than two times higher than the video in both the AP and ML directions though the average values were less than 142 mm/s^2 .

The mean RMS differences between the COM displacements are given in Fig. 4 for the three conditions tested and for each axis. The RMS COM displacement were relatively low and averaged 3.5 mm and ranged between 0.43 to 6.93 mm . The lowest mean COM displacement RMS values were observed in the static condition with a mean of $0.68 \pm 0.22 \text{ mm}$. The lowest COM displacement RMS values observed in the vertical direction with a mean of $1.21 \pm 0.70 \text{ mm}$ while the highest were found in the ML and AP direction $4.7 \pm 3.04 \text{ mm}$. Though the RMS differences for the self-imposed AP and ML oscillations were 4 times higher, no significant difference was found in the COM excursion determined by the either the video or the model method.

The Pearson coefficients of correlation between the COM displacements given by the video system and those estimated by the model for the three testing conditions and for each axis are given Table I. These are well over 0.8 except for the vertical direction.

V. DISCUSSION

The second objective was to validate the model. Since the ground reaction forces reflect the accelerations of the center of mass of the body, force-plate data can be use as reference values to compare the performance of both the 3D model and the video method. Based on the RMS acceleration differences, both the model and the video methods can be used adequately for assessing standing equilibrium. These RMS differences were considered relatively low especially for quiet standing condition. In the self-imposed conditions, the RMS acceleration differences were higher but still low.

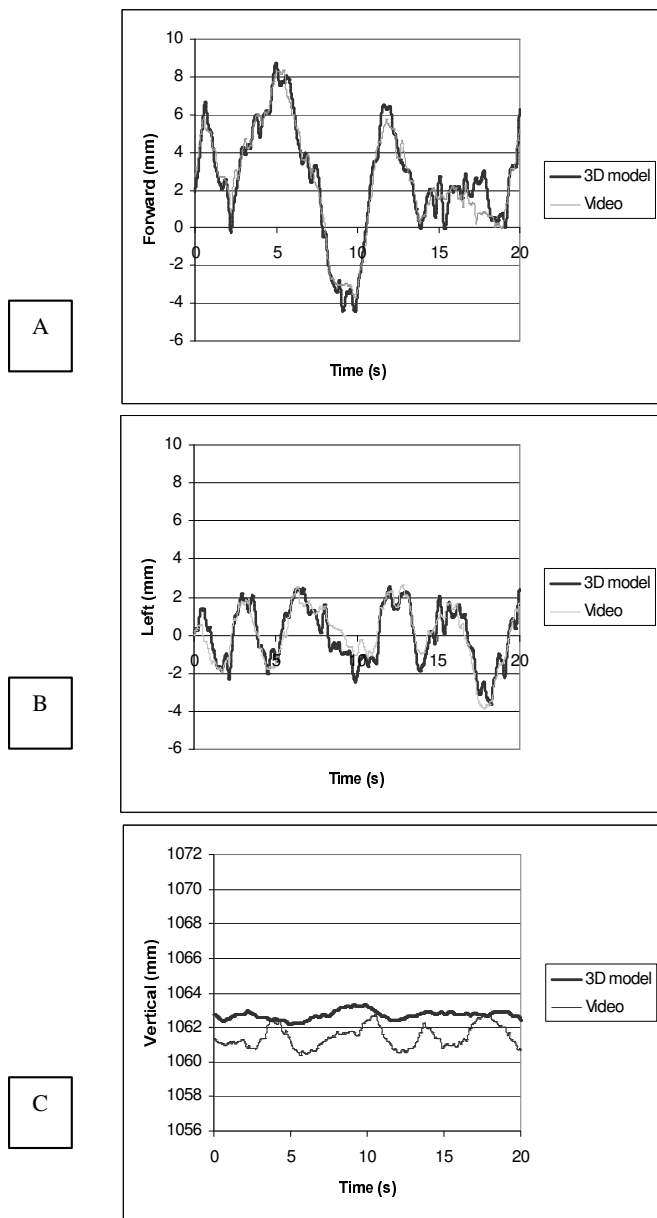


Fig. 2. Excursion of the COM measured by the video and calculated by the model for a sample subject during quiet standing for the a) antero-postero, b) medio-lateral and c) vertical axes.

TABLE I

	STATIC	Self-imposed AP	Self-imposed ML
Antero-posterior	0.98 (0.03)	1.00 (0.00)	0.84 (0.06)
Medio-lateral	0.93 (0.05)	1.00 (0.00)	1.00 (0.00)
Vertical	0.41 (0.33)	0.81 (0.23)	0.43 (0.23)

Pearson coefficient of correlation of the COM displacements calculated between video and the model data for the three test conditions (static, self-imposed antero-posterior and medio-lateral directions) along each axis.

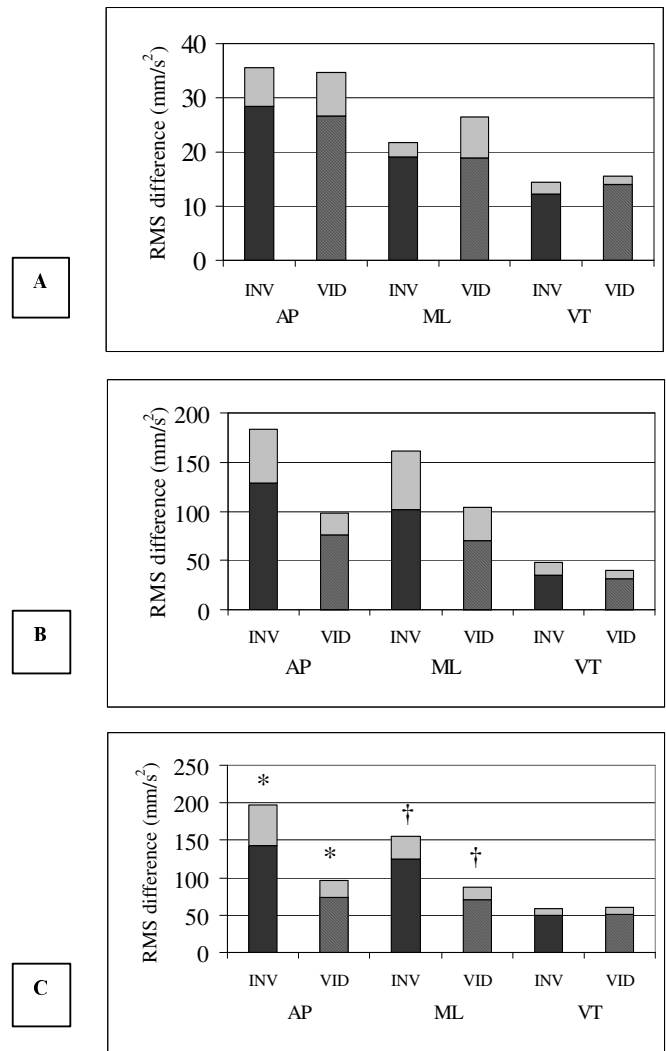


Fig. 3. RMS acceleration differences between the force plate data and those calculated by means of the video system (VID) or estimated by the model (INV) during the a) static, b) self-imposed antero-posterior and c) self-imposed medio-lateral oscillations for the antero-posterior (AP), the medio-lateral (ML) and vertical (VT) axes.

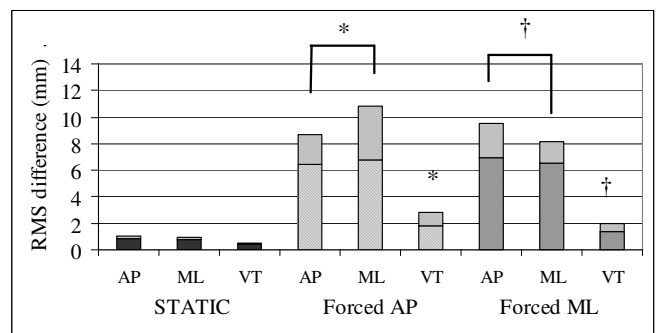


Fig. 4. RMS displacement differences between the excursion of the COM calculated by means of the video system (VID) and those estimated by the model (INV) during the static, self-imposed antero-posterior and self-imposed medio-lateral oscillations for the antero-posterior (AP), the medio-lateral (ML) and vertical (VT) axes.

Few reported validity results on their model when reporting COM displacements. Winter et al. [22] reported correlation coefficients between the difference between the center of pressure measured by the force plate and the COM by a video-based system that were generally above 0.9 for the AP direction with slightly lower values in the ML direction.

The third objective was to compare the antero-posterior, medio-lateral and vertical displacements of the COM measured by a video-based system with those obtained by the model. RMS differences in the COM displacements were reported in various studies for quiet standing only. Zatsiorsky and King [18] found RMS differences to be slightly more than 9 mm between video and their method to estimate the COM excursion while Levine and Mizrahi [10] reported RMS differences ranging from 1.01 to 4.66 mm. Morasso and al [6] validated their model by means of an inverted pendulum stabilized by springs. They defined a percentual amplitude error from the estimated COM positions given by their model and those given by a video method. It was 8.95% of the input range. Our average COM displacement RMS values (0.68 mm) were much lower than those reported by Levine and Mizrahi [10] for quiet standing and within their range for self-imposed oscillations (4.7 mm). To our knowledge, no one validated their models for self-imposed oscillations.

Rabuffetti and Baroni [14] reported the coefficients of correlation between the COM displacements determined by a double integration technique and a video-based system for the ground contact phase of a jump. The mean values ranged between 0.956 and 1.000. The lowest coefficient was 0.162 calculated in the transverse axis where the smallest displacements of the body center of mass were expected. For quiet standing, the horizontal plane correlation coefficients found by Zatsiorsky and King [18] ranged from 0.79 to 0.94. We have found similar coefficients of correlation ranging between 0.84 and 1.00. A low coefficient of correlation was found in the vertical direction (0.41) where the displacements were smaller. This could be attributed to a small COM vertical excursion of 1.1 ± 0.4 mm in the static condition. The high correlation (0.81) for the self-imposed AP oscillation can be explained by greater vertical amplitude of 5.3 ± 1.3 mm.

A three-dimensional inverse dynamic approach was used to estimate the center of mass of the human body from force plate data during standing without any major assumption. The height of the COM was not assumed to be constant [6][10] though its position with respect to the ankle was of constant length. Since Newton's first equation was used to eliminate the linear acceleration term in the model, there was no need to determine the constants resulting from the double integration of the accelerations [12][14], make assumption on the forces and moments [18] or develop iterative processes [6][10]. The time history excursion of the COM was not assumed dependant on the center of pressure as in the low pass filter method [15], [11][17] since there is a non-linear relationship between the COP and COM. Because the angular acceleration term was replaced by a moment, the rate of change of the angular momentum was not considered either constant nor negligible [10].

All the above assumptions seem justified when evaluating quiet standing. They may be violated when standing is perturbed by mechanical devices [25] or by bending or moving a body segment [26]. Previously, the vertical displacement of the COM could only be estimated using a video-based system. With the present model, the COM vertical excursion is computed as well as the horizontal displacements.

Though the proposed model does not rely on any major simplifying assumptions, a limitation resides in the estimation of the distance between the center of mass and the midpoint between the ankle. This distance was assumed to be constant. Furthermore, this distance was measured while the subject was lying supine with the effect of the gravity removed. The standing height of the center of mass could be slightly lower. Notwithstanding this limitation, the proposed model behaved well in a static condition and reasonably well in self-imposed oscillations. It was able to provide the horizontal and vertical displacements of the COM which only video based systems were able to provide to date.

VI. CONCLUSION

A three-dimensional inverse dynamic approach was used to estimate the excursion of the center of mass of the human body using a single force plate during standing. No major simplifying assumption was introduced in the model. Since the height of the COM is not fixed, the proposed method provides the vertical trajectory of the COM. The RMS acceleration differences were considered relatively low especially for the quiet standing condition. Both the model and the video method can be used adequately for assessing standing equilibrium. In quiet standing, the model was able to provide a good agreement with the COM displacements obtained from the video system. For the self-imposed oscillations, the model was relatively good though the RMS values were higher. For all three testing conditions, the coefficients of correlation of the COM displacements were relatively high between the model and the video method.

APPENDIX

Fig. A1 illustrates a schematic representation of the feet in an arbitrary coordinate system located at O with the axes oriented as in Fig. 1. $\vec{C}_{b-f} = -\vec{C}_{f-b}$ is the moment of the body less the feet acting on the feet at midpoint between the ankles, C. Based on the second equation of motion, the summation of moments about C is

$$\vec{C}_{b-f,C} + \vec{M}_{\vec{F}_{gf},C} + \vec{M}_{\vec{W}_f,C} = -\vec{C}_{f-b-f,C} + \vec{M}_{\vec{F}_{gf},C} + \vec{M}_{\vec{W}_f,C} \vec{0} \quad A1$$

or in it's expanded form

$$\vec{C}_{f-b-f,C} = \vec{P}_a \vec{C} \times \vec{F}_{gf} + \vec{G}_f \vec{C} \times \vec{W}_f \quad A2$$

To illustrate that the COM excursion remains the same regardless of the point about which the summation of moments is taken, equation A2 will be compared to a summation of moments taken about the left ankle at C'.

Moments about C' are

$$\vec{C}_{fb-fC'} = \vec{P}_a \vec{C}' \times \vec{F}_{gf} + \vec{G}_f \vec{C}' \times \vec{W}_f \quad A3$$

Expressing $\vec{P}_a \vec{C}'$ in terms of $\vec{P}_a \vec{C}$ and $\vec{C} \vec{C}'$, ($\vec{P}_a \vec{C}' = \vec{P}_a \vec{C} + \vec{C} \vec{C}'$) and $\vec{G}_f \vec{C}'$ in terms of $\vec{G} \vec{C}$ and $\vec{C} \vec{C}'$ ($\vec{G}_f \vec{C}' = \vec{G} \vec{C} + \vec{C} \vec{C}'$), equation A3 becomes

$$\vec{C}_{fb-fC'} = (\vec{P}_a \vec{C} + \vec{C} \vec{C}') \times \vec{F}_{gf} + (\vec{G} \vec{C} + \vec{C} \vec{C}') \times \vec{W}_f \quad A4$$

Equation A4 is then expanded to

$$\vec{C}_{fb-fC'} = \vec{P}_a \vec{C} \times \vec{F}_{gf} + \vec{G} \vec{C} \times \vec{W}_f + \vec{C} \vec{C}' \times \vec{F}_{gf} + \vec{C} \vec{C}' \times \vec{W}_f \quad A5$$

Equation A5 is basically equation A1 transposed to point C' by a moment transfer. The first two terms of the left-hand side of equation A5 are found in equation A1 while the last two terms correspond to the moment transfer.

Since the moment transfer terms will be taken in the summation of moments about the center of mass of the body less the feet, the point about which the summation of moments is taken will not affect the COM excursion. Consequently, setting the origin between the ankles does not exclude studies involving individuals who usually do not stand with 50% of their weight on each foot.

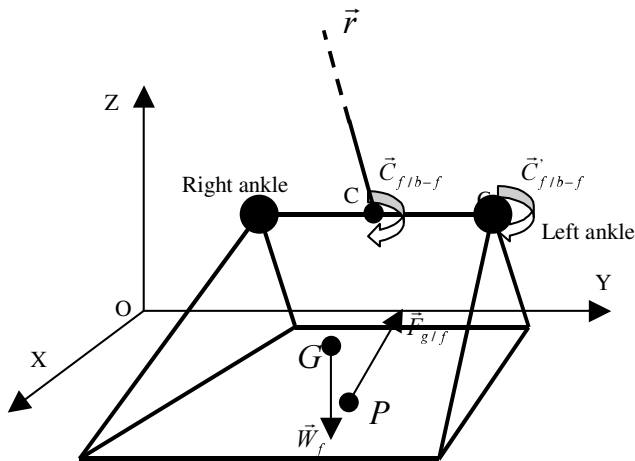


Fig. A1. Free body diagram of the feet illustrating the moment transfer from a point located at the midpoint of the ankles, C to a position lying on the left ankle, C' . The symbols are described in the Appendix.

ACKNOWLEDGMENT

The authors wish to express their gratitude to Mrs. Colette Stérol, Mr Jacques Delacroix and Mr Richard Kazmarek, for their technical assistance. Financial support was given in part by the Région Nord-Pas de Calais (France), Direction Régionale à la Recherche et à la Technologie, the Association of Colleges and Universities of Canada, the Natural Sciences and Engineering Research Council of Canada, The Hospital for Sick Children Foundation and by a French NATO Senior Guest Scientist scholarship awarded to Paul Allard.

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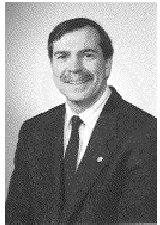
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