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Validation of a multi-body optimization with knee kinematic models including ligament constraints

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ABSTRACT

Motion analysis aims at evaluating the joint kinematics but the relative movement between the bones and the skin markers, known as soft tissue artifact (STA), introduces large errors. Multi-body optimization (MBO) methods were proposed to compensate for the STA. However, the validation of the MBO methods using no or simple kinematic constraints (e.g., spherical joint) demonstrated inaccurate in vivo kinematics. Anatomical constraints were introduced in MBO methods and various ligament constraints were proposed in the literature. The validation of these methods has not been performed yet.

The objective of this study was to validate, against in vivo knee joint kinematics measured by intra-cortical pins on three subjects, the model-based kinematics obtained by MBO methods using three different types of ligament constraints.

The MBO method introducing minimized or prescribed ligament length variations showed some improvements in the estimation of knee kinematics when compared to no kinematic constraints, to degree-of-freedom (DoF) coupling curves, and to null ligament length variations. However, the improvements were marginal when compared to spherical constraints. The errors obtained by minimized and prescribed ligament length variations were below 2.5° and 4.1 mm for the joint angles and displacements while the errors obtained with spherical joint constraints were below 2.2° and 3.1 mm. These errors are generally lower than the errors previously reported in the literature.

As a conclusion, this study presented encouraging results for the compensation of the STA by MBO and for the introduction of anatomical constraints in MBO. Personalization of the geometry should be considered for further improvements.

1. Introduction

Motion analysis based on skin markers aims at evaluating the kinematics of the underlying bones. However, the relative displacement between the markers and the bones, known as soft tissue artifact (STA) (Akbarshahi et al., 2010; Leardini et al., 2005; Peters et al., 2010), introduces large errors and inconsistency in this kinematic estimation. Several methods were proposed to compensate for STA, such as multi-body optimization (MBO) methods (Andersen et al., 2009; Duprey et al., 2010; Lu and O'Connor, 1999; Reinbolt et al., 2005). The latter methods optimize the position and orientation of a multi-body model of the limb to minimize the distances between the model-determined and the measured marker trajectories. The resulting model-based kinematics is thus dependent from the constraints imposed by the model: it is necessary that the kinematic model well represents the general joint kinematics.

The validation of the model-based kinematics obtained with the MBO methods remains limited. Previous studies (Andersen et al., 2010; Stagni et al., 2009) showed that simple kinematic constraints (e.g., spherical or hinge joints) are inefficient for the estimation of accurate in vivo knee kinematics, especially the joint displacements. Li et al. (2012) had the same observation when

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introducing degree-of-freedom (DoF) coupling curves at the joint (Walker et al., 1988) to constrain the bone relative motion. However, recently, the introduction of anatomical constraints in the MBO was proposed, especially by taking into account the articular surfaces and the ligaments. Duprey et al. (2010) used parallel mechanisms with surface contact conditions and ligament length constancy (Parenti-Castelli and Sancisi, 2013) to represent the anatomical constraints of the knee. Three ligaments were considered: anterior cruciate ligament (ACL), posterior cruciate ligament (PCL) and medial collateral ligament (MCL). Gasparutto et al. (2013) further introduced deformable ligaments with a minimal ligament length variation for ACL, PCL, MCL and lateral collateral ligament (LCL). Bergamini et al. (2011) introduced prescribed ligament length variations for ACL, PCL, deep and superficial MCL and LCL as a function of knee flexion angle. As such, the model-based kinematics revealed physiological patterns for the joint angles and displacements (Duprey et al., 2010; Gasparutto et al., 2013) but a validation has not been performed yet.

Therefore, the objective of the study is to validate, against in vivo knee joint angles and displacements measured by intracortical pins, the model-based kinematics obtained by three MBO methods introducing anatomical constraints at the knee. For comparison, MBO methods with no kinematic constraints, spherical constraints, and DoF coupling curves are also studied. For that purpose, the optimization framework of Duprey et al. (2010) is used. It was previously developed to be able to introduce different sets of kinematic constraints and further developments in the present study allowed deformable ligaments to be introduced. In addition, a knee kinematic model allowing the definition of the different ligament constraints on the same geometry was also specifically developed in this study. The previous model (Feikes et al., 2003) used by Duprey et al. (2010) indeed had neither the possibility of adding a 4th ligament nor any information on the lengthening of the ligaments during knee flexion.

2. Material and methods

2.1. Knee kinematic model

The knee kinematic model was based on in vitro studies (Ottoboni et al., 2010; Parenti-Castelli and Sancisi, 2013). It was composed of four ligaments (ACL, PCL, MCL and LCL) and two medial and lateral sphere-on-plane contacts. The model geometry (i.e., sphere centre, plane orientation, origin and insertions of ligaments) was calibrated on cadaver data in order to obtain the best-fit parallel mechanism featuring the ACL, PCL and MCL, and to find the most isometric LCL fibre during passive flexion (Tables 1–2, (Parenti-Castelli and Sancisi, 2013)). Then, the experimentally measured ligament length variations were fitted with polynomial functions of the knee flexion angle θ (Table 3).

2.2. Constraints introduced in MBO

The principle of MBO is to minimize the square of distances between model-determined and measured skin marker positions by considering the kinematic constraints (Andersen et al., 2009; Duprey et al., 2010; Lu and O'Connor, 1999; Reinbolt et al., 2005). MBO was applied to two segments in the current study, namely the thigh and shank. The initial guess used as input for the MBO was computed from the skin markers (i.e., the segment coordinate systems (SCSs) are directly constructed at each sampled instant of time). Six different methods were used. First, no kinematic constraints (*N*) were introduced at the knee joint. This special case of MBO is equivalent to a least square segment pose estimation (e.g., Söderkvist and Wedin, 1993). Second, a spherical joint (*S*) was introduced (Lu and O'Connor, 1999). The centre of the joint was the midpoint between the medial and lateral epicondyles. Third, DoF coupling curves (CC) were introduced (Walker et al., 1988). The other three methods considered anatomical constraints, all featuring the same medial and lateral sphere-on-plane contact conditions, but different ligament models. Accordingly, fourth, the ACL, PCL and MCL with constant lengths (ΔL_0) were introduced (Duprey et al., 2010). No variations of the three ligament lengths, $\tilde{d}^l$ (l=3, 4, 5) given in Table 2, were allowed during motion. Fifth, the ACL, PCL, MCL and LCL minimized ligament length variations ($\Delta L_{\min}$) were introduced by a penalty-based method (Gasparutto et al., 2013). The four reference lengths, $\overline{d}^l(\theta)$ (l=3, 4, 5, 6), were the mean length of each ligament during motion. These lengths were computed by using the polynomial functions in Table 3, with the knee flexion angle, θ, computed from the thigh and shank SCSs as constructed for the initial guess of the MBO. MBO was performed by minimizing ligament length variations with respect to $\overline{d}^l(\theta)$. Sixth, the ACL, PCL, MCL and LCL prescribed ligament length variations (ΔL_θ) were introduced by the same penalty-based method. The four lengths, $d^l(\theta)$ (l=3, 4, 5, 6), were also computed by using the polynomial functions in Table 3, and MBO was performed by minimizing ligament length variations with respect to $d^l(\theta)$.

Table 1
Knee model geometry.

Segment	Anatomical point or orientation vector	Coordinates (in mm) or components in the thigh/shank SCSs *X*	*Y*	*Z*
Femur	Medial condyle centre	0.2	3.4	−23.2
	Lateral condyle centre	−3.3	2.1	26.2
	ACL origin	−6.8	7.5	9.2
	PCL origin	−2.7	−1.1	−2.2
	MCL origin	2.7	5.8	−47.6
	LCL origin	3.2	2.3	36.2
Tibia	Medial tibial plateau	−2.1	−28.6	−19.1
	Lateral tibial plateau	−2.8	−26.1	24.4
	Medial normal	0.067	0.989	−0.127
	Lateral normal	−0.088	0.994	0.061
	ACL insertion	12.8	−26.1	−0.9
	PCL insertion	−25.8	−38.1	−3.5
	MCL insertion	2.1	−117.1	−5.8
	LCL insertion	−24.3	−48.0	37.1

The thigh segment coordinate system (SCS) was defined according to Wu and Cavanagh (1995): the *Y*-axis is the unitary vector from the midpoint between epicondyles to the hip joint centre. The *X*-axis is the unitary cross product between the *Y*-axis and the vector from the medial to the lateral epicondyle. The *Z*-axis is the unitary cross product between the *X*-axis and the *Y*-axis. The origin is the midpoint between the medial and lateral epicondyles. The shank SCS is superimposed to the thigh SCS in anatomical position (0° of flexion).

Table 2
Parallel mechanism parameters.

Medial condyle sphere radius ($\tilde{d}^1$) (in mm)	Lateral condyle sphere radius ($\tilde{d}^2$) (in mm)	Isometric ACL ligament lengths ($\tilde{d}^3$) (in mm)	Isometric PCL ligament lengths ($\tilde{d}^4$) (in mm)	Isometric MCL ligament lengths ($\tilde{d}^5$) (in mm)
32.3	28.3	40.5	43.3	129.7

Table 3
Coefficients of the polynomial interpolation of the experimental ligament lengths.

	ACL (l=3)	**PCL** (l=4)	**MCL** (l=5)	**LCL** (l=6)
a_0^l	40.1	43.8	129.7	57.5
a_1^l	−1.0e−01	1.2e−01	3.0e−02	2.0e−01
a_2^l	−5.6e−04	−1.8e−03	4.2e−03	−8.9e−03
a_3^l	1.3e−04	−3.7e−04	1.8e−04	−1.1e−03
a_4^l	3.6e−06	−1.1e−05	3.2e−06	−3.3e−05
a_5^l	4.2e−08	−1.3e−7	2.3e−08	−4.6e−07
a_6^l	2.3e−10	−7.9e−10	2.9e−11	−3.0e−09
a_7^l	5.4e−13	−1.8e−12	−2.3e−13	−7.6e−12

The seventh order polynomial used to fit the experimental ligament length variations (in mm) with a least-square method has the following form: $d^l(\theta)=a_0^l+a_1^l\theta+a_2^l\theta^2+a_3^l\theta^3+a_4^l\theta^4+a_5^l\theta^5+a_6^l\theta^6+a_7^l\theta^7$ with θ as the flexion angle (in degree, in the range −110° to 0°).