Journal of Biomechanics 47 (2014) 50–58

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

A 3D lower limb musculoskeletal model for simultaneous estimation of musculo-tendon, joint contact, ligament and bone forces during gait

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ARTICLE INFO

Article history:
Accepted 12 October 2013

Keywords:
Musculoskeletal modeling
Musculo-tendon force
Joint contact force
Ligament force
Bone force
Musculoskeletal structures interaction
Static optimization
Model validation

ABSTRACT

Musculo-tendon forces and joint reaction forces are typically estimated using a two-step method, computing first the musculo-tendon forces by a static optimization procedure and then deducing the joint reaction forces from the force equilibrium. However, this method does not allow studying the interactions between musculo-tendon forces and joint reaction forces in establishing this equilibrium and the joint reaction forces are usually overestimated. This study introduces a new 3D lower limb musculoskeletal model based on a one-step static optimization procedure allowing simultaneous musculo-tendon, joint contact, ligament and bone forces estimation during gait. It is postulated that this approach, by giving access to the forces transmitted by these musculoskeletal structures at hip, tibiofemoral, patellofemoral and ankle joints, modeled using anatomically consistent kinematic models, should ease the validation of the model using joint contact forces measured with instrumented prostheses. A blinded validation based on four datasets was made under two different minimization conditions (i.e., C1 – only musculo-tendon forces are minimized, and C2 – musculo-tendon, joint contact, ligament and bone forces are minimized while focusing more specifically on tibiofemoral joint contacts). The results show that the model is able to estimate in most cases the correct timing of musculo-tendon forces during normal gait (i.e., the mean coefficient of active/inactive state concordance between estimated musculo-tendon force and measured EMG envelopes was C1: 65.87% and C2: 60.46%). The results also showed that the model is potentially able to well estimate joint contact, ligament and bone forces and more specifically medial (i.e., the mean RMSE between estimated joint contact force and in vivo measurement was C1: 1.14BW and C2: 0.39BW) and lateral (i.e., C1: 0.65BW and C2: 0.28BW) tibiofemoral contact forces during normal gait. However, the results remain highly influenced by the optimization weights that can bring to somewhat aphysiological musculo-tendon forces.

1. Introduction

Many musculoskeletal structures (e.g., bones, ligaments, muscles, tendons) are solicited to perform a movement (Pandy and Andriacchi, 2010). The understanding of these structures function and the interaction between them (Cleather and Bull, 2011; Collins and O'Connor, 1991; Pandy and Andriacchi, 2010) remains a challenge that could allow in the future assisting clinicians in term of diagnosis and treatments in case of orthopedic or neurologic disorders. In vivo measurements of musculo-tendon, joint contact, ligament and bone forces exist (Behrmann et al., 2012; Bergmann et al., 2001; Bey and Derwin, 2012; Beynnon and Fleming, 1998; D'Lima et al., 2008; Lu et al., 1998), but the protocols are invasive and inappropriate for a daily clinical use (Fleming and Beynnon, 2004). Consequently, 3D musculoskeletal modeling of the lower limb has been proposed (Al Nazer et al., 2008; Anderson and Pandy, 2001; Cleather and Bull, 2011; Crowninshield and Brand, 1981; Fraysse et al., 2009; Glitsch and Baumann, 1997; Hu et al., 2013; Lenaerts et al., 2008; Moissenet et al., 2012a; Pierrynowski and Morrison, 1985; Seireg and Arvikar, 1975; Stansfield et al., 2003; Wehner et al., 2009) as an interesting alternative and several models have been developed (Arnold et al., 2010; Delp et al., 1990; Klein Horsman et al., 2007). These models allow solving the muscular redundancy problem and estimating the forces transmitted by these musculoskeletal structures during a movement (Chèze et al., 2012; Erdemir et al., 2007). However, the validation of these models is arduous (Lund et al., 2012). Electromyographic (EMG) signals are often used to evaluate the estimated musculo-tendon forces but only provide a qualitative (Dumas et al., 2012; Modenese et al., 2011; Neptune et al., 2001; Selk Ghafari et al., 2009; Stansfield et al., 2003; Thelen and Anderson, 2006) or semi-quantitative validation (Giroux et al., 2013; Kaufman et al., 1991; Prilutsky and Zatsiorsky, 2002). An alternative is to collect joint

