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TECHNICAL NOTE

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.*, 1976). 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 (Blankevoort, 1991; Kok, 1993; Lewis *et al.*, 1989; Mommersteeg *et al.*, 1996; Woo *et al.*, 1991).

In previous studies, we proposed an alternative concept of the ligaments to evaluate their functions. They were considered, conceptually, as collections of several tensile elements with variable orientations, lengths and mechanical properties, rather than as one-dimensional structures (Mommersteeg *et al.*, 1996). This structural concept was proposed in order to explain the functional role of the ligaments in knee-joint motion from the loaded and unloaded tensile elements (Mommersteeg *et al.*, in press). The aim of the present study was to verify this new structural concept of human knee ligaments.

In earlier work (Mommersteeg *et al.*, 1995b), the geometry of the ligaments was described by the three-dimensional insertion sites of different tensile elements which were determined by anatomical analyses of the fiber orientations in the ligaments. The forces in these tensile elements during knee motions were determined from the relative positions of the ligament insertion sites, specifying the lengths of the elements, and their mechanical parameters (Mommersteeg *et al.*, in press). These parameters, defining the force–length relationships of the tensile elements, were identified using a combination of mathematical analysis and experimental assessment of ligament behavior in laboratory bench tests (Mommersteeg *et al.*, 1996).

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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 was assumed 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 function 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 of one knee-joint specimen were compared to model predictions, using a three-dimensional computer simulation knee 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 (Blankevoort, 1991), used to simulate the experiment (i).

(i) *Experiment: knee-joint motions.* One cadaveric knee joint, donor aged 71, freshly frozen in plastic bags at $-20°C$, was slowly thawed at room temperature. Although this procedure does not markedly affect the mechanical properties of fibrous tissues *in vitro* (Woo *et al.*, 1986), they are probably different from the *in vivo* state (Matthews and Ellis, 1968). The fibula was fixed to the tibia with a Kirchner wire. All periarticular connective soft tissues were removed, so that only the ligaments remained to connect the femur to the tibia/fibula complex. No signs of knee pathology were present. The femur and the tibia were cemented at their ends in polymethylmethacrylate (PMMA). The knee specimen was positioned in a specially designed motion and loading rig (Blankevoort *et al.*, 1988) and surrounded by a plastic bag to maintain a relatively high humidity. Before testing, the knee ligaments were preconditioned by cyclically loading the tibia in several positions of the knee. The flexion of the femur was constrained to flexion angles of successively 0°, 15°, 30°, 60° and 90°. At each of these flexion angles, an axial force of 150 N was combined successively with anterior and posterior forces of 50 and 100 N. At flexion angles of 0°, 15° and 30° of flexion, the axial force of 150 N was combined successively with varus and valgus moments of 8 and 16 Nm. For each flexion angle, axial rotations of the tibia were constrained during anterior–posterior loading as well as during varus–valgus loading. The relative positions of the bones were determined with Röntgen Stereophotogrammetric Analysis (RSA, Selvik, 1974; Blankevoort *et al.*, 1988). For this purpose, six tantalum pellets (0.8 mm diameter) were inserted in each bone before mounting the knee in the motion rig. The three-dimensional positions of these pellets were measured with an accuracy of 50 μm (de Lange *et al.*, 1985).

(ii) *Geometry of the articular surfaces of femur and tibia.* After the experiment, the three-dimensional geometry of the articular surfaces was measured relative to the 0.8 mm pellets in the femur and the tibia by a stereophotogrammetric method with an accuracy of 96 μm (Meijer *et al.*, 1989).

(iii) *Ligament forces as functions of their relative insertion orientations (Mommersteeg* et al., *1995a).* The insertion sites of the four major ligaments were surrounded by 0.5 mm tantalum pellets. The positions of the 0.5 mm pellets were related to the 0.8 mm pellets in the femur and the tibia by applying RSA. Subsequently, each ligament was isolated as a bone–ligament–bone preparation. For each preparation a series of tensile tests was performed. The relative orientations of the femoral and tibial insertion sites were varied systematically, though not *per se* as in the *in situ* situation. The ligament forces were measured with a load-cell. The relative insertion positions were determined using RSA. In this way, relationships between the relative orientations of the bones and the mechanical behavior of the ligaments were determined.

