Evaluation of a Cruciate Ligament Model: Sensitivity to the Parameters During Drawer Test Simulation

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The knowledge of how cruciate ligaments stabilize the knee joint could be very useful during the execution of daily living activities for the development of clinical procedures. The objective of this study was to evaluate a cruciate ligament model that could achieve this knowledge while avoiding any destructive measurements in living healthy subjects. Subject-specific geometries and kinematic data, acquired from a living subject, were the foundations of the devised model. Each cruciate ligament was modeled with 25 linear-elastic elements and their geometrical properties were subject specific. The anteroposterior drawer test was simulated, and the sensitivity to the reference length and the elastic modulus was performed. Laxity, anterior, and posterior stiffness were calculated and compared with the literature. The laxity was most sensitive to reference length but fitted the literature well considering the reference length estimated from the subject. Both stiffnesses were most sensitive to elastic modulus variations. At full extension, anterior stiffness overestimated the literature, but at 90° good comparisons with the literature were obtained. Posterior stiffness showed smaller overestimations. The devised model, when properly improved, could evaluate the role of the cruciate ligaments of a living subject during the execution of daily living activities.

**Keywords:** 3-D quasi-static model, in-vivo knee kinematics, subject-specific anatomical geometry

The knee joint is a key structure of the human locomotor system. The main function of the knee joint is expressed by two main characteristics: (a) mobility consists of allowing a wide range of motion of the shank with respect to the thigh, and (b) stability consists of the capability to resist external loads. These apparently conflicting characteristics are contemporaneously achieved by the concurrent action of different active and passive anatomical structures and by the conformity of the articular surfaces. An injury to any of these anatomical structures alters the function of the whole joint. Thus, a good knowledge of the in vivo biomechanical function of each anatomical subunit is of fundamental importance and of great clinical interest for the development of new and effective rehabilitative and surgical procedures. A particular interest for the passive structures is demonstrated by almost 8 million injury-related visits, 478,000 total knee replacements, and 9,000 other repair of cruciate ligaments performed in the United States in 2004 as reported by the American Association of Orthopaedic Surgeons (American Association of Orthopaedic Surgeons, 2004).

Most of what we know about the function of the cruciate ligaments has been provided from in vitro experimental studies under different loading conditions, e.g., during the anteroposterior drawer test or simulated muscle loads (Fukubayashi et al., 1982; Markolf et al., 1996; Sakane et al., 1997; Hoher et al., 1999; Li et al., 1999; Harner et al., 2000; Kanamori et al., 2000). Nowadays, very few in vivo experimental studies (Beynnon et al., 1997; Beynnon & Fleming, 1998; Cerulli et al., 2003) have been performed about the function of the cruciate ligaments. Beynnon et al. evaluated the local deformation of the anterior cruciate ligament in a living subject by means of strain gauges, but did not report the function of the whole cruciate.
In the literature, many models were proposed to reproduce the biomechanical behavior of the cruciate ligaments (Wismans et al., 1980; Blankevoort et al., 1991; Zavatsky & O’Connor, 1992a, 1992b, 1993; Gill & O’Connor, 1996; Lu & O’Connor, 1996; Shelburne & Pandy, 1997; Mommersteeg et al., 1997; Zavatsky & Wright, 2001; Defrate et al., 2004; Li et al., 2005; Moglo & Shirazi-Adl, 2005). Among these, there are many bidimensional models, which were usually aimed to evaluate the role of the cruciate ligaments in the sagittal plane. More complex 3-D models considered different anatomical structures too (e.g., passive and active structures, articular surfaces and contact), but they often featured high computational weight and parameters derived from cadaver and/or nonhomogeneous sources. By contrast, knee models, which featured a high level of subject specificity, were also developed and proposed in the literature (Blankevoort & Huiskes 1996); nevertheless, they were mainly based on extensive mechanical acquisitions performed on a cadaveric knee specimen. Thus, these models were representative of the considered cadaveric knee but not of any other physiological knee. Indeed, this could be the reason why these models were not widely exploited in the clinical context. More recently, Li et al. (2004) proposed a numerical, subject-specific knee model that evaluates the in vivo function of the cruciate ligaments during the weight-bearing knee flexion. However, they studied only static positions of the knee joint and did not evaluate forces expressed by the cruciate ligaments. This evaluation, performed on a selected subject, could be very important in providing new guidelines for the surgical treatment of the ligament injuries.

Since the objective of the present work was focused on the mechanical behavior of the cruciate ligaments, other anatomical structures, such as collaterals and articular surfaces, and their relative interactions were neglected. The cruciate ligaments are passive structures; thus, they develop force depending on the relative kinematics between the femur and the tibia, which were both mathematically defined by the operator (anteroposterior translations) and experimentally acquired on the selected subject (passive flexion).

From the relative kinematics between the bony segments, the length and the deformation of the cruciates can be calculated, but if we want to quantify the forces of the cruciate ligaments, their mechanical properties have to be estimated. In our approach, because invasive measurements were avoided for practical and ethical reasons, no mechanical properties of the cruciate ligaments were available from the selected healthy subject, and thus some mechanical parameters had to be necessarily considered from the literature.

The theory of the quasi-linear viscoelasticity (Fung, 1994) demonstrated that the mechanical behavior of a complex system can be described using several simple mechanical models properly connected. The physiological behavior of the cruciate ligaments was assumed to be reproduced by a model with many linear fibers (25 fibers), instead of few nonlinear ones (Wismans et al., 1980; Blankevoort et al., 1991; Mommersteeg et al., 1997; Moglo & Shirazi-Adl, 2005). Thus, the nonlinear mechanical behavior of each cruciate ligament is provided by the progressive recruitment of the linear fibers, which occurs in function of the relative kinematics between the femur and the tibia. The only mechanical parameter needed in the model to mechanically characterize the cruciates is the elastic modulus. Additional parameters, needed to implement the nonlinear characteristic of the whole cruciate in each of the 25 fibers, would have to be defined by the literature, because they cannot be quantified from the selected subject, further losing subject specificity. And they could introduce more experimental errors.

The purpose of our study was to develop a model of the cruciate ligaments that reproduces the physiological behavior of the knee during the anteroposterior drawer test. Sensitivity of the model with respect to the reference length and the elastic modulus of each ligament fiber was estimated and compared with results reported in the literature.

### Methods

A Caucasian male (height 1.68 m, weight 62 kg, and age 30 years), free of musculoskeletal problems, gave his informed consent. The subject underwent a high-resolution NMR scan of his right knee by means of a 1.5-tesla Gemsow scanner (GE Medical Systems, Milwaukee, WI); see details in Table 1. A qualified operator imposed a slow passive-flexion movement to the knee under analysis, which in approximately 1.5 s went from full extension to about 100° of flexion. The motion was recorded by means of a video fluoroscope (SBS 1600, Philips Medical Systems, the Netherlands) at 10 images per second, keeping the knee inside the fluoroscopic field of view.

For each NMR image, the outer contour of each anatomical structure of interest was detected and outlined with a manual 2-D segmentation technique using the software Amira (Indeed–Visual Concepts GmbH, Berlin, Germany). The resulting stacks of segmented

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<th>Table 1 Parameters of the NMR Scanning Procedure</th>
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images were interpolated, generating external surfaces of the distal femur, of the proximal tibia, and of the anterior and the posterior cruciate ligaments (ACL and PCL respectively; Stagni et al., 2004; Figure 1a, 1b). Using Amira, 3-D anatomical insertion areas of the cruciate ligaments were estimated as the impressions of the cruciate ligament geometries on the external bony surfaces. Each anatomical insertion area was described with a 3-D cloud of points (Figure 1c), which were called anatomical insertion points.

From each anatomical insertion area, composed by \( N \) points, the inertia tensor \( I \) was calculated according to the Equation 1, considering \( m_i \) equal to 1 for all points (Goldstein et al., 2002). Then, the anatomical insertion points were projected on the plane passing through the first and the second principal axes (Corazza et al., 2005).

\[
I = \sum_{i=1}^{N} m_i \begin{bmatrix}
(x_i^2 + z_i^2) & -x_i y_i & -x_i z_i \\
-x_i y_i & (x_i^2 + z_i^2) & -y_i z_i \\
-x_i z_i & -y_i z_i & (x_i^2 + y_i^2)
\end{bmatrix}
\] (1)

On this local plane, the projected anatomical insertion points were outlined automatically with a quadratic equation. Ellipses were obtained for all anatomical insertion areas. For each ellipse, 25 planar modeled insertion points were defined on the elliptical area: 1 in the center of the ellipse, 12 uniformly distributed on the contour of the evaluated ellipse, and 12 uniformly distributed along the contour of an ellipse having the same center and semiaxes half of the previous ones. Finally, 3-D modeled insertion points have been defined fitting the 25 planar modeled insertion points on each 3-D anatomical-insertion area using the thin plate splines method (TPS; Bookstein, 1989; Figure 2a).

The procedure used to join the femoral insertion points with the tibial ones took the anatomical twisting of the fibers into account. Coherently with the physiological external twist of the ACL, the tibial insertion area of this ligament was rotated externally by 90° with respect to the femoral one. In other words, the fiber joining the most anterior insertion point lying on the femoral insertion

**Figure 1** — Reconstructed anatomical geometries: (a) Anterior and (b) posterior view of the bony geometries. (c) Anterior view of the bony geometries and the ligament insertion areas (dotted regions) on the femur and the tibia

**Figure 2** — Definition and connection of the insertion points: (a) A 3-D anatomical insertion area (mesh) and the 25 modeled insertion points (empty circles). Elliptical patterns (schematic ellipses) are maintained after the deformations of the TPS. (b) Anterior and (c) posterior cruciate ligament ordering pattern of the fibers.
area, joined the most lateral insertion point on the tibial insertion area (Figure 2b). On the contrary, the tibial insertion area of the PCL was rotated internally by 90° with respect to the femoral one (Figure 2c).

The accurate 3-D pose of the femur and tibia was reconstructed using an automatic iterative procedure frame by frame. The reconstruction was based on the tangency condition between the fluoroscope projection lines and the surface of the bony geometries. The accuracy of the reconstruction process was assessed in prosthesized knees to be better than 1.5° and 1.5 mm for relative translations and rotations, respectively (Zuffi et al., 1999). In fluoroscopic images of natural knees, the contour of the cortical bone is generally less defined with respect to the contour produced by the metallic components of a knee implant. Nevertheless, the low sampling rate of the fluoroscopic acquisitions (10 fps) used in this work allowed us to obtain images of sufficient contrast to clearly identify and define the projections of the knee bones. Moreover, in the process of the reconstruction of the bony kinematics, the use of MRI-based with respect to more accurate CT-based geometries generated errors similar of the theoretical accuracy of the technique, which were both near approximately 2 mm for translations and 1.5° for rotations (Fregly et al., 2005; Moro-oka et al., 2007).

The value of each anatomical insertion area was known from the NMR data set, which was divided into 25 insertion subareas. Each subarea was proportional to the square of the distance of the modeled-insertion point from its adjacent ones after the TPS deformation. The cross-sectional area $A_j$ of each fiber $j$ was calculated as the mean value between the femoral and the tibial insertion subareas. The sum of all cross-sectional areas of each cruciate ligament was approximately 110 mm² for the ACL and 157 mm² for the PCL.

The reference length $l_{0j}$ of each fiber $j$ was defined according to Goodfellow’s hypothesis (Goodfellow & O’Connor, 1978) as the maximal length reached by each fiber during passive flexion in physiological conditions.

To estimate the stiffness of each fiber, since an estimation of the elastic modulus was not available from the selected subject, mean values and standard deviations were taken from the literature (Butler et al., 1992; Race & Amis, 1994). The considered values were 284 ± 140 MPa and 155 ± 120 MPa for the anterior and the posterior fibers bundle of the ACL (Butler et al., 1992), respectively, and 248 ± 119 MPa and 145 ± 69 MPa for the anterior and the posterior fibers bundle of the PCL (Race & Amis, 1994), respectively. According to the anatomical description of the cruciate ligament insertions provided by Harner and colleagues (1999), each fiber $j$ was assigned to one of the two fiber bundles depending on the position of its 3-D modeled insertion points onto the femoral and tibial anatomical insertion areas. Then the mean value of the elastic modulus $E_j$ was considered for each fiber $j$ of the relative fiber bundle.

Using linear springs, Hooke’s law was considered and the stiffness coefficient $K_j$ of each fiber $j$ was calculated according to Equation 2. The force $F_j$ and the extension $\Delta l_j$ of each spring $j$ were expressed as a combination of the stress $\sigma_j$, the strain $\varepsilon_j$, the reference length $l_{0j}$, the cross-sectional area $A_j$, and the elastic modulus $E_j$.

$$K_j = \frac{F_j}{\Delta l_j} = \frac{\sigma_j A_j}{\varepsilon_j l_{0j}} = \frac{E_j A_j}{l_{0j}}$$

The force vector of each fiber $j$ was

$$\begin{cases} F_j &= -K_j \left( \left\| \vec{l}_j \right\| - l_{0j} \right) \quad \left\| \vec{l}_j \right\| - l_{0j} \geq 0 \\ F_j &= 0 \quad \left\| \vec{l}_j \right\| - l_{0j} < 0 \end{cases}$$

where the vector $\vec{l}_j$ connected, at each frame, the two modeled insertion points of each fiber $j$, from the tibia to the femur.

The mechanical system, composed by the bone and ligament mechanical model, was implemented in ADAMS/View 2005 (MSC Software, Santa Ana, CA). For each position along the passive-flexion path, the pose of the femur was fixed in the global reference system, and a total translation of 20 mm (Markolf et al., 1976; Piziali & Rastegar, 1977; Butler et al., 1980; Race & Amis, 1996) was imposed on the tibia along the anteroposterior (A/P) tibial direction in steps of 0.1 mm. No proximodistal and mediolateral translations, and no rotations were allowed.

The parameters defined by Markolf (Markolf et al., 1976, 1978) were estimated from the A/P component of the forces. These parameters were the laxity, and the anterior and posterior stiffnesses, and they were considered in different studies focused on the modeling of the knee (Wismans et al., 1980; Blankevoort & Huiskes, 1996; Mommersteeg et al., 1996; Race & Amis, 1996; Bendjaballah et al., 1998; Moglo & Shirazi-Adl, 2003). The laxity (Markolf et al., 1978) was defined as the tibial translation necessary to reach a specified level of A/P force, ±100 N or ±200 N. The anterior and the posterior stiffnesses (Markolf et al., 1976) were defined as the slopes of the tangents to the A/P force restraint curve versus the A/P tibial displacement at ±100 N.

The first sensitivity analysis was performed with respect to the reference length. The simulation of the drawer test was repeated varying the value of the reference length $l_{0j}$ estimated directly from the passive-flexion path in the range of ±5% with a 1% step.

The second sensitivity analysis was performed with respect to the elastic modulus. The simulation of the drawer test was repeated 81 times. This was the number of the combinations that we obtained assigning independently to each fiber bundle the elastic modulus equal to the mean value and the mean value ±1 standard deviation reported in the literature (Butler et al., 1992; Race & Amis, 1994).

The A/P tibial restrain force, at 2, 4, and 6 mm of posterior translation, was computed versus knee flexion angle. These computations were compared with the
experimental measurements reported by Race and Amis (1996). All the postprocessing was performed using the software Matlab 7 (MathWorks, Inc.).

Results
Predictably, underestimations of the reference length (~5%) produced bigger A/P forces because of a faster recruitment of the fibers, whereas overestimations of the reference length (+5%) produced smaller forces. The ACL expressed A/P restraint quicker than the PCL at full extension, but, over 20° of flexion, the two cruciate ligaments showed a similar capability to resist A/P translations.

As shown in Figure 3, estimations of the laxity parameter were obtained and compared with the experimental results reported in the literature (Markolf et al., 1981). At 20° of flexion, all variations of the reference length produced laxity values within the experimental range. At the full extension and at 90° of flexion, only overestimations of the reference length greater than 1% produced laxity values out of the experimental range. Thus, the devised model estimated laxity in accordance with the literature when the reference length was obtained from the passive-flexion path of the selected subject. Anterior and posterior stiffness estimations were less sensitive to variations of the reference length parameter and were often close to the experimental mean values (Markolf et al., 1976, 1978, 1984).

During the second sensitivity analysis, we found that only 18 combinations of elastic modulus values, of 81 in total, were actually independents, because the ACL and PCL produced restraint only for anterior and posterior tibial translations, respectively. The anterior and the posterior stiffnesses were the parameters most sensitive to variations of the elastic modulus. Both these parameters were compared with the experimental measurements reported in the literature (Markolf et al., 1978, 1984).

As shown in Figure 4 at 90° of flexion, anterior stiffness reached values quite similar to the experimental results (Markolf et al., 1978, 1984). At the full extension, mean value of the estimated anterior stiffness was approximately three times bigger than the experimental mean values. At 20° of flexion, estimations were about double. The variability of the anterior stiffness depended on the knee flexion angle. The smallest variability was obtained at 45° of flexion, and referring to it, at 20° and at 90° of flexion the variability was more than double, whereas at full extension was about six times greater.

As shown at 90° of flexion in Figure 5, the devised model estimated posterior stiffness values very similar to the experimental measurements. At full extension, the estimated mean value of the posterior stiffness was at the upper bound of the experimental 95% confidence interval, whereas at 20° of flexion, estimations were

![Figure 3](image-url)

Figure 3 — Laxity calculated at ±200N at different flexion angles and superimposed with the experimental data reported by Markolf et al. in 1978 and 1981. Mean values (full circles) plus and minus two standard deviations (vertical bars) are shown for experimental results.
Figure 4 — Anterior stiffness calculated at different flexion angles and superimposed with the experimental data reported by Markolf et al. in 1978 and 1984. Mean values (circles) plus and minus two standard deviations (vertical bars) are shown in figure.

Figure 5 — Posterior stiffness calculated at different flexion angles and superimposed with the experimental data reported by Markolf et al. in 1978 and 1984. Mean values (circles) plus and minus two standard deviations (vertical bars) are shown in figure.
approximately double the experimental results. The trend of the variability of the posterior stiffness was opposite of that of the anterior stiffness. The smallest variability was obtained at full extension and at 90° of flexion. A little more was obtained at 20° of flexion and the largest one was calculated at 45° of flexion.

The laxity parameter was less sensitive to variations of the elastic modulus. Estimations of the laxity obtained at ±200 N were 10.5 ± 0.4 mm at 20° of flexion and 5.7 ± 0.6 mm at 90°, quite close to the experimental measurements of 9.6 ± 2.1 and 7.4 ± 1.9 mm, respectively (Markolf et al., 1984). At full extension, the estimated laxity, 12.1 ± 0.5 mm, was a slightly greater than the experimental one, 7.7 ± 2.3 mm (Markolf et al., 1984).

**Discussion**

A 3-D quasi-static model of the cruciate ligaments was developed using bony geometries and kinematic data acquired from a living subject. Each ligament was modeled with 25 linear-elastic elements, paying attention to the anatomical twist of the fibers. The reference length and the cross-sectional area were derived from the selected subject by means of 3-D video fluoroscopy and NMR, respectively. Exploiting only bioimaging technologies, an estimation of the elastic modulus was not available from the subject; thus, experimental values were considered from the literature (Butler et al., 1992; Race & Amis, 1994). The sensitivity of the model with respect to the reference length and with respect to the elastic modulus was performed. Laxity, anterior stiffness, and posterior stiffness were calculated and compared with experimental measurements (Markolf et al., 1978, 1981, 1984) to validate preliminarily the devised model.

Recently, the contribution of the impingement of cruciate ligaments on bony structures attracted particular attention in sports medicine research. The devised model lacked this modeling aspect. First, this aspect would have significantly increased the computational heaviness of the model without ensuring a commensurate increase in its accuracy. Second, this model was devised to be used in physiological conditions (i.e., during daily living activities and not during sports activities), during which the conditions needed to produce the impingement of the cruciates on the bones usually did not occur.

Despite some simplifications, the devised model demonstrated physiologically meaningful relationships between the relative femur/tibia movements and the tensioning of the cruciate ligaments. The ACL was the only responsible for restraining anterior translations of the tibia; otherwise, only PCL opposed to tibial translations along the posterior direction (Markolf et al., 1976; Mommersteeg et al., 1996; Piziali & Rastegar, 1977). Moreover, it is important to note the different responses of the model under variations of the different mechanical parameters. In fact, as the laxity was more sensitive to variations of the reference length parameter, the anterior and posterior stiffnesses were more sensitive to variations of the elastic modulus. This behavior was expected because it is quite clear that variations of the reference length parameter change the level of the recruitment of each single fiber, but not its stiffness, whereas variations of the elastic modulus parameter change the stiffness but not the recruitment pattern of the fibers.

Before comparing the obtained estimations with the literature, it’s important to notice that the considered in vitro studies were often performed on intact knees from cadavers, and the measured restraint was the result of the concurrent action of different anatomical structures. Several authors (Piziali & Rastegar, 1977; Piziali et al., 1980; Butler et al., 1980; Fukubayashi et al., 1982; Shoemaker & Markolf, 1985; Race & Amis, 1996) measured the effects of the cruciate ligaments resection on the A/P restraint during the drawer test. The A/P stabilization role of the *secondary structures*—i.e., ligaments, capsule, menisci, articular surface and contact—depends on the magnitude of the tibial translation and on the knee flexion angle. The A/P restraint provided by these structures was negligible for A/P translations lower than 5–10 mm (Piziali & Rastegar, 1977; Piziali et al., 1980; Shoemaker & Markolf, 1985; Race & Amis, 1996; Sakane et al., 1997; Vogrin et al., 2000). However, Race and Amis (1996) reported that, at the full extension along the posterior direction, even for small translations of the tibia the *secondary structures* may play a relevant role. Since the proposed model did not include the *secondary structures*, best fittings with the literature were expected at large flexion angles. Regarding the posterior restraint versus the flexion angle (Race & Amis, 1996), the behavior of the experimental results was reproduced by the model from 20° to 90° of flexion, exactly where the *secondary structures* resist less than the cruciate ligaments to A/P loads.

The laxity estimated at ±200 N fitted very well with the experimental results reported by Markolf (Markolf et al., 1978, 1981), specially regarding the high sensitivity of the model with respect to the reference length (Figure 3). Although the reference length was a critical parameter, all estimations—obtained with the reference length estimated from passive-flexion path—were never in disagreement with the literature. As expected, at large flexion angle, both anterior and posterior stiffnesses were reproduced very well by the devised model. At 20° of flexion, posterior stiffness was slightly bigger than the experimental values (Markolf et al., 1978; Markolf et al., 1984), but the comparison was again good at full extension (Figure 5). Anterior stiffness obtained at 20°, and even more the one estimated at full extension, was rather larger than the experimental results (Figure 4) reported in the two in vivo studies (Markolf et al., 1978, 1984). Nevertheless, the estimation was comprised within the measurements reported in the in vitro study of Markolf (Markolf et al., 1976).

Causes of the overestimations observed in the anterior and the posterior stiffnesses could be at least two:
the first one was the not allowed intra–extra rotation of the tibia, whereas the second one was the overestimation of the cross-sectional area. In fact, the tibia can physiologically rotate around its long axis to reach the best orientation of the cruciate ligaments to resist the A/P loads (Fukubayashi et al., 1982; Bendjaballah et al., 1998), whereas this rotation was not allowed in the performed simulations. Moreover, as reported by Harner and colleagues (Harner et al., 1995, 1999), the ligament insertion area should be at least three times greater than the mid-substance cross-sectional area. Thus, the cross-sectional area considered for each ligament fiber could have been overestimated, producing a comparable overestimation on the anterior and the posterior stiffnesses. Both these aspects will be tackled in the next scheduled development of the model to estimate these effects on the mechanical behavior.

Although the proposed model was more simple than other models presented in the literature (Wismans et al., 1980; Blankevoort et al., 1991; Mommersteeg et al., 1997; Moglo & Shirazi-Adl, 2005), similar results were obtained with respect to experimental data. If we consider the difficulty in characterizing soft tissues, even using in vitro measurements as demonstrated by the dispersion of the data reported in the literature, a more complex model would not necessarily imply more precise estimations. Thus, the use of linear-elastic mechanical properties for each ligament fiber seems to be not so wrong. Even more true, if the target is to analyze a living subject without any invasive mechanical tests, excluding obviously in vitro or ex vivo conditions. In fact, during the evaluation of a living subject in physiological conditions, the invasive measurements necessary for the characterization of more complex models, originally developed for ex vivo applications, cannot be performed on the subject. Thus, the potential of such models can be nullified by the errors that can occur in the definition of the parameters, such as when these have to be necessarily considered by measurements performed on a different subject.

This model featured an adequate level of accuracy in combination of a very low complexity. This latter aspect allows us to apply the devised model during the execution of daily living activities. In fact, acquiring the subject-specific kinematics during a daily living activity and applying them to the model, as easily done in this work for the passive-flexion path, it is possible to estimate the role of the cruciate ligaments during the selected motor task.

Future developments of this research will consist in acquisitions of a set of living subjects to evaluate the reliability of the devised methodology. In addition, data based on these individuals will be acquired with a new mechanical apparatus suitably designed to (1) synchronously record forces/torques imposed manually to the tibia by means of a 6-axis load cell, and the bony kinematics by means of the video-fluoroscopy; (2) estimate the elastic modulus on a living subject. The first point will give us the capability to validate the model with experimental results acquired from the same living subject, and the second point will give the model the subject-specific feature.

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References


