A 3D QUASI-STATIC MODEL TO INVESTIGATE BIOMECHANIC FUNCTION OF THE KNEE CRUCIATE LIGAMENTS USING SUBJECT SPECIFIC GEOMETRY AND KINEMATIC DATA

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Abstract: The knee joint is a key structure of the human locomotor system. The modelling approach is the only possible way to estimate the biomechanic function of the different anatomical sub-structures during daily living activities. Subject specific geometry and kinematic data are the foundations of the 3D quasi-static model adopted for the present work. Two knee cruciate ligament models are presented considering the influence of the anatomical twist of the fibres. The loads developed by the modelled ligaments during chair rising-sitting motor tasks were estimated. Two different reference length definitions were considered for each ligament fibre: literature and subject specific experimental reference length. The force ranges obtained strongly depended on the adopted model and the reference length resulted the most sensitive parameter. Larger differences between the Untwisted and the TwoTwisted models were observed in the anterior cruciate ligament model, while smaller differences were observed in the posterior cruciate ligament. In each case, the reference length taken from literature produced higher forces with respect to the subject specific one. The TwoTwisted model was effective in evaluating loads in the knee cruciate ligament during the execution of daily living activities. The results agreed with physiology, although the linearity of the mechanical characteristic assumed for each fibre.

Introduction

In the human knee joint, the harmonious interaction among all its different anatomical sub-units provides the well known mobility and stability characteristics. The knowledge of the biomechanic function of the knee passive structures, like the cruciate ligaments, is of fundamental importance and of great clinical interest for the development of new effective rehabilitative and surgical procedures. This interest is demonstrated by over 8 million of injury related visits for knee symptoms by physicians and in emergency rooms, 381000 total knee replacements and 12000 other repair of cruciate ligaments performed in the USA in 2002 as reported by the American Association of Orthopaedic Surgeons (AAOS) [1].

During its normal function, the knee lets the shank move with respect to the thigh, maintaining the stability under external articular load and torque. This is the result of several contributions: inter-segmental contact loads, ligaments tensioning, muscle forces, inertia of body segments. Thus, if we want to quantify the contribution of each anatomical structure, the only possible way is a modelling approach.

The problem of the knee modelling has been tackled from different points of view and at different levels of complexity. Many two-dimensional models on sagittal plane were proposed in literature and several of these were based on a four–bar linkage modelling approach [2-6]. These models allowed to investigate the function of the knee ligaments only in the sagittal plane in different loading conditions [7;8]. Three-dimensional mathematical and finite elements models were also developed [9-12]. These can include sub-models of anatomical articular surfaces and contact forces, of articular deformations, of different passive structures, like ligaments, capsule and menisci, and of active structures like muscles. Nevertheless, these complex models were unusable in physiological context for their computational weight. The logical evolution of this approach would be the evaluation of a 3D model during the execution of a motor task characteristic of daily living activity [13;14]. In this context, even if the model is designed properly for a specific application, its potential can be nullified by the errors resulting from the anatomical, geometrical and mechanical parameters definition. In the cited papers [2;3;7-9;11;12], these errors were due to disagreement in the origins of parameters and inputs, which were often obtained from different and non-homogeneous sources.

So in this work special attention was paid to the input data and parameter, in particular to the geometry influencing mechanics. Subject specific geometry and kinematics data are the foundations of the 3D quasi-static model adopted. The cruciate ligament models differ in the twisting of the fibres and in the definition of their reference length.

The aim of this study was the evaluation of the biomechanic role, in terms of forces, of the anterior and the posterior cruciate ligaments during a chair rising-sitting motor task.

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Materials and Methods

Subject and experimental acquisitions: The selected subject (male, height 168 cm, weight 62 kg, and age 30 years) underwent a high resolution nuclear magnetic resonance (NMR) scan of his right knee with a 1.5T Gemsow scanner (*GE Medical Systems, Milwaukee, Wisconsin*) [14], as reported in Table 1.

Table 1: The NMR scanning procedure parameters

Parameter	Value
Scanning sequence	Spin Echo (T1 weighted)
Number of slices	54
Pixel spacing	0.037x0.037 (cm·cm)
Scanned region length (across the knee)	15.9 (cm)
Slice thickness	2.5 (mm)
Slice spacing	3 (mm)

The subject performed chair rising-sitting motor tasks while acquired by means of fluoroscopy (*SBS 1600, Philips Medical System Nederland B.V.*) at 10 images per second. The knee under analysis was kept inside the fluoroscopic field of view during the execution of the selected task. Moreover, for the detection of the subject specific fibres reference length [2;3], passive flexion was performed with the help of a qualified operator and acquired by means of the same fluoroscopic set-up. Nine repetitions of the chair rising-sitting task and four continuous passive flexions were acquired.