(iv) *Bundle insertion site geometry (Mommersteeg* et al., *1995b).* Obstructing bone parts and synovium were removed from the bone–ligament–bone preparations, to obtain a clear view of the fiber orientations of the ligaments from all sides. These preparations were fixed at both ends in a plastic support. The relative positions of the ligament insertions were determined by applying RSA for the 0.5-mm pellets. The contour-lines of the ligament insertion sites were measured with a 6-dof digitizer as well with an accuracy of 0.35 mm (3Space Isotrak, Polhemus Navigation Sciences, Colchester, VT, U.S.A.; Sidles *et al.*, 1988). Because the measuring principle of this system is based on electromagnetic signals, metals and instruments which affect these signals were avoided between the transmitter and the receiver. Subsequently, ligament bundles were identified using the fiber orientations as a guide. In the ACL, PCL, MCL and LCL, 7, 6, 3 and 3 bundles were identified, respectively (Fig. 1). These bundles were separated by a blunt edge, and removed one by one. Subsequently, the contour-lines of the insertion sites of these bundles were measured with the 6-dof digitizer. The insertion sites of each bundle were assumed to be concentrated at two points, the geometric centers of the perimeters.

(v) *Force–length relationships of the bundles (Mommersteeg* et al., *1996).* The force–length relationships of the fiber bundles defined were determined by combining the series of tensile tests (iii) with separate multi-bundle models of the ligaments (Mommersteeg *et al.*, 1996). In these models the bundles were represented by tensile elements. The three-dimensional coordinates of these elements' insertion sites were measured as described in (iv). The model calculates the ligament forces as a function of the relative positions of the femoral and tibial insertion sites and two unknown model parameters for each line element. These unknown parameters, defining the force–length relationships of the line elements, are a stiffness parameter k and the length at which the line element takes up force, the zero-force length L_0. These parameters were determined by simulating the tensile tests with the model, and subsequently, minimizing the differences in ligament forces, measured in the experiment and calculated with the model for all relative positions of the bones by adjusting the two parameters. This procedure resulted in values for k and L_0 for all line elements, and thus in their force–length relationships.

(vi) *The three-dimensional whole-joint model.* The geometric data of the specimen (ii), (iv) and the parameters characterizing the mechanical behavior of the ligaments (v) were used as input for a mathematical knee model. A graphic representation of this model is shown in Fig. 1. The model describes the position of the femur relative to the tibia for a given configuration of external

