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AN INVERSE DYNAMICS MODELING APPROACH TO DETERMINE THE RESTRAINING FUNCTION OF HUMAN KNEE LIGAMENT BUNDLES

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Abstract— During knee motion, the fiber bundles of ligaments are nonuniformly loaded in a recruitment pattern which is different for successive knee-joint positions. As a result, the restraining functions of these ligaments are variable. To analyze the relative restraint contributions of the fiber bundles in different knee-joint positions, a new method was developed. Its application was illustrated for the cruciate ligaments of one knee-joint specimen.

The methods developed to estimate bundle forces comprise five steps. First, the three-dimensional motions of a knee specimen are measured for anterior–posterior forces, using Röntgen Stereophotogrammetric Analysis. Second, bone–ligament–bone tensile tests are performed to evaluate the mechanical properties of these structures in several relative orientations of the bones. Third, multiple fiber bundles are identified in each ligament, based on the main fiber orientations. Fourth, the nonlinear force–length relationship of each functional bundle, as defined by a stiffness and a recruitment parameter, is determined by combining the multidirectional tensile tests with a multi-line-element ligament model. Finally, the information obtained is combined in a whole-joint computer model of the knee, to determine the internal forces in the initial kinematic experiment, using an inverse dynamics approach.

The technique appeared to be extremely time consuming and technologically involved. However, it was demonstrated to be useful and effective. The preliminary results reveal that the fiber bundle restraints are extremely sensitive to the knee flexion angle and the restraining forces are highly variable within the ligaments. For both cruciate ligaments, a gradual transition was demonstrated in load transfer from the posterior bundles to the more anteriorly positioned ones during knee flexion. Furthermore, it appeared that relatively high forces were carried by only a few fiber bundles at each flexion angle. Based on these preliminary results, it is concluded that the determination of forces in multiple ligament bundles is important for the understanding of failure mechanisms of ligaments. In particular, alternate loading of different fiber bundles suggests that successful operative reconstruction of the cruciate ligaments may not be achieved simply by a one-bundle preparation. Copyright © 1996 Published by Elsevier Science Ltd.

Keywords: Knee model; Knee ligaments; Ligament bundles; Function.

INTRODUCTION

Human knee ligaments are geometrically complex and mechanically nonhomogeneous tissues. As an effect of these characteristics, the loading patterns of these structures are typically three-dimensional in nature: they transfer loads nonuniformly, in several directions, displaying variable load–displacement characteristics during knee motions. In this way, in perfect tuning with other knee structures, they are able to provide the knee with sufficient stability and movability. Understanding these complex mechanisms underlying the function of knee ligaments is important for the evaluation of partial ligament ruptures as well as for the design of ligament replacements.

The role the ligaments play in knee mechanics has been studied by the measurement of ligament forces, using a variety of methods. Direct measurements of forces in the four major ligaments of the knee were performed *in situ* using buckle transducers (Lewis *et al.*, 1982, 1989; Ahmed *et al.*, 1987, 1992). Selected bundles of knee ligaments were instrumented and the loads in two

flexion angles of the joint measured. To obtain whole ligament forces, Markolf *et al.* (1991, 1993) and Wascher *et al.* (1993) connected a load-cell to the tibial insertion sites of the anterior and posterior cruciate ligaments of the knee. These studies have provided considerable information concerning the *in situ* forces developed within the cruciate ligaments. In order to determine the restraining functions of the ligaments, however, the directions of the forces relative to the externally applied forces have to be taken into account. In recent studies, the functions of the anterior cruciate ligament and its major bundles in restraining an anterior tibial force of 100 N were determined by selective cutting (Blomstrom *et al.*, 1993; Guan *et al.*, 1991). In these studies the load-transmitting role of the removed structure was determined. A disadvantage of this approach is that the removal of (part of) a structure could influence the loading patterns of the remaining knee structures as an indirect effect.