Knee Geometrical Model: A 3D tiled surface geometrical representation of the distal femur, the proximal tibia, and the insertion areas of the anterior and posterior cruciate ligaments was generated from the NMR dataset using the software Amira (*Indeed - Visual Concepts GmbH, Berlin, Germany*). For each NMR slice, the outer contour of the structures of interest was detected and outlined with an entirely manual 2D segmentation technique. The resulting stacks of contours were interpolated to generate polygonal surfaces of each structure [14], as shown in Figure 1 and Figure 2.

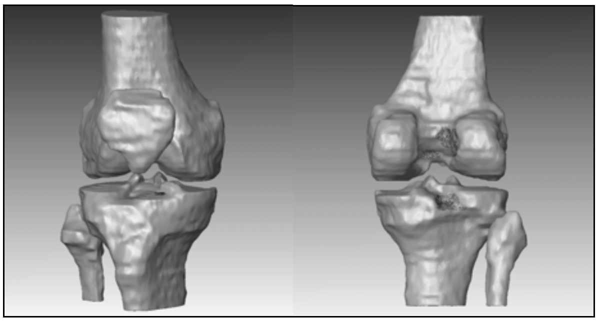

Figure 1: Anterior and posterior view of the complete geometrical knee model. The areas of insertion of ligaments are the dotted regions on the femur and the tibia

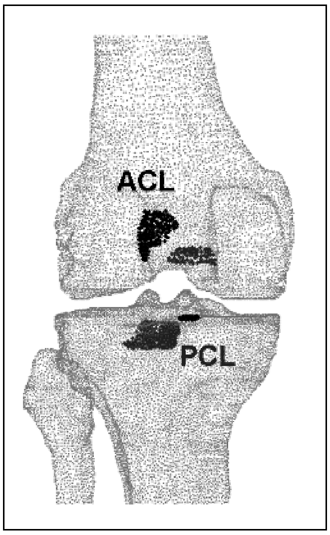

Figure 2: Anterior view of the bony geometry and the ligament insertion areas.

Ligament Geometrical Model: The anatomical insertion areas of both cruciate ligaments were described by a set of points. These were also calculated by the software Amira as prints of the ligament geometrical volume on the 3D bony surface, see Figure 2. The inertia tensor was calculated from each cloud of points and its principal axes and planes were calculated. The anatomical points were then projected on the first principal plane. A quadratic equation for each plane insertion area was estimated to fit the contour line of the projected anatomical points, and in every cases an ellipse was obtained. The planar insertion points were then selected uniformly mapping 25 points on these elliptical areas. The 25 points were distributed: 1 in the centre 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 centre and semi-axes half of the previous ones. The 25 planar insertion points selected on each elliptical area were then fitted on the 3D anatomical insertion area using the “thin plate splines” (TPS) method [15] as shown in Figure 3.

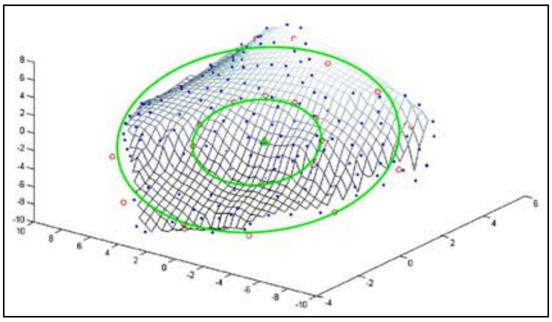

Figure 3: A 3D anatomical insertion area with the 25 fitted by TPS fibre insertion points mapped on two elliptical contours

Two joining method between the femoral and tibial insertion points were considered in this study, taking into account the anatomical twisting of the ligament fibres. The first was called *Untwisted* and the fibres jointed the insertion areas by means of a *consistent* bundle of fibres [16], in other words the anterior fibre on the femoral insertion area was the anterior fibre of the tibial insertion area, and so on for each fibre of the ligament (Figure 4-a). The second was called *TwoTwisted* and it considered the anatomical twisting of the two cruciate ligaments. The anterior cruciate ligament had the order of the fibres on the tibial insertion area rotated by 90° laterally with respect to the femoral insertion area coherently with the anatomical external twist of the ligament. The posterior cruciate ligament had the order of the fibres on the tibial insertion area rotated by 90° medially with respect to the femoral insertion area [16] (Figure 4-b).

