A Three-Dimensional Kinematic and Dynamic Study of the Lower Limb During the Stance Phase of Gait Using an Homogeneous Matrix Approach

Nathalie Doriot* and Laurence Chèze

***Abstract*—This paper presents a method to analyze three dimensional kinematics and dynamics of lower limb during walking. The proposed method is based on a homogeneous matrix concept, derived from robotics and using compact, expressive notation convenient for computer applications. The major advantage of this method is that no hypothesis is required on the joint model, which makes it applicable to complex and pathologic joints. Kinematic data are computed from 3-D trajectories of markers collected by a motion analysis system. External forces applied on the leg are measured synchronously during the stance phase of gait. Angular velocity components obtained using the homogeneous matrix method are displayed for three subjects and compared with those obtained from the same experimental data using a helical axis method. Then, intersegmental moments calculated from the inverse dynamic part of the homogeneous matrix method are shown on the same subjects. Kinematic results indicate that there are no significant differences between the methods, thus demonstrating the reproducibility of the stance phase of gait in the sagittal plane. Use of this synthetic homogeneous method developed for both kinematics and dynamics of rigid bodies demonstrates good promise for applications in biomechanics.**

***Index Terms*—Biomechanics, dynamics, kinematics, legged locomotion, three-dimensional displays.**

I. Introduction

IN the study of kinematics and dynamics of the human lower limb, many methods have been developed, first only in the sagittal plane and then in the three-dimensional (3-D) space. The method most widely used to compute 3-D kinematics is based on the modeling of each lower limb joint as a sequence of three hinges. The main disadvantage of this method is that the results depend both on the sequence order: abduction/adduction, rotation, and flexion/extension [1] or flexion/extension, abduction/adduction, and rotation [2], [3]; they also depend on the definition of the axes about which the rotations are expressed: a fixed laboratory frame [1] or mobile axes [2], [3].

Another way to obtain kinematics is based on the helical axis concept which is interesting because the description is unique. The classical method to compute the joint kinematics is to calculate the helical motion associated with the relative displacement of two adjacent segments between two consecutive positions (e. g., displacement of the tibia between positions i and $i+1$, the femur segment being assumed motionless). However, this method is not suited to compute the joint kinematics from video system data, because the displacement between two successive positions is often limited. The results are, therefore, very sensitive to measurement errors. An alternative use of the helical axis concept avoids this problem and gives better results. It consists of characterizing directly the displacement between pelvis/femur, femur/tibia, and tibia/foot at each time [2], [4]. Another method developed in the field of robotics, expanded by Legnani [11], can be applied to biomechanical problems to calculate the kinematics and dynamics of a movement. To validate the kinematic part of this last method, based on an homogeneous matrix concept (HMMd), the angular velocity vector is compared to that computed from the helical axis method which corresponds to the same motion description [12].

As far as the inverse dynamics is concerned, the classical method uses the vectorial equations of Newton–Euler to compute the intersegmental forces and moments using the exterior efforts and the movement measured [1], [5], [6]. The homogenous matrix method, which is also based on an iterative Newton–Euler formulation, presents the advantage of giving directly these intersegmental efforts from simultaneous video and force platform data, without requiring any assumption on the joint kinematics.

II. Materials and Methods

A. Experimental Protocol

The normal gait movement is measured by using a noninvasive opto-electronic Motion Analysis system (Santa Rosa, CA) consisting of six-monochrome cameras synchronized with a six degrees of freedom force platform.

Fourteen retro-reflective markers are stuck on the lower limb over anatomical landmarks: right/left antero/postero superior iliac spines, greater trochanter, medial and lateral epicondyles, tibial tuberiosity, head of the fibula, medial and lateral malleoli, calcaneus, and first and fifth metatarsal heads [Fig. 1(a)].

The 3-D trajectories of these markers are tracked at 60 Hz and the ground reaction forces and moments are sampled at 600 Hz.

The data presented are obtained from six healthy male voluntary subjects $(S1, \ldots, Si, \ldots, S6)$ whose mean age and mean mass are 24 years (from 20 to 28 years) and 77 Kg (from 69 to 85 Kg) respectively. Both interindividual and intraindividual comparisons are displayed.

Two distinct movements are recorded. The first one corresponds to one gait cycle, the subject being asked to walk at com-

Manuscript received March 26, 2002; revised May 1, 2003. *Asterisk indicates corresponding author.*

*N. Doriot is with the Laboratoire de Mécanique de l'Appareil Locomoteur de l'Université Claude Bernard Lyon1, 43 bd du 11 Novembre 1918 69622 Villeurbanne Cedex, France (e-mail: doriot@inrets.fr).

L. Chèze is with the Laboratoire de Mécanique de l'Appareil Locomoteur de l'Université Claude Bernard Lyon1, 69622 Villeurbanne Cedex, France.

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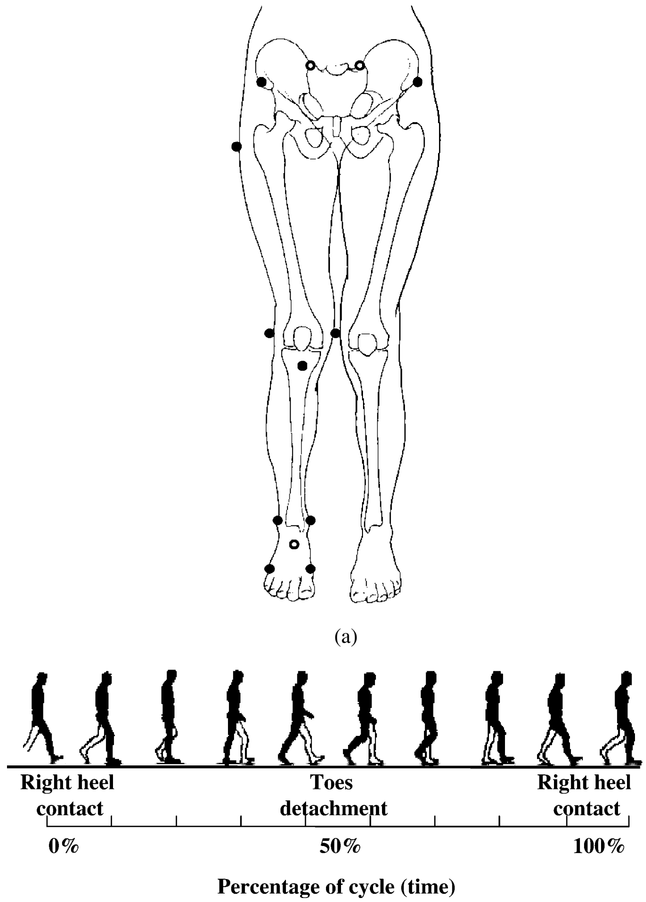
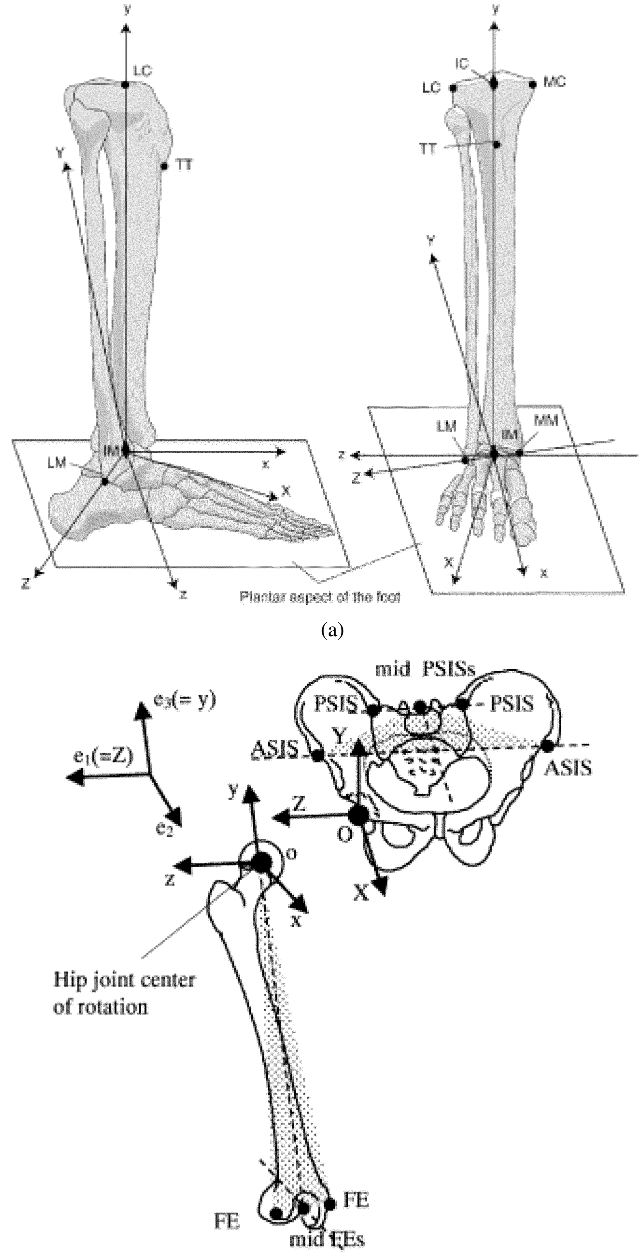


(a)

(b)

Fig. 1. (a) Markers position. (b) Gait cycle (Viel [19]).



(a)

(b)

Fig. 2. (a) Illustration of the tibia/fibula coordinate system (XYZ) and the calcaneus coordinate system (xyz) with the ankle joint complex in the neutral position [10]. (b) Illustration of the pelvic coordinate system (XYZ), femoral coordinate system (xyz), and the joint coordinate system for the right hip joint [10].

fortable speed. Several tests are realized but only those for which the right foot lands suitably on the force plate are analyzed. No particular recommendations are given to the subject concerning steps on the force plate, it's a "free gait" according to Viel [19]. Then the subject performs a circumduction motion of the leg which allows one to determine the hip joint center [7].

B. Kinematics

The 3-D trajectories of cutaneous markers are corrected by a low-pass filtering (Butterworth, fourth-order, with a cutoff frequency of 5 Hz) followed by a solidification of each body segment [8]. This adjustment, segment by segment, is preferred to the global optimization proposed by Lu and O'Connor [9] which enforces the *a priori* choice of a joint model.

As shown in Fig. 2, the segment reference frames are defined according to the International Society of Biomechanics (ISB) recommendation [10].

From the "transformation matrix" ${}^0_i\mathrm{T}$ defining both the orientation and the position of the reference frame $\mathrm{R_i} = (\mathrm{O_i}, \vec{x}_i, \vec{y}_i, \vec{z}_i)$ of segment i with respect to the fixed reference frame $\mathrm{R_0} = (\mathrm{O}, \vec{x}, \vec{y}, \vec{z})$, the matrix $[\mathrm{W}_{i/0}]_{\mathrm{R_0}}$ which describes both the linear and angular velocities and the matrix $[\mathrm{H}_{i/0}]_{\mathrm{R_0}}$ which contains both the linear and angular accelerations are obtained.

Actually, for each body segment i and each gait picture, we determine the following matrices:

$${}^0_i\mathrm{T} = \begin{bmatrix} \vec{x}_i & \vec{y}_i & \vec{z}_i & \mathrm{OO_i} \\ 0 & 0 & 0 & 1 \end{bmatrix}_{\mathrm{R_0}},$$

$$\left[\mathrm{W}_{\frac{i}{0}}\right]_{\mathrm{R_0}} = \begin{bmatrix} \left[\Omega_{\frac{i}{0}}\right] & & & \vec{\mathrm{V}}_{\frac{i}{0}}(\mathrm{O}) \\ 0 & 0 & 0 & 0 \end{bmatrix}_{\mathrm{R_0}} = \left(\dot{{}^0_i\mathrm{T}}\right)\cdot\left({}^0_i\mathrm{T}\right)^{-1}.$$

Note that $\left({}^0_i\mathrm{T}\right)^{-1}$ denotes the inverse matrix of ${}^0_i\mathrm{T}$ and $\left(\dot{{}^0_i\mathrm{T}}\right)$ is the first-order derivative of the matrix ${}^0_i\mathrm{T}$, computed using a centerd fourth-order finite difference, followed by a low-pass filtering.

Let $\vec{V}_{i/0}(O)$ be the velocity of the point O (called the pole), belonging to the body i, that in the considered instant of time is passing through the origin of the reference frame. The "skew-symmetric" matrix $[\Omega_{i/0}]_{R0}$ has the following form:

$$\left[\Omega_{\frac{i}{0}}\right]_{R_0} = \begin{bmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{bmatrix}_{R_0}$$

with

$$\vec{\Omega}_{\frac{i}{0}} = \begin{pmatrix} \omega_x \\ \omega_y \\ \omega_z \end{pmatrix}_{R_0}$$

expressing the angular velocity of the body i with respect to the fixed frame R_0.

The angular velocity vectors $\vec{\Omega}_{i/0}$ are compared with those obtained using the helical axis method.

The matrix

$${}^0_iR = \begin{bmatrix} \alpha_{11} & \alpha_{12} & \alpha_{13} \\ \alpha_{21} & \alpha_{22} & \alpha_{23} \\ \alpha_{31} & \alpha_{32} & \alpha_{33} \end{bmatrix}_{R_0}$$

is deduced from

$${}^0_iT = \begin{bmatrix} & {}^0_iR & & OO_i \\ 0 & 0 & 0 & 1 \end{bmatrix}_{R_0}.$$

From the equality between ${}^0_iR$ and

$$[Q(\theta)]_{R_0} = \begin{bmatrix} {k_x}^2 v\theta + c\theta & k_y k_x v\theta - k_z s\theta & k_z k_x v\theta + k_y s\theta \\ k_x k_y v\theta + k_z s\theta & {k_y}^2 v\theta + c\theta & k_z k_y v\theta - k_x s\theta \\ k_x k_z v\theta - k_y s\theta & k_y k_z v\theta + k_x s\theta & {k_z}^2 v\theta + c\theta \end{bmatrix}_{R_0}$$

with $(v\theta = 1 - \cos\theta)$, the rotation angle θ around $\vec{k}$ of the segment reference frame R_i with respect to R_0 is given by

$$\theta = \cos^{-1}\left(\frac{\alpha_{11} + \alpha_{22} + \alpha_{33} - 1}{2}\right) \quad \text{and}$$

$$\vec{k} = \begin{cases} k_x = \frac{\alpha_{32} - \alpha_{23}}{2\sin\theta} \\ k_y = \frac{\alpha_{13} - \alpha_{31}}{2\sin\theta} \\ k_z = \frac{\alpha_{21} - \alpha_{12}}{2\sin\theta} \end{cases}.$$

At the first position, the frames R_0 and R_i are superimposed.

The comparison between the two methods is demonstrated by Woltring [12]. He showed that the angular velocity vector $\vec{\Omega}_{i/0}$ obtained from the matrix

$$\left[\Omega_{\frac{i}{0}}\right]_{R_0} = [\dot{Q}(\theta)] \cdot [Q(\theta)]^{-1} = \left({}^0_i\dot{R}\right) \cdot \left({}^0_iR\right)^{-1}$$

is equal to $[M(\theta)](d(\vec{k}\theta)/dt)$ with $[M(\theta)] = \vec{k}\vec{k}^t + (\sin(\theta/2)/(\theta/2))([Q(\theta/2)] - \vec{k}\vec{k}^t)$. Finally, the matrix

$$\left[H_{\frac{i}{0}}\right]_{R_0} = \begin{bmatrix} & \left[G_{\frac{i}{0}}\right]_{R_0} & & \vec{\gamma}_{\frac{i}{0}}(O) \\ 0 & 0 & 0 & 0 \end{bmatrix} = \left({}^0_i\ddot{T}\right) \cdot \left({}^0_iT\right)^{-1}$$

is calculated where $[G_{i/0}]_{R_0}$ corresponds to $[\dot{\Omega}_{i/0}] + [\Omega_{i/0}]^2$, and $\vec{\gamma}_{i/0}(O)$ is the acceleration of the pole with respect to the reference frame R_0.

The second-order derivative $({}^0_i\ddot{T})$ is computed in the same way as the one used for $({}^0_i\dot{T})$. Actually the finite difference method involves high-frequency numerical noise which is removed by the two successive filterings.

From the matrices ${}^0_iT$ and $[W_{i/0}]_{R_0}$, one can directly compute the joint kinematics by using the velocity composition rule

$$\left[W_{\frac{i}{i-1}}\right]_{R_0} = \left[W_{\frac{i}{0}}\right]_{R_0} - \left[W_{\frac{i-1}{0}}\right]_{R_0}$$

and then the relative velocity matrix is transformed from the frame R_0 to the local frame R_{i-1}

$$\left[W_{\frac{i}{i-1}}\right]_{R_{i-1}} = \left({}^0_{i-1}T\right)^{-1} \cdot \left[W_{\frac{i}{i-1}}\right]_{R_0} \cdot {}^0_{i-1}T.$$

The (3 × 3) upper-left part of this matrix contains the angular joint velocity $\vec{\Omega}_{i/i-1}$ expressed in the reference frame of the upper segment R_{i-1}.

Note that the expression of $\vec{\Omega}_{i/i-1}$ in the lower segment reference frame is by definition the same.

C. Inverse Dynamics

It consists in a rising strategy, writing the dynamic equilibrium of each body segment. First, the stance foot (on which the external forces are measured) is isolated in order to calculate the intersegmental loads on the ankle, then the shank is isolated to obtain the loads on the knee and so on until the hip.

To implement this method, a generalization of the homogeneous operator in dynamics is proposed by Legnani [11]. For this, three new matrices are introduced.

First, the ground/foot actions measured by the force plates form the matrix

$$[\Phi_{ext./foot}]_{R_0} = \begin{bmatrix} 0 & -m_z & m_y & f_x \\ m_z & 0 & -m_x & f_y \\ -m_y & m_x & 0 & f_z \\ -f_x & -f_y & -f_z & 0 \end{bmatrix}$$

with

$$\vec{F}_{ext./foot} = \begin{pmatrix} f_x \\ f_y \\ f_z \end{pmatrix}_{R_0} \quad \text{and} \quad \vec{M}_{ext./foot}(O) = \begin{pmatrix} m_x \\ m_y \\ m_z \end{pmatrix}_{R_0}.$$

The gravity acceleration matrix is given by

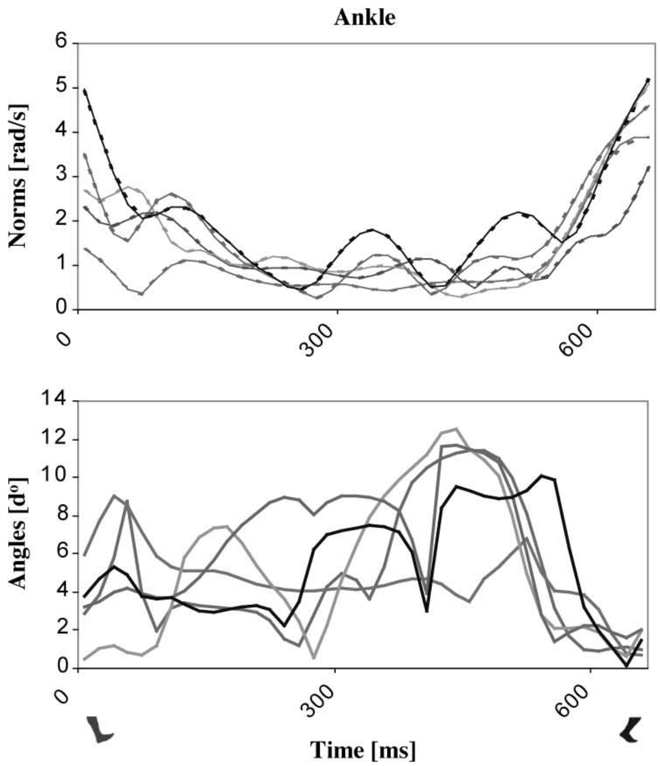
$$\lfloor H_g \rfloor_{R_0} = \begin{bmatrix} & [0] & & \vec{g} \\ 0 & 0 & 0 & 0 \end{bmatrix}.$$

Finally, a "pseudoinertial" matrix $[J_i]_{Ri}$ has to be defined from the mass m_i and the gravity center position G_i. The Zatsiorsky and Seluyanov's regression equations [13] allow one to define the inertial matrix $I(G_i)_{R_i}$. This matrix is transferred to the origin O_i of the segment frame by using the Koënig's theorem

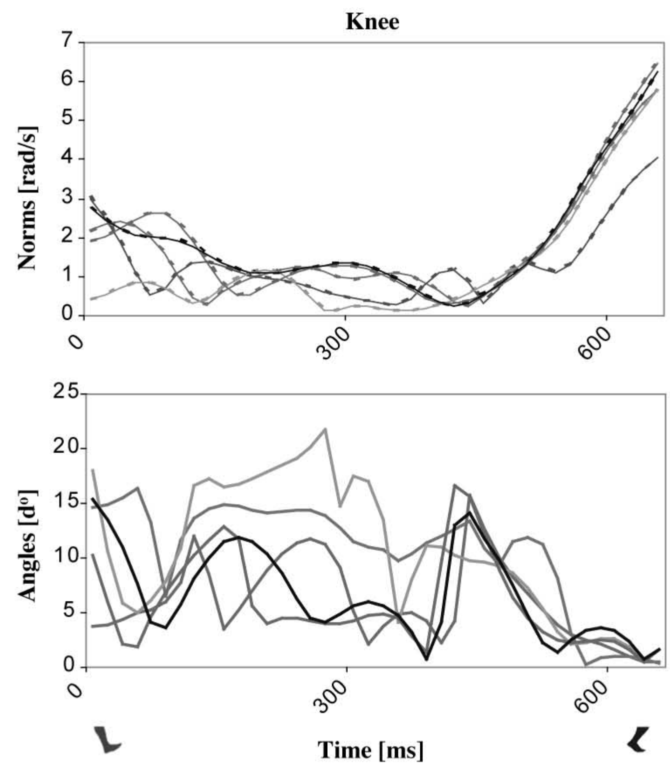
$$I(O_i)_{R_i} = I(G_i)_{R_i} + I_{R_i}$$

where I_{R_i} is the matrix corresponding to the whole mass m_i affected to the point G_i and $I(O_i)_{R_i}$ is the matrix

$$I(O_i)_{R_i} = \begin{bmatrix} A & -F & -E \\ -F & B & -D \\ -E & -D & C \end{bmatrix}.$$



(a)



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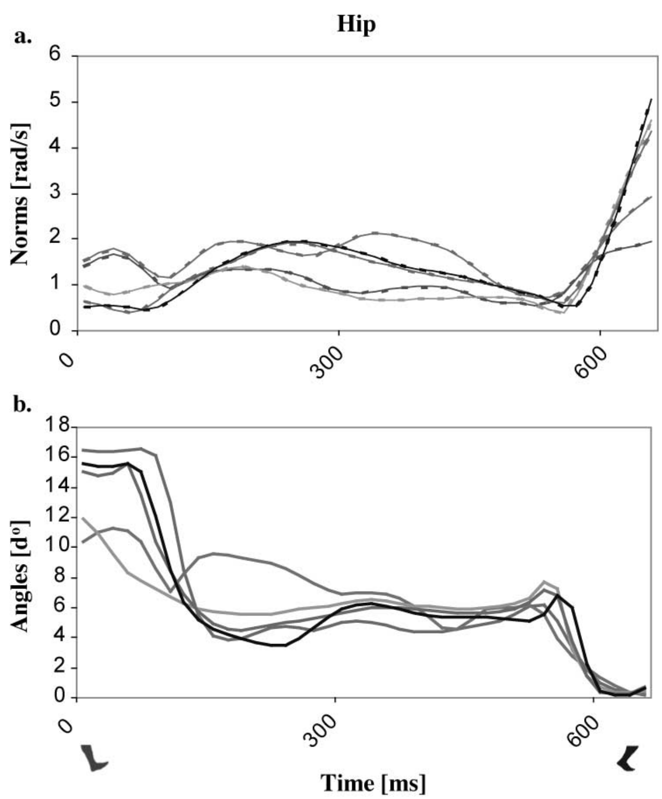

(c)

Fig. 3. Comparison between the two methods; (a) angular velocity vector norms and (b) angles between the two vectors for ankle joint of three subjects during the stance phase of gait. —— HMMD - - -helical axis **S1 S2 S3-1 S3-2 S3-3**.

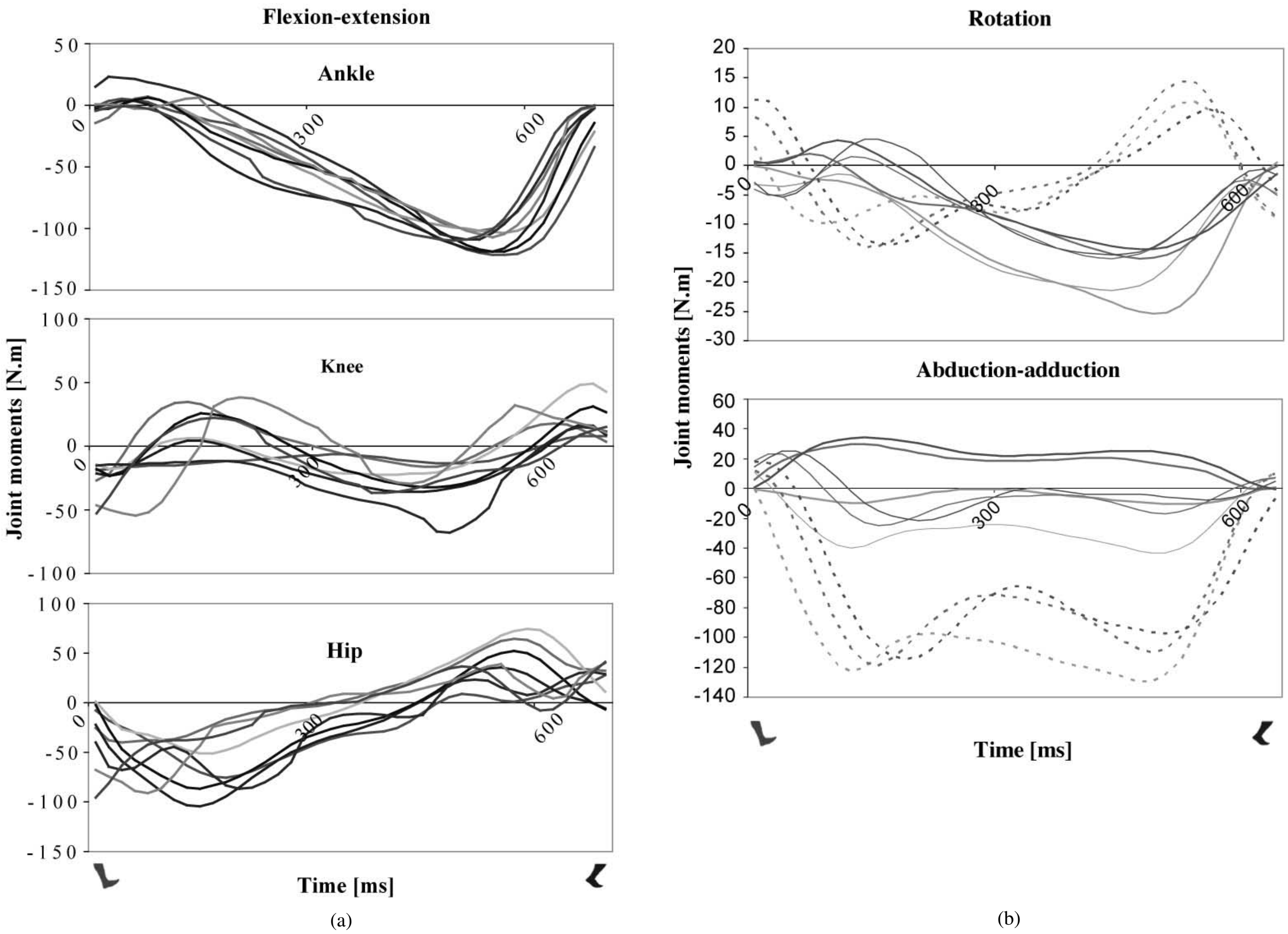

Fig. 4. (a) Flexion/extension joint moments for each joint and subject during the stance phase of gait. **S1 S2 S3-1 S3-2 S3-3 S4 S5 S6**. (b) Rotation and abduction/adduction joint moments for each joint of the subject S3 during the stance phase of gait. —— Ankle —— Knee - - - Hip, **S3-1 S3-2 S3-3**.

The pseudoinertial matrix is finally derived by

$$[J_i]_{Ri} = \begin{bmatrix} \frac{S}{2} - A & F & E & m_iX_i \\ F & \frac{S}{2} - B & D & m_iY_i \\ E & D & \frac{S}{2} - C & m_iZ_i \\ m_iX_i & m_iY_i & m_iZ_i & m_i \end{bmatrix}$$

with

$$S = A + B + C \quad \text{and} \quad (O_iG_i)_{R_i} = \begin{pmatrix} X_i \\ Y_i \\ Z_i \end{pmatrix}.$$

The calculation is performed iteratively as described in the following algorithm.

1) Computation of the absolute acceleration matrix of the segment i.

2) Transfer of the pseudoinertial matrix in the absolute frame.

3) Computation of the matrix $[A_{i/0}]_{R_0}$ containing the dynamic forces and moments of the segment i.

4) Calculation of the matrix $\lfloor \Phi_{g/i} \rfloor_{R_0}$ expressing the action of the gravity on the body i.

Evaluation of the intersegmental loads between adjacent segments $i - 1/i$, by writing the dynamic equilibrium equation of the segment (5) and then transferring the operator in the local frame (6).

$$\left[\Phi_{\frac{4}{5}}\right]_{R_0} = [\Phi_{\mathrm{foot./ext}}]_{R_0}.$$

For $t = 0$, T (stance phase of gait)

For $i = 4$ (foot), 1 (pelvis)

$$\left[H_{\frac{i}{0}}\right]_{R_0} = \left({}^0_i\ddot{T}\right) . \left({}^0_iT\right)^{-1} \tag{1}$$

$$[J_i]_{R_0} = {}^0_iT \, . \, [J_i]_{Ri} \, . \, {}^0_iT^t \tag{2}$$

$$\left[A_{\frac{i}{0}}\right]_{R_0} = \left[H_{\frac{i}{0}}\right]_{R_0} . [J_i]_{R_0} - [J_i]_{R_0} . \left[H_{\frac{i}{0}}\right]^t_{R_0} \tag{3}$$

$$\left[\Phi_{\frac{g}{i}}\right]_{R_0} = [H_g]_{R_0} . [J_i]_{R_0} - [J_i]_{R_0} . [H_g]^t_{R_0} \tag{4}$$

$$\left[\Phi_{\frac{i-1}{i}}\right]_{R_0} = \left[A_{\frac{i}{0}}\right]_{R_0} + \left[\Phi_{\frac{i}{i+1}}\right]_{R_0} - \left[\Phi_{\frac{g}{i}}\right]_{R_0} \tag{5}$$

$$\left[\Phi_{\frac{i-1}{i}}\right]_{R_{i-1}} = \left({}^0_{i-1}T\right)^{-1} . \left[\Phi_{\frac{i-1}{i}}\right]_{R_0} . {}^0_{i-1}T \tag{6}$$

The end for i

The end for t.

III. Results

The upper panels of Fig. 3 shows the kinematic results using both the helical axis method and the HMMd.

In order to compare the gait of several subjects (intervariability) and of one particularly (intravariability), and also to display more legible curves, only three gait recorders of the subject S3 and one for the two first subjects (S1 and S2) are presented in the upper panels of Fig. 3.

The angular velocity vector is characterized by its norm (upper panels of Fig. 3), and its direction. The comparison between the two methods, concerning the direction of the angular velocity vectors, is described by the spatial angle formed by them (lower panels of Fig. 3).

The norms are almost the same for the two methods. Differences are observed between directions for low values of the norm: actually, the maximum angle between the two vectors is about 22° and corresponds to an angular velocity near zero.

Fig. 4 displays the dynamic results obtained from the HMMd.

The general shape of the flexion/extension joint moments are rather similar for all experiments and all subjects [Fig. 4(a)]. It confirms the well-known reproducibility of the stance phase of gait in the sagittal plane for distinct subjects [17]. On the contrary, the motions of rotation and abduction-adduction being particular to each person, the corresponding dynamic results are presented only for one subject (S3).

As far as the flexion/extension moments are concerned, the overall shape of the curves is consistent with data from other investigators [1], [5], [6], [14]–[16].

IV. Conclusion

The homogeneous method developed for both kinematics and dynamics of rigid-body systems has demonstrated favorable results for the biomechanical application discussed in this paper. From the positions of the reference frames embedded on each body segment, and a very concise synthetic formulation, joint angles and moments are calculated in all three planes, which becomes increasingly important when analysing pathological gait for example. No significant differences can be noted between our results and those published by other investigators, even if the used methods are not exactly comparable (local axes definition and calculation methods are different between authors). For this main reason, the ISB have proposed a standardization for both methods and reference frames [7].

The HMMd is valid for the kinematic and dynamic descriptions of a joint, from both the clinical interpretation and result accuracy points of view.

For this paper, the Newton-Euler algorithm based upon homogeneous matrix concept, developed by Legnani in 1996, has been implemented in C and matlab languages in the purpose to determine intersegmental forces and moments between two body segments. Actually, this compact and expressive formulation is very well adapted to matrix oriented programming languages. The software is used in our laboratory for various clinical studies (podiatry [18], rolling chair, walking simulator developing, ...) or sportive applications (rowing).

In the future, the model of the lower limb will be completed, by taking into account the ligaments and the muscles which balance the intersegmental loads between two consecutive body segments. Actually, this further step is necessary in order to finally obtain the contact reaction forces on each joint, which can be useful for example for prosthesis design. Another clinical application can be the detection of muscular control deficiency.

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Nathalie Doriot received the Ph.D. degree for the development of a dynamic model of the gait including muscle and joint contributions, from Lyon1 University, Lyon, France, in December 2001.

She is involved in the European project REALMAN, consisting of integrating technology for dynamic simulation and advancing visualization of human motion in virtual environments, as a Postdoctoral Researcher at the French Institute for Transport and Safety Research, Bron, France.

Laurence Chèze joined the Claude Bernard University, Lyon, France, in 1994 where she is is Associate Professor of Mechanics. She is the head of the university team of the Biomechanics and Human Modeling laboratory.

Her main research interests are the kinematical and dynamical modeling of human movements, in both ergonomical and clinical contexts. She has published 10 papers in international reviews, 12 in national journals, and about 70 communications in international and national symposiums in the biomechanical field.