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http://dx.doi.org/10.1016/j.jbiomech.2013.10.015

contact forces using instrumented prostheses (Bergmann et al., 2001; Brand et al., 1994; D'Lima et al., 2008; Lin et al., 2010; Stansfield et al., 2003). Indeed, since musculo-tendon forces have a primary role in the joint contact force generation (Herzog et al., 2003), a pertinent estimation of joint contact forces should be a reflection of the quality of the estimated musculo-tendon forces and consequently it should provide a quantitative validation metric for the validation of musculoskeletal models (Lund et al., 2012).

In such a context, it is necessary to ensure a good interaction between musculoskeletal structures in the model in order to assess the quality of the estimated musculo-tendon forces from the joint contact forces data. However, studies are often based on a traditional two-step method (Chèze et al., 2012; Erdemir et al., 2007). First, the musculo-tendon forces are computed using a static optimization procedure, and second, the joint reaction forces (i.e., joint contact and ligament forces) are deduced from the dynamics equation using the optimal muscular solution. Even if this method complies with the forces equilibrium, it does not allow studying the musculoskeletal structures interaction in establishing this equilibrium (Cleather and Bull, 2011). Based on this observation, Lin et al. (2010) proposed an optimization procedure estimating simultaneously musculo-tendon and joint contact forces using a deformable model of the knee. However, ligaments were omitted and the study was limited to the knee joint that may bring a negative impact on the bi-articular musculo-tendon forces estimation (Fraysse et al., 2009). Optimization based methods computing simultaneously musculo-tendon, joint contact and knee ligaments forces have also been proposed recently (Cleather and Bull, 2011; Hu et al., 2013). However, the estimated forces were not validated.

The first aim of this study is to propose a 3D lower limb musculoskeletal model based on a one-step static optimization procedure allowing simultaneous musculo-tendon, joint contact, ligament and bone forces estimation during normal gait. It is postulated that this approach, by giving access to the forces transmitted by these musculoskeletal structures at both hip, tibiofemoral, patellofemoral and ankle joints, modeled using anatomically consistent kinematic models (Di Gregorio et al., 2007; Feikes et al., 2003; Sancisi and Parenti-Castelli, 2011a), should ease the validation of the model using joint contact forces measured with instrumented prostheses. The second aim is to perform a blinded model validation based on the datasets provided for the “Grand Challenge Competition to Predict in Vivo Knee Loads” organized by Fregly et al. (2012).

2. Material and methods

2.1. 3D lower limb musculoskeletal model

A previously described (Dumas et al., 2012; Moissenet et al., 2012a) 3D lower limb musculoskeletal model, consisting of pelvis, thigh, shank and foot segments and 43 muscular lines of action, was extended by adding the patella to perform this study (Fig. 1). Hip, tibiofemoral, patellofemoral and ankle joint kinematic models are all based on anatomical considerations (Fig. 1). Hip joint is modeled by a spherical joint. Tibiofemoral joint is modeled by a parallel mechanism made of two sphere-on-plane contacts (i.e., medial and lateral) and three isometric ligaments (i.e., anterior cruciate ligament – ACL, posterior cruciate ligament – PCL and medial collateral ligament – MCL) (Feikes et al., 2003). The choice of these ligament combination was made for kinematic reasons in order to ensure a 1-DOF model (Ottoboni et al., 2010; Sancisi and Parenti-Castelli, 2011b). Patellofemoral joint is modeled by a hinge joint between the patella and the femur and an isometric ligament (i.e., the patellar tendon – PT) between the patella and the tibia (Sancisi and Parenti-Castelli, 2011a). Ankle joint is modeled by a parallel mechanism made of a spherical joint and two isometric ligaments (between tibia and calcaneus – TiCaL and between fibula and calcaneus – CaFiL) (Di Gregorio et al., 2007). In the same way as for the tibiofemoral joint, this ligament combination is the one that most fittingly models the joint movement. Then, in order to compute muscular lever arms, a widely-used generic musculoskeletal geometric model (Delp et al., 1990) was adapted to our model (Fig. 1).