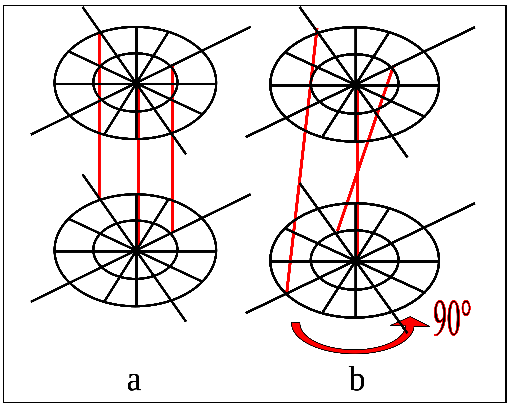

Figure 4: Untwisted (a) and twisted (b) ordering pattern of the fibres for the Untwisted and TwoTwisted model.

Kinematics: The accurate 3D bone pose in space was reconstructed by means of an iterative procedure using a technique based on tangent condition between the projection lines and the surface of the geometrical model. The accuracy of this technique was assessed to be 1 degree for rotations and 1 mm for translations [17]. The two more repeatable passive flexion motions were used for the evaluation of the reference length value, as better described in the next paragraph.

Mechanical Ligament Properties: The two cruciate ligaments, in both models, were modelled with 25 different linear-elastic elements . The elastic modulus *E* of each ligament was the same and it was considered constant from literature equal to 175 MPa [3]. The reference length l_{0j} of each fibre *j* was defined in two different ways: the first length dataset was considered constant from literature (RL_{lit}), equal to 37 mm for the anterior cruciate ligament and to 35 mm for the posterior cruciate ligament fibres [3]; the second was defined according to O'Connor's hypothesis [18], as the maximal length RL_{flex} reached by each fibre during passive flexion. From the NMR dataset the total insertion area was known. The relative cross-sectional area A_j for each fibre was calculated proportionally to the distance of each modelled insertion point from its adjacent ones.

The stiffness coefficient K_j was calculated for each fibre *j* with the equation (1) where *E*, l_{0j} and A_j were the variables mentioned above.

$$K_j = \frac{E \cdot A_j}{l_{0j}} \tag{1}$$

The force expressed from each fibre was shown in equation (2) where ΔL_j was the difference between instant length l_j and the reference length l_{0j} of the fibre.

$$F_j = -K_j \cdot \Delta L_j \tag{2}$$

The total ligament force was the vectorial sum of all fibre forces of the ligament. Obviously, the force expressed by each fibre was imposed to be zero if the distance between its two insertions was smaller than the reference length.

Simulation and Post-Processing Tools: The mechanical system, composed from the knee and the ligament geometrical models and the mechanical properties of the ligament, were implemented and animated with the acquired kinematics in ADAMS/View 2005 (*MSC.Software Corporation 2 MacArthur Place Santa Ana,CA 92707 USA*). This simulator of mechanical systems allowed to estimate each variable in the model, in particular, for each relative position between the femur and the tibia. The three components of the forces, anterior-posterior, proximal-distal and medial-lateral NMR projections, and the magnitude for each fibre were calculated and exported for both cruciate ligaments considering the Untwisted and the TwoTwisted models and the RL_{lit} and the RL_{flex} data sets. Post-processing elaborations were computed with Matlab 7 (*The MathWorks, Inc, MA 01760-2098*). The three components were set to zero when the magnitude of the force was positive (compression), see equation (2). All these forces were transposed to the anatomical tibial reference system. Mean and standard deviation were calculated for each force component versus knee flexion-extension angle.

Results

Anterior cruciate ligament: The three components of the force showed a similar behaviour for each analysed modelling condition. The Untwisted model, considering the RL_{lit} values set, developed the maximal force in maximal extension along all direction, and its action decreased with knee flexion (Figure 5). The TwoTwisted model, considering the same reference length values set, generated little force decreasing from 0° to 35° flexion angle. From 40° flexion angle this model didn't produce any forces increasing the flexion angle. In both models no marked differences were observed between flexion and extension movements.

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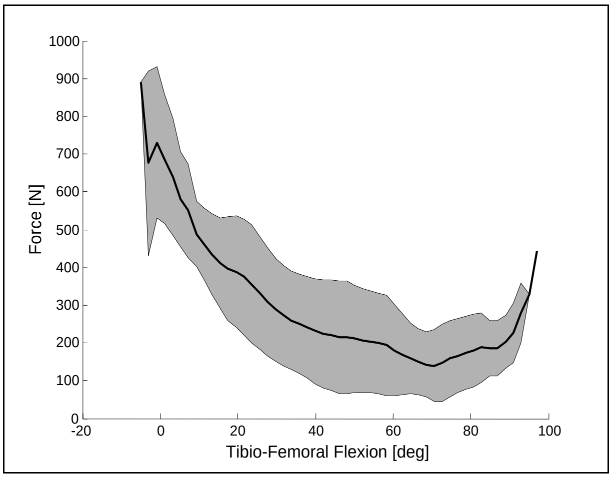

Figure 5: Mean anterior-posterior force component plus and minus a standard deviation of the anterior cruciate ligament RL_{lit} Untwisted model (flexion motor task).

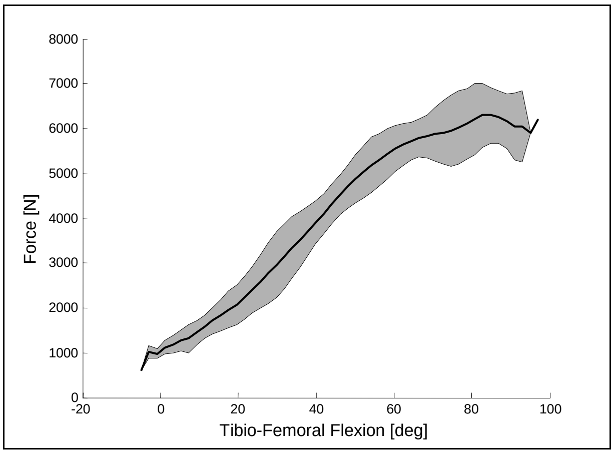

Figure 6: Mean anterior-posterior force component plus and minus a standard deviation of the posterior cruciate ligament RL_{lit} TwoTwisted model (flexion motor task).

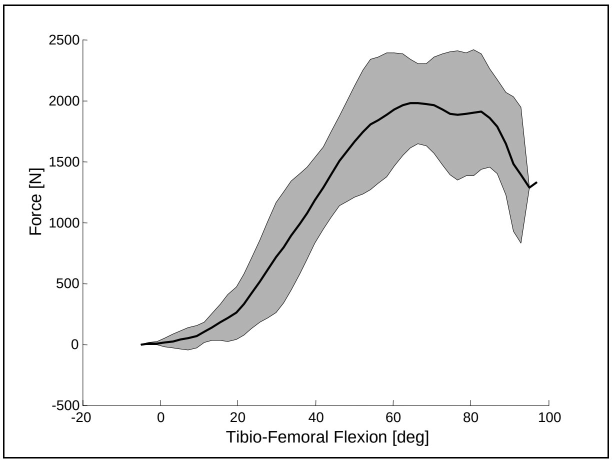

Figure 7: Mean anterior-posterior force component plus and minus a standard deviation of the posterior cruciate ligament RL_{flex} TwoTwisted model (flexion motor task).

The Untwisted model, considering the RL_{flex} values set, expressed its maximum force at the full extension. The forces quickly decreased until reaching small forces at 10°. From 10° to 80° were expressed quite little and constant forces. At 80° flexion angle a little forces recovery was noticed. The whole behaviour described was much evident in flexion with respect to extension. Big differences were obtained with the TwoTwisted model, where null forces were noticed in every component and at each flexion-extension angle.