An alternative to direct measuring techniques is the indirect assessment of ligament forces from inverse dynamics models (Huiskes, 1992). If the positions of the ligament insertion sites are known relative to the geometry of the bones, and the force–displacement characteristics of the ligament as well, the ligament force can be calculated from prescribed motions of the bones relative to each other. This method was applied frequently in the

past (Blankevoort *et al.*, 1991; Crowninshield *et al.*, 1976; Edwards *et al.*, 1970; Grood and Hefzy, 1982). These studies were limited in scope, because usually the ligaments were arbitrarily represented by a limited number of line elements, disregarding their three-dimensional multibundle structures. In addition, force-displacement characteristics of the ligaments were usually estimated from literature values. Finally, the difficult problem of the zero-force length, required for these calculations, was not solved. The method was improved by Hollis *et al.* (1988) and Takai *et al.* (1993), who determined the relation between force and length for the entire ACL as well as for portions of it by the application of anterior tibial forces in different flexion angles of the knee (Takai *et al.*, 1993) or by bone-bundle-bone testing (Hollis *et al.*, 1988).

In order to assess the resultant force vectors and force distributions in all human knee ligaments simultaneously, a new method is presented here. This method comprises the inverse dynamics modeling approach to its full potential. In order to illustrate the utility of the new method, it was applied for the cruciate ligaments of one knee-joint specimen. The resulting loading patterns of these structures during anterior-posterior loading of the tibia at different flexion angles of the knee joint are presented and compared with literature data.

METHODS AND MATERIAL

The protocol (Fig. 1) consists of the determination of (i) the joint motions associated with externally applied anterior-posterior forces, (ii) the relationships between ligament forces and the relative positions of the bones, (iii) the insertion site geometry of bundles identified in the ligaments, (iv) the bundle force-length relationships by combining (ii) and (iii) in a separate line-element model for each ligament and, finally, (v) the forces in the bundles by combining (i), (iii) and (iv) in a whole-joint model of the knee. The protocol was applied for one cadaveric knee joint, aged 71, which was freshly frozen in a plastic bag at -20°C , and slowly thawed at room temperature at the time of usage. Any signs of knee pathology were absent.

Fig. 1. Schematic diagram representing the inverse dynamics modeling approach.

To record the kinematics of the knee joint for anterior-posterior loading of the tibia (i), the femur as well as the tibia were transected at a distance of about 15 cm from the joint line and cemented at their ends in polymethylmethacrylate (PMMA). The fibula was fixed to the tibia with a Kirchner wire. The knee specimen was positioned in a specially designed motion and loading apparatus (Blankevoort *et al.*, 1988). Before testing, the knee ligaments were manually preconditioned by cyclically moving the tibia through endo/exorotation, varus/valgus and anterior/posterior pathways at flexion angles of 0, 15, 30, 60 and 90° . Subsequently, the femur was fixed in flexion angles of 0, 15, 30, 60 and 90° . In the first flexion series a compressive force of 150 N was applied at each flexion angle. Tibial rotation was free. In subsequent flexion series, this compressive force was combined with anterior and posterior forces of 100 N. In these series, tibial rotation angle was fixed for each flexion angle at the same value as obtained by free rotation in the first flexion series. In each flexion and loading position the relative positions of the bones were determined with Röntgen Stereophotogrammetric Analysis (RSA, Blankevoort *et al.*, 1988; Selvik, 1974). Therefore, six tantalum pellets (0.8 mm diameter) were inserted in each bone before mounting the bones in the motion and loading apparatus.

To determine the relationship between ligament forces and the relative positions of the insertion sites (ii), 0.5 mm tantalum pellets were inserted near the insertion sites of the cruciate ligaments. 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 bone-ligament-bone preparation a series of tensile tests was performed in which the relative orientations of the femoral and tibial insertion site were varied such that all ligament fibres were tensed at least once. The orientations of the bones were not anatomical *per se*. The ligament forces were measured with a load-cell. The relative insertion positions were determined with RSA. In this way, relationships between the relative positions of the bