A GLOBAL VERIFICATION STUDY OF A QUASI-STATIC KNEE MODEL WITH MULTI-BUNDLE LIGAMENTS*

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Abstract—The ligaments of the knee consist of fiber bundles with variable orientations, lengths and mechanical properties. In concept, however, these structures were too often seen as homogeneous structures, which are either stretched or slack during knee motions. In previous studies, we proposed a new structural concept of the ligaments of the knee. In this concept, the ligaments were considered as multi-bundle structures, with nonuniform mechanical properties and zero force lengths. The purpose of the present study was to verify this new concept.

For this purpose, laxity characteristics of a human knee joint were compared as measured in an experiment and predicted in a model simulation study. In the experiment, the varus–valgus and anterior–posterior laxities of a knee-joint specimen containing the ligaments and the articular surfaces only, were determined. From this knee-joint, geometric and mechanical parameters were derived to supply the parameters for a three-dimensional quasi-static knee-joint model. These parameters included (i) the three-dimensional insertion points of bundles, defined in the four major knee ligaments, (ii) the mechanical properties of these ligament, as functions of their relative insertion orientations and (iii) three-dimensional representations of the articular surfaces. With this model the experiments were simulated. If knee-model predictions and experimental results agree, then the multi-bundle ligament models are validated, at least with respect to their functional role in anterior–posterior and varus–valgus loading of the joint.

The model described the laxity characteristics in AP-translation and VV-rotation of the cadaveric knee-joint specimen reasonably well. Both display the same patterns of laxity changes during knee flexion. Only if a varus moment of 8 N m was applied and if the tibia was posteriorly loaded, did the model predict a slightly higher laxity than that measured experimentally.

From the model–experiment comparisons it was concluded that the proposed structural representations of the ligaments and their mechanical property distributions seem to be valid for studying the anterior–posterior and varus-valgus laxity characteristics of the human knee-joint.

Keywords: Knee; Model simulation; Knee ligaments; Knee laxity; Biomechanics.

INTRODUCTION

Human knee ligaments are complex multi-fiber structures. In concept, however, these ligaments are considered, too often, as purely uniaxial homogeneous structures, which are either stretched or slack. In studies evaluating the tensile behavior of the ligaments, for example, uniaxial tensile tests were performed (Butler et al., 1978; Kennedy et al., 1976; Noyes et al., 1978). Tensile load and elongation were usually measured in only one direction. In functional analyses, forces were determined in whole ligaments (Markolf et al., 1993), in selected parts of the ligaments (Ahmed et al., 1987), or in two or three fiber bundles (Blomstrom et al., 1993; Takai et al., 1993). In surgical reconstructions, anterior cruciate ligaments were replaced by grafts that act more-or-less as single bundles. Representations or reconstructions of the ligaments which ignore the normal fiber-bundle organisations of the ligaments have been shown to be inadequate to reproduce normal functional behavior.
In the analytical determination of the loading and unloading patterns of the tensile elements in the ligaments, some implicit assumptions were made. First, the bone-ligament-bone tests were performed in a series of variable relative orientations of the bones, which do not necessarily represent anatomic, in situ orientations (Mommersteeg et al., 1995a). When evaluating the functional role of the ligaments, it is important to note that the properties determined in these bench tests are unaltered for in situ orientations of the ligaments. Second, it was assumed that all fiber bundles, except those of the MCL, can be represented by straight lines running from the tibial to the femoral insertion site in all positions of the knee without piercing any bone. For the MCL the bony interaction with the proximal edge of the tibia was taken into account. Third, it was assumed that the interaction between fiber bundles can be neglected. Blomstrom et al. (1993) and Takai et al. (1993) measured the effect of the interaction between the anterior and the posterior fiber bundles on the behavior of the anterior cruciate ligament during anterior tibial loading. Although only minor effects were noted, these effects might play a role for loading configurations or bundle definitions not considered by the authors of these two papers.

The aim of the present study was to determine whether the new structural concept for modeling knee ligaments, with the underlying assumptions as outlined above, is suitable to describe joint laxity of an experimental analogy. For this purpose, experimental measurements of knee laxity at varus–valgus and anterior–posterior loading were compared to model predictions, using a three-dimensional mathematical knee simulation model. The model input parameters were obtained from the experimental specimen.

METHODS AND MATERIAL

The methods involved experiments and computer simulations. The protocol was to determine (i) the joint motions associated with varus–valgus and anterior–posterior loading in a motion rig, (ii) the geometry of the femoral and tibial articular surfaces, (iii) the mechanical behavior of the ligaments in various relative positions of their insertion sites in a series of tensile tests, (iv) the three-dimensional coordinates of the insertion sites of several bundles defined in the four major knee ligaments, and (v) the parameters which describe the mechanical behavior of the ligaments as a function of their relative insertion orientations; the mechanical properties of the cartilage were derived from literature values (Mow et al., 1982; Walker and Hajek, 1972). Finally, (vi) the geometric and mechanical data were used to supply a three-dimensional mathematical model of the knee (Blankenvoort et al., 1991). The geometric and mechanical data were combined into a mathematical knee model. A graphic representation of this three-dimensional geometry of the articular surfaces was measured with an accuracy of 50 μm (de Lange et al., 1985).

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loads and kinematic constraints. These kinematic constraints were in the present study axial and flexion–extension rotations as determined in the experiment by RSA (i). The remaining degrees of freedom of motion are unconstrained. The positions of the femur relative to the tibia are found by solving a set of quasi-static equilibrium equations containing the external forces and moments acting on the femur and the tibia, constrained forces and moments, ligament loads and articular contact loads. The model accounts for deformable articular surfaces (Blankevoort et al., 1991) and for an interaction between the medial collateral ligament and the bone (Blankevoort and Huiskes, 1991). The origin of the tibial coordinate system is located at the most posterior apex of the tibial insertion site of the ACL. The origin of the femoral coordinate system is located 15 mm proximal to this point in the extended knee-joint. The ligament forces \( F_j \) and moments \( M_j \) were summations of the forces \( F_j \) and moments \( M_j \) in its tensile elements \( j \), according to

\[
F_j = \sum F_j, \quad M_j = \sum M_j.
\]

The bundle force \( F_j \) and moment \( M_j \) acting on the tibia are expressed by

\[
F_j = F_j y_j, \quad M_j = D \times F_j,
\]

where \( y_j \) is the unit vector pointing from the tibia to the femoral insertion site of line element \( j \) and \( D \) is the vector pointing from the origin of the femoral coordinate system to the tibial insertion site of the line element \( j \). The tensile force \( F_j \) in a line element \( j \) is assumed to vary according to the square of the strain \( \varepsilon_j \) of this line element as

\[
F_j = k_j \varepsilon_j^2, \quad \varepsilon_j > 0, \quad F_j = 0, \quad \varepsilon_j \leq 0,
\]

in which \( k_j \) is a parameter which defines the structural stiffness 2ks at each strain level and \( \varepsilon_j \) is the strain in line element \( j \), which is calculated from its actual length \( L_j \) and its zero-force length \( L_{0j} \) according to

\[
\varepsilon_j = (L_j - L_{0j})/L_{0j}.
\]

The actual length \( L_j \) follows directly from the kinematic variables (i) and the coordinates of their insertion sites (iv).

For each flexion step, the anterior–posterior laxities as well as the varus–valgus laxities were calculated and compared to the experimental results.

**RESULTS**

At both combined anterior–posterior and axial loading, and combined varus–valgus and axial loading, the model gave reasonable predictions of the experimental results. In general, slightly higher deviations between model and experiment were found at lower load levels than at higher load levels. When applying anterior forces of 50 and 100 N to the tibia, the anterior translations were very well described by the model (Fig. 2). For posteriorly directed forces, the predictions of the posterior translations were approximately 2 mm higher than measured experimentally, during the whole range of flexion. Applying a varus moment of 8 Nm, the model predictions of the varus rotations were higher than the experimental results, in particular in extension of the knee [Fig. 3(a)]. Applying a valgus moment of 8 Nm, the model predictions of the valgus rotations were slightly lower in 30° of flexion [Fig. 3(a)]. For higher load levels, the varus- and valgus-rotations were predicted very well by the model, except for the valgus rotation at 30° of flexion [Fig. 3(b)].
Fig. 3. The varus–valgus rotations as a function of flexion, while applying varus and valgus moments of 8 N m (a) and 16 N m (b) to the tibia for the model simulations (solid lines) and the experimental specimen (data-points). Laxity values were predicted discretely by the model at the same flexion angles as in the experiment. Between these points interpolations were performed. The model predictions of the varus and valgus rotation at 8 N m moment application were higher than the experimental results in extension of the joint. For other joint angles and at 16 N m load application, the model and experiment agree well.

Similar variations in total laxity of the knee were found during knee flexion for the model simulation and the experiment. In both cases, the total anterior–posterior laxity was maximal at 30° of flexion and decreased towards full extension and further flexion (Table 1). The varus–valgus laxity increased from 0° to 30° of flexion (Table 2). In general, the model predicted slightly higher laxities than those measured.

**DISCUSSION**

Mathematical models are simplified representations of reality by which some aspects of the often complex reality become comprehensible. The knee-joint model in the present study is confined to the femoral and tibial articular surfaces and the knee-joint ligaments. Before the model can be applied to understand the mechanisms by which the ligaments and the articular surfaces stabilize the knee joint, or to design new knee or ligament prostheses, the model representations of these structures have to be verified. The only way to perform this task is to integrate mathematical modeling and experimental testing. A physical analogy of the mathematical model has to be constructed, its geometry and mechanical parameters determined and its mechanical behavior compared to that of the mathematical model. This was performed in the present study, with the aim to experimentally verify a new ligament representation, in which fiber bundles were considered as the functional units. In a former version of the model (Blankevoort, 1991), ligament stiffnesses were assumed to be uniformly distributed among the line elements in each ligament. The advantage of the present ligament representation is that it allows for the implementation of nonuniform mechanical characteristics and fiber orientations of each ligament in knee-joint models. Future improvements of this model should be directed to the inclusion of the time-dependent behavior of the ligaments and the mechanical behavior of the menisci, the capsule and the patellofemoral complex. Because these structures were omitted here, the results presented cannot be compared to in vivo laxity values.

In the past, mathematical knee-joint models have been verified only to a certain extent. In most cases, model results were compared to average experimental data reported in the literature (Wismans et al., 1980; Essinger et al., 1989). Only Blankevoort and Huiskes (1996) compared experimental and numerical results for individual knee specimens, although not all
Model parameters were derived from the intact experimental specimen, as done here. The recruitment parameters were identified in an optimization process based on the objective that the knee model behaves like an intact knee-joint specimen. The strength of the present validation study is that the experimental and mathematical models were analogous with respect to both geometry and mechanical parameters, hence comparable. These parameters were specific for the ligaments in the particular knee specimen of which the motion characteristics were described by the model.

This validation study is limited because we used only one knee-joint specimen. This restriction was set because the experimental analysis is very time-consuming. As a consequence, all that was proven here is that the model can mimic the mechanical laxity behavior of a knee joint, if the relevant mechanical properties of the ligaments and the articular geometry in that knee joint are used as input for the model. In addition, this would also work if the same experiment were repeated with another knee joint. It was not shown here, however, that the model can now represent an arbitrary knee joint with the same numerical values for the relevant properties.

Another restriction of the study is the number of load cases applied. In addition to the anterior–posterior and varus–valgus loads applied, we tried to apply endo- and exorotation moments. In these loading situations both the experimental and numerical knee-joints appeared to be mechanically unstable. In the experiment, luxation occurred. In the model simulations, no equilibrium position of the joint could be found. The joint structures represented appeared to be inadequate to restrain the externally applied axial moments.

The laxity characteristics of the knee model depend on the descriptions of the ligaments and the articular surfaces. Assuming that the articular surfaces are well represented (Blankevoort et al., 1991), the ACL is mainly responsible for the translations of the tibia in the anterior direction, the PCL for the translations of the tibia in the posterior direction, the MCL for the valgus rotation and the LCL for the varus rotation (Gollehon et al., 1987; Grood et al., 1981; Markolf et al., 1976; Mommersteeg et al., in press; Fizial et al., 1980; Seering et al., 1980). It was shown that the laxity of a knee specimen can be described reasonably well with our model. Only when a varus moment of 8 Nm was applied and when the tibia was posteriorly loaded did the model predict slightly higher laxities than those measured. Because of this, it might be that the representations of the LCL and the PCL were not precise or incomplete. For the PCL and the LCL, the bony interactions with the tibia and femur, respectively, were not taken into account. This bony edge elongates the fiber bundles of these ligaments in the anatomical specimen without any changes in the distances between their insertion sites. As a result, the forces in these structures may have been higher in reality than in the model, thus creating a higher resistance against displacement or rotation of the knee.

It can be concluded from the model-experiment comparisons that the model is suitable to describe an experiment realistically. The collections of line elements appeared to be a suitable representation of the ligaments in studying the laxity characteristics of a human knee. Furthermore, it seems that mechanical parameters, derived from tensile testing in nonanatomical ligament orientations can be extrapolated to in situ orientations.

REFERENCES