Posterior cruciate ligament: The Untwisted and TwoTwisted RL_{lit} models were very similar each other in each component except to the force range. In the anterior-posterior direction all models generated an increasing force from the full extension until the full flexion. In the Untwisted model smaller forces were calculated with respect to the TwoTwisted model. An high repeatability in both flexion and the extension motor tasks was noticed in both model (Figure 6). In the proximal-distal direction the global behaviour in both models were quite similar but a low repeatability was observed. Force increased from the full extension until reaching the maximum value at about 50° flexion angle. From 50° to 90° force was quite constant for the Untwisted model, whereas in the TwoTwisted model force decreased lightly. In the medial-lateral direction, in both models, forces increased from the full extension to 60° flexion angle and decreased from 60° to full flexion. Two models had two differences along medial-lateral direction. First, in the Untwisted model the starting force was null, whereas the TwoTwisted model generated already at the full extension about the maximum value reached by the Untwisted model. Second, the force value reached at the full flexion by the TwoTwisted model was equal to the its starting value, whereas in the Untwisted model was smaller of the starting value of an half of the maximum value reached. Like in anterior-posterior direction the TwoTwisted expressed bigger forces with respect to the Untwisted model. With the RL_{flex}, big differences were observed between both models, but no between the three components. The Untwisted model didn't expressed forces from 0° to 40° flexion angle. From 40° very little and noisy forces was obtained. Whereas the TwoTwisted model showed very little forces from 0° to 20° flexion angle. Then forces increased until reaching a force plateau from 50° to 70°. Increasing flexion angle forces decreased lightly, see Figure 7. This last behaviour was fewer visible in the extension motor task for the anterior-posterior and the proximal-distal components. In these two last cases was also evident a lower repeatability.

Discussion

Anterior and posterior cruciate ligament geometrical models were implemented using parameters from a single selected subject. The mechanical effect of the anatomical twist of the ligament fibres and the influence

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of the definition of their reference length were investigated.

Modelling the anterior cruciate ligament relevant differences were noticed between the Untwisted and the TwoTwisted model: the Untwisted model was always active and ready to develop force. With RL_{lit} the developed forces resulted twice as much those developed with $RL_{flex.}$ On the other hand, the TwoTwisted model produced no force considering the RL_{flex} value. With RL_{lit} the TwoTwisted model produced little force contribution was noticed from full extension to 40°, but no fibre recruitment from 40° to full flexion. More force was developed considering the Untwisted model. The results obtained with the TwoTwisted model considering the RL_{flex} were physiologically meaningful: during the execution of living activities, like the chair rising-sitting, the continuous co-contraction of the femoral quadriceps and biceps substitute the stabilizing action of the anterior cruciate ligament [7].

Modelling the posterior cruciate ligament qualitative differences were observed between the Untwisted and the TwoTwisted model only considering the RL_{flex} parameter. The Untwisted model showed a no activity from full extension to 40° and a limited and noisy contribution from 40° to full flexion. Using the RL_{lit} parameter no qualitative differences were observable between two models. Nevertheless the TwoTwisted model developed bigger forces until a factor of three, in the medial-lateral component with respect to the Untwisted. The TwoTwisted model considering the RL_{lit} parameter developed bigger forces until a factor of five with respect to the same model considering RL_{flex}. Whereas the Untwisted model, considering RL_{flex} or RL_{lit}, developed bigger qualitative and qualitative differences. Also in this case, the TwoTwisted model with RL_{flex} produced most physiologically meaningful results: the force plateau showed from 50° to 70° agreed with the flexion range considered for the stabilizing action of the posterior cruciate ligament in a lot of prosthesis models.

The reference length values obtained from the passive flexion kinematics were verified in previous validation work, using anterior-posterior drawer test at different flexion angle. These geometrical knee models reproduced qualitatively and quantitatively very well experimental data from literature.

Conclusions

The proposed models were effective in evaluating strains and loads in the anterior and posterior cruciate ligament during the execution of daily living activities. Both models were very sensitive to the reference length parameter and very large differences between Untwisted and TwoTwisted were achieved.

The modelling approach pulled attention towards the importance of the geometrical arrangement of the ligament fibres in term of twist. The anatomical twist must be modelled in order to replicate the specific mechanical action developed by the ligaments for the stabilization of the knee joint.

As expected, although the linearity of the mechanical characteristic assumed for each fibre, considering the more anatomical fibre twist, in TwoTwisted model coupled with reference length calculated from the specific subject more physiologically meaningful results were obtained.

In the future the simplicity and the specificity of these geometrical knee models will allow to estimate the knee cruciate ligament forces during whatever living motor task in every clinical subject. This will be very important to advise clinicians in ligament reconstructions: before to choose the better ligament insertion point and the reference length of the ligament; after to choose the better rehabilitative procedure.

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