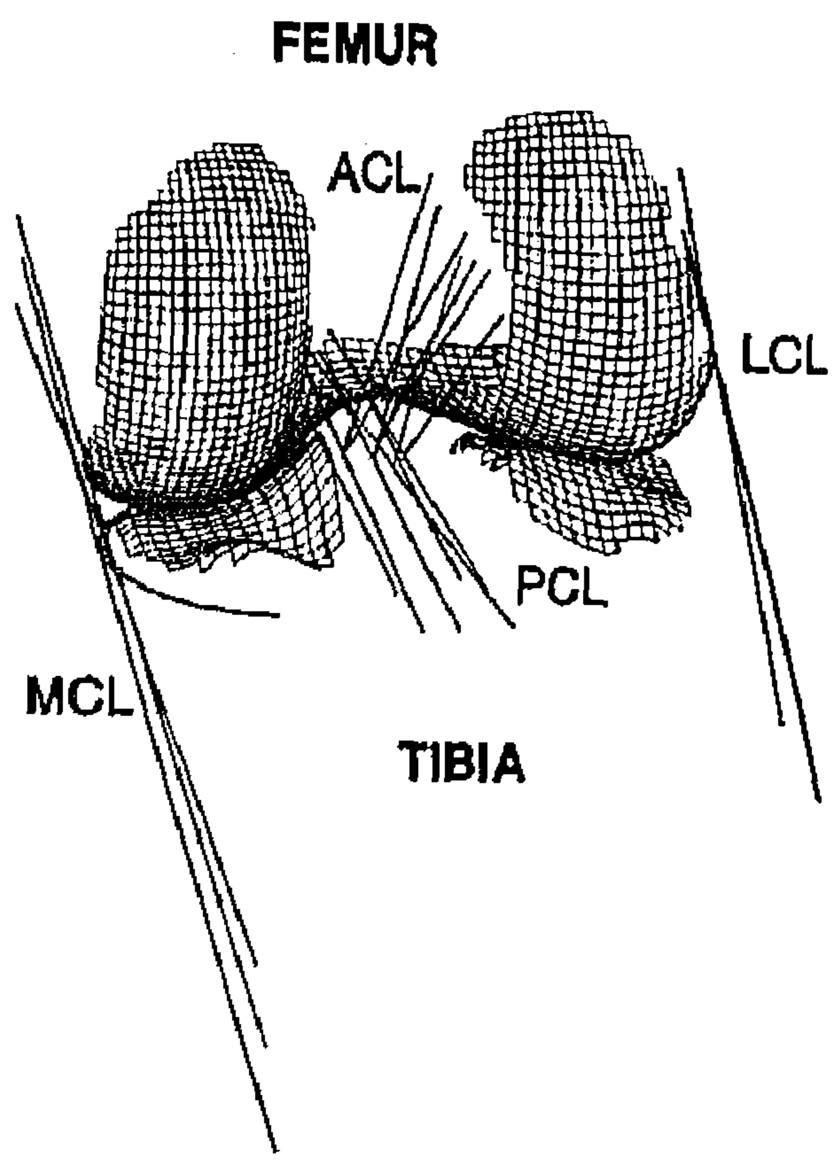


Fig. 1. Graphic representation of the three-dimensional knee-joint model in a posterior–medial view, including the femoral condyles, the tibial surfaces and the line elements which model the anterior cruciate ligament (ACL), the posterior cruciate ligament (PCL), the medial collateral ligament (MCL) and the lateral collateral ligament (LCL). The medial edge is represented by a spatial circle segment. The joint position shown is extension.

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 ($\mathbf{F}_l$) and moments ($\mathbf{M}_l$) were summations of the forces ($\mathbf{F}_j$) and moments ($\mathbf{M}_j$) in its tensile elements j, according to

$$\mathbf{F}_l = \sum \mathbf{F}_j, \qquad \mathbf{M}_l = \sum \mathbf{M}_j. \tag{1}$$

The bundle force $\mathbf{F}_j$ and moment $\mathbf{M}_j$ acting on the tibia are expressed by

$$\mathbf{F}_j = F_j \mathbf{v}_j, \qquad \mathbf{M}_j = \mathbf{D} \times \mathbf{F}_j, \tag{2}$$

where $\mathbf{v}_j$ is the unit vector pointing from the tibial to the femoral insertion site of line element j and $\mathbf{D}$ 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 ε_j of this line element as

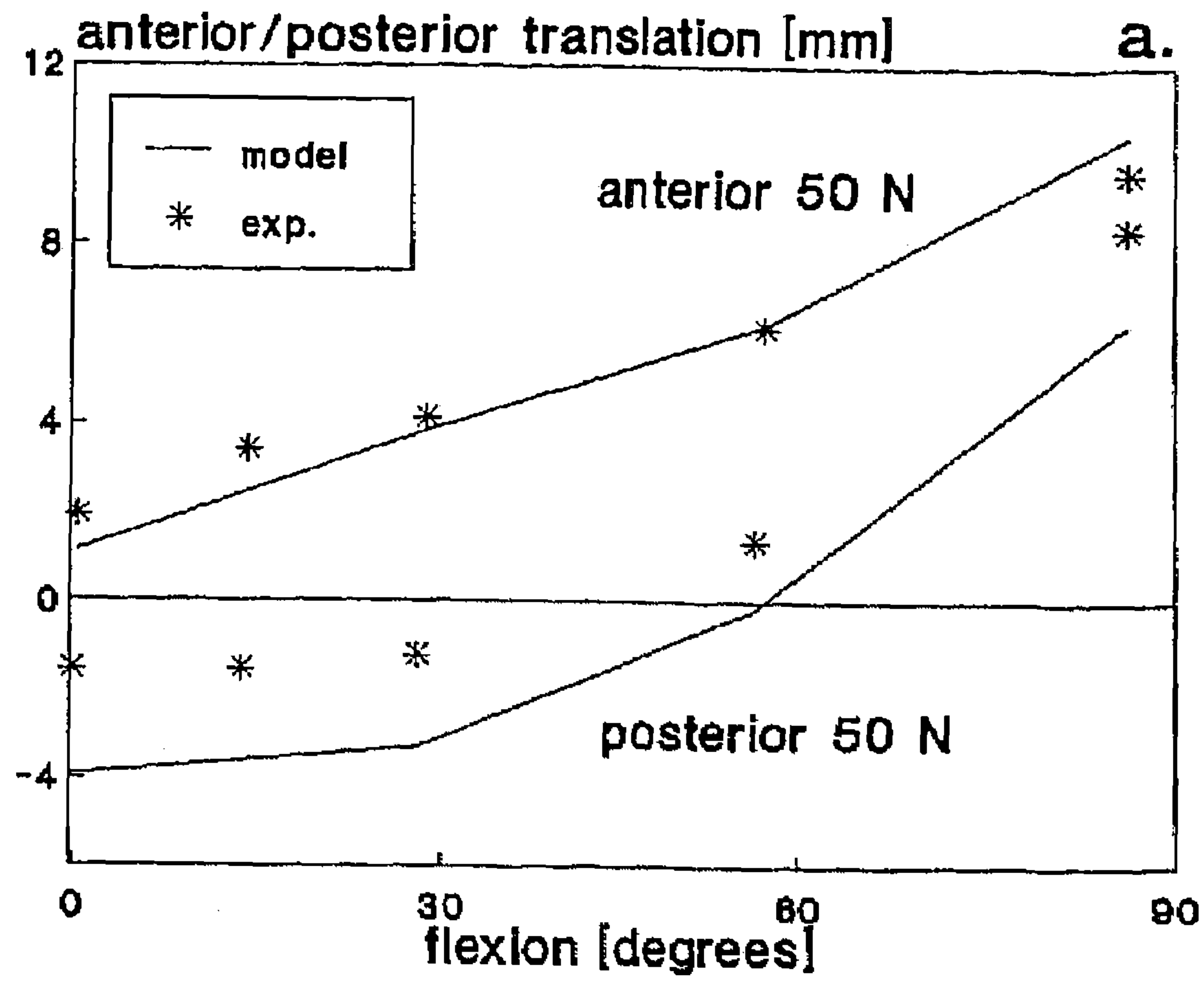
$$F_j = k_j \varepsilon_j^2, \quad \varepsilon_j > 0, \qquad F_j = 0, \quad \varepsilon_j \leqslant 0, \tag{3}$$

in which k_j[N] is a parameter which defines the structural stiffness $2k\varepsilon$ at each strain level and ε_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

$$\varepsilon_j = (L_j - L_{0j})/L_{0j}. \tag{4}$$

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.



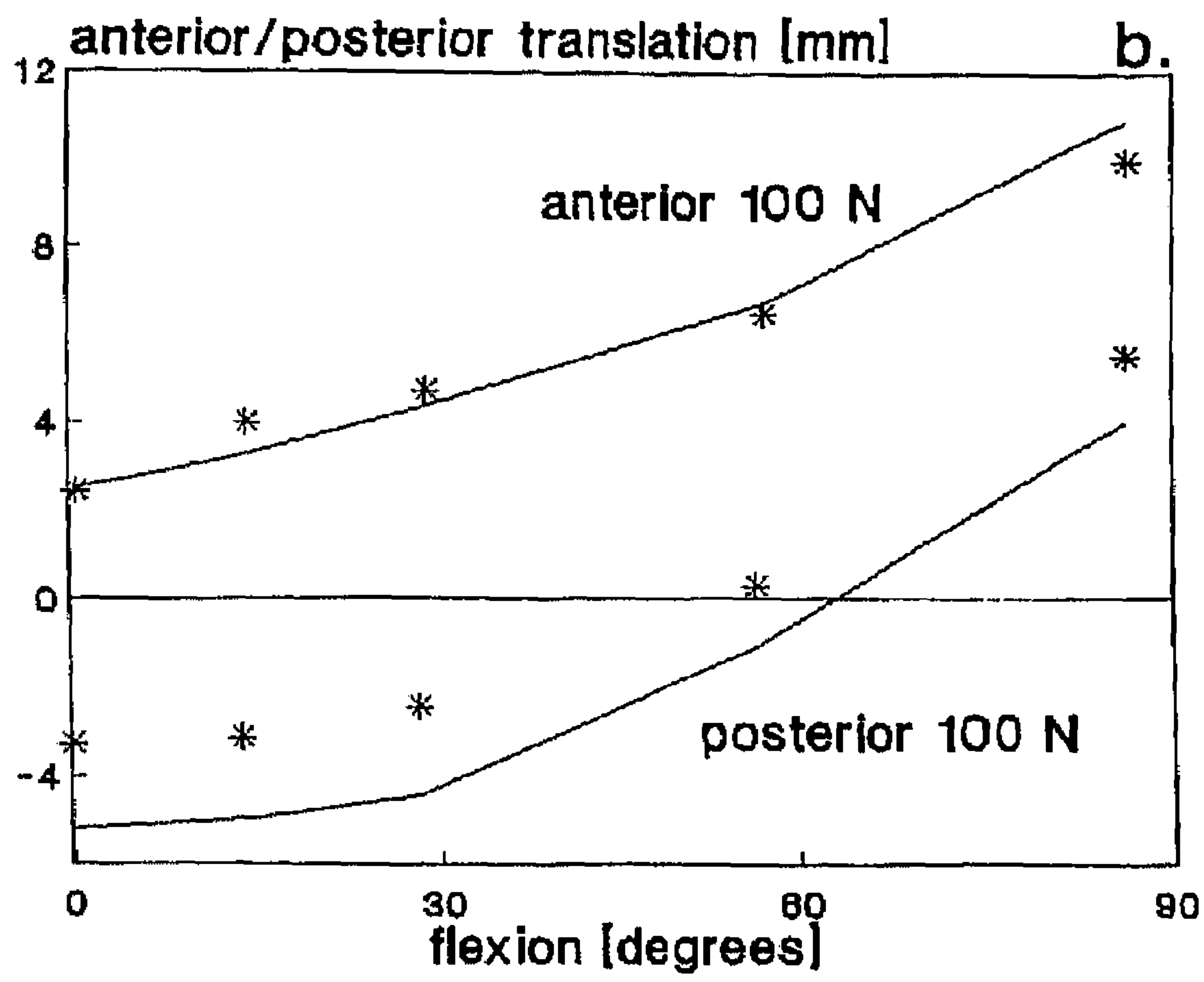


Fig. 2. The anterior–posterior translations as a function of flexion, while applying anterior and posterior forces of 50 N (a) and 100 N (b) to the tibia for the model simulations (solid lines) and the experimental specimen (data points). Laxity values were predicted discretely by the model at the same flexion angles as in the experiment. Between these points intrapolations were performed. The anterior translations were very well predicted by the model, while the translations with posterior forces were ± 2 mm higher than measured experimentally.

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 N m, 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 N m, the model predictions of the valgus rotations were higher than the experimental results in extension of the knee and 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)].