2.2. Computation framework

First, each segment position is defined using generalized coordinates (Dumas and Chèze, 2007) that consist, for each segment i, in two position vectors (i.e., the proximal P_i and distal D_i joint centers) and two unitary direction vectors (i.e., $\mathbf{u}_i$ and $\mathbf{w}_i$) (Fig. 1)

$$Q_i = [\mathbf{u}_i \mathbf{r}_{P_i} \mathbf{r}_{D_i} \mathbf{w}_i]^T \tag{1}$$

These parameters correspond to a classic set of natural coordinates (Garcia de Jalon and Bayo, 1994). Details of the segment parameters can be found in Dumas and Chèze (2007).

Second, the kinematic constraints $\boldsymbol{\Phi}^k$ and the associated Jacobian matrix $\mathbf{K}^k$ are defined for each joint. Since 12 parameters represent the six degrees-of-freedom of each segment, rigid body constraints $\boldsymbol{\Phi}^r$ have to be considered in addition to the kinematic constraints with the associated Jacobian matrix $\mathbf{K}^r$ (Duprey et al., 2010).

Third, a constrained multi-body optimization (Duprey et al., 2010; Moissenet et al., 2012a) is performed in order to obtain consistent segments positions $\mathbf{Q}$, velocities $\dot{\mathbf{Q}}$ and accelerations $\ddot{\mathbf{Q}}$.

Fourth, the full dynamics equation of the lower limb is written. In contrast with the classical approach, the dynamics equation of the whole kinematic chain is used here (Pennestrì et al., 2007), introducing the musculo-tendon forces and the Lagrange multipliers instead of motor joint moments (Dumas et al., 2012; Moissenet et al., 2012a)

$$\mathbf{G}\ddot{\mathbf{Q}} + \mathbf{K}^T \boldsymbol{\lambda} = \mathbf{E} + \mathbf{L}\mathbf{f} \tag{2}$$

where $\mathbf{G}$ is the generalized mass matrix, $\ddot{\mathbf{Q}}$ is the consistent generalized accelerations, $\mathbf{K} = [\mathbf{K}^k \quad \mathbf{K}^r]$ is the Jacobian matrix of both joint kinematic and rigid body constraints, $\boldsymbol{\lambda}$ is the Lagrange multipliers, $\mathbf{E}$ is the external forces, including both weight and ground reaction forces and moments, $\mathbf{L}$ is the generalized muscular lever arms and $\mathbf{f}$ is the musculo-tendon forces.

Fifth, Eq. (2) gives a direct access to the unknowns only composed of the musculo-tendon forces and the Lagrange multipliers corresponding straightforwardly to the joint contact, ligament and bone forces (Moissenet et al., 2012a). At this level, the traditional two-step method can be used (Moissenet et al., 2012a) when a parameter reduction (Garcia de Jalon and Bayo, 1994) is introduced. This parameter reduction means that all the Lagrange multipliers can be removed from Eq. (2). However, a selection of Lagrange multipliers can also be introduced in the objective function with the partial parameter reduction described below.