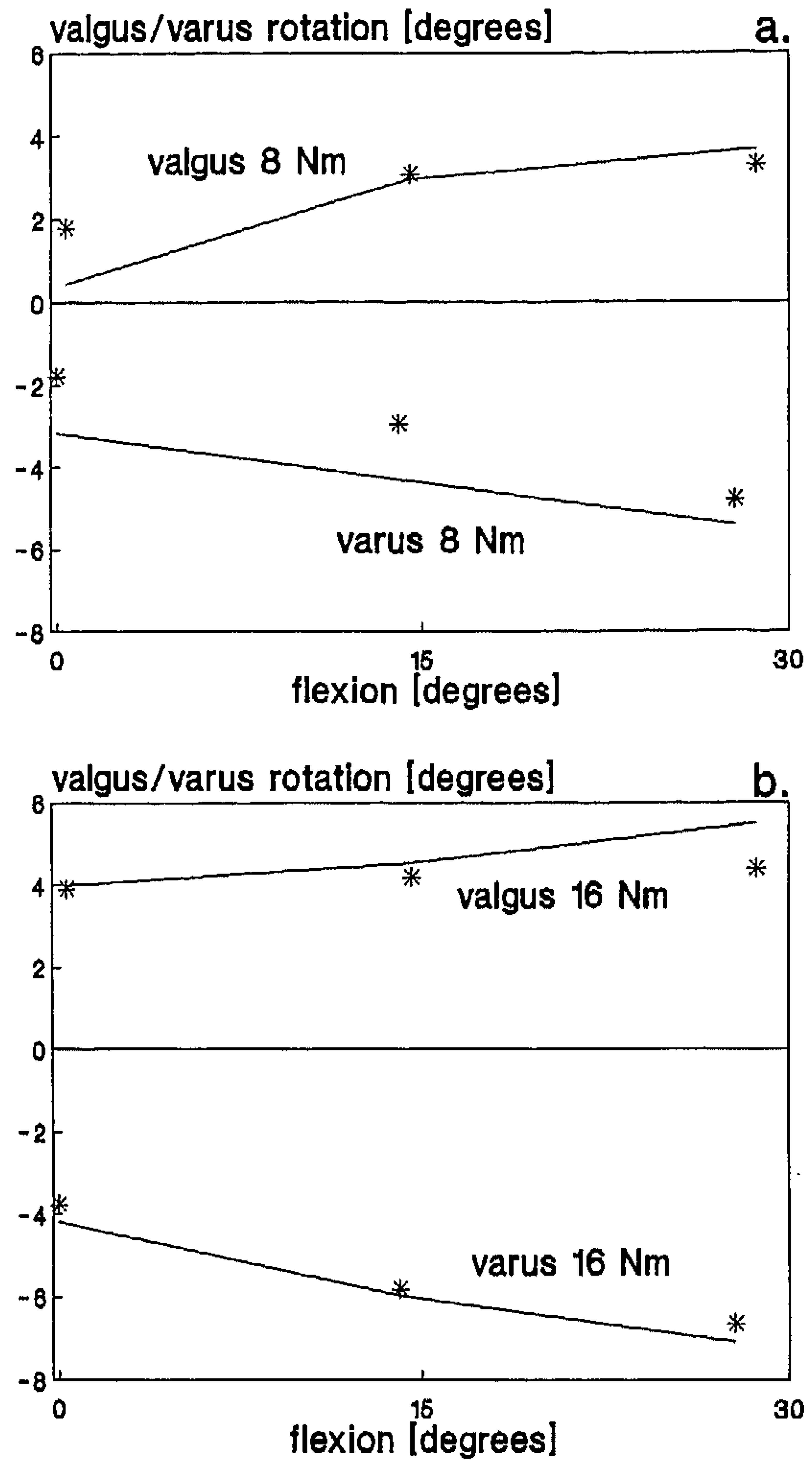


Fig. 3. The varus–valgus rotations as a function of flexion, while applying varus and valgus moments of 8 N m (a) and 16 N m (b) to the tibia for the model simulations (solid lines) and the experimental specimen (data-points). Laxity values were predicted discretely by the model at the same flexion angles as in the experiment. Between these points intrapolations 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.

Table 1. Total anterior–posterior laxity [mm]

Load	Flexion	Total anterior–posterior laxity [mm] Experiment	Model
50 N	0	3.5	5.1
	15	4.9	6.1
	30	5.4	7.1
	60	4.7	6.4
	90	1.3	4.2
100 N	0	5.7	7.7
	15	7.1	8.3
	30	7.2	8.8
	60	6.2	7.8
	90	4.4	6.9

Note. Variations in total anterior–posterior laxity during knee flexion are equal for the model simulation and the experiment. However, the model predicted slightly higher laxities than measured.

Table 2. Total varus–valgus laxity [degrees]

Load	Flexion	Total valgus–varus laxity [degrees] Experiment	Model
8 N m	0	3.6	3.6
	15	6.0	7.2
	30	8.0	9.1
17 N m	0	7.7	8.2
	15	10.0	10.5
	30	11.0	12.6

Note. Variations in total varus–valgus laxity during knee flexion are equal for the model simulation and the experiment. However, the model predicted slightly higher laxities than measured.

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; Piziali *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 N m 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.

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