2.3. Partial parameter reduction and one-step optimization

The following linear system can be obtained from dynamics Eq. (2):

$$\begin{bmatrix} \mathbf{L} & -\mathbf{K}_1^T & -\mathbf{K}_2^T \end{bmatrix} \begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \\ \boldsymbol{\lambda}_2 \end{bmatrix} = \mathbf{G}\ddot{\mathbf{Q}} - \mathbf{E} \tag{3}$$

where $\boldsymbol{\lambda}_1$ are the Lagrange multipliers that we want to introduce in the objective function, $\boldsymbol{\lambda}_2$ are all the other ones, and $\mathbf{K}_1$ and $\mathbf{K}_2$ the associated Jacobian matrices. The second group of Lagrange multipliers $\boldsymbol{\lambda}_2$ can be canceled from Eq. (3) by projecting the system on the kernel of $\mathbf{K}_2$, using the projection matrix $\mathbf{Z}_{\mathbf{K}_2}$

$$\begin{aligned} &\mathbf{Z}_{\mathbf{K}_2}^T \mathbf{L}\mathbf{f} - \mathbf{Z}_{\mathbf{K}_2}^T \mathbf{K}_1^T \boldsymbol{\lambda}_1 - \mathbf{Z}_{\mathbf{K}_2}^T \mathbf{K}_2^T \boldsymbol{\lambda}_2 = \mathbf{Z}_{\mathbf{K}_2}^T \left(\mathbf{G}\ddot{\mathbf{Q}} - \mathbf{E} \right) \\ &\Leftrightarrow \mathbf{Z}_{\mathbf{K}_2}^T \mathbf{L}\mathbf{f} - \mathbf{Z}_{\mathbf{K}_2}^T \mathbf{K}_1^T \boldsymbol{\lambda}_1 = \mathbf{Z}_{\mathbf{K}_2}^T \left(\mathbf{G}\ddot{\mathbf{Q}} - \mathbf{E} \right) \\ &\Leftrightarrow \mathbf{Z}_{\mathbf{K}_2}^T \begin{bmatrix} \mathbf{L} & -\mathbf{K}_1{}^T \end{bmatrix} \begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \end{bmatrix} = \mathbf{Z}_{\mathbf{K}_2}^T \left(\mathbf{G}\ddot{\mathbf{Q}} - \mathbf{E} \right) \end{aligned} \tag{4}$$

where $\mathbf{Z}_{\mathbf{K}_2}$ is a matrix composed of the eigenvectors of the square matrix $\mathbf{K}_2{}^T\mathbf{K}_2$ corresponding to the null eigenvalues. If necessary, the Lagrange multipliers $\boldsymbol{\lambda}_2$ can be computed a posteriori using the optimal solution $\begin{bmatrix} \mathbf{f} & \boldsymbol{\lambda}_1 \end{bmatrix}^T$ (Moissenet et al., 2012a).

The unknowns $\begin{bmatrix} \mathbf{f} & \boldsymbol{\lambda}_1 \end{bmatrix}^T$, corresponding respectively to the musculo-tendon forces and the selected joint contact, ligament and bone forces, are then introduced in a one-step optimization procedure in order to solve the muscular redundancy problem. A typical static optimization procedure is used and defined as follows (Dumas et al., 2012; Moissenet et al., 2012a, 2012b):

$$\begin{aligned} &\min_{\begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \end{bmatrix}} J = \tfrac{1}{2} \begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \end{bmatrix}^T \mathbf{W} \begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \end{bmatrix} \\ &\text{constraint to : } \begin{cases} \mathbf{Z}_{\mathbf{K}_2}^T \begin{bmatrix} \mathbf{L} & -\mathbf{K}_1^T \end{bmatrix} \begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \end{bmatrix} = \mathbf{Z}_{\mathbf{K}_2}^T \left(\mathbf{G}\ddot{\mathbf{Q}} - \mathbf{E} \right) \\ \begin{bmatrix} \mathbf{f} \\ \boldsymbol{\lambda}_1 \end{bmatrix} \geq \mathbf{0} \end{cases} \end{aligned} \tag{5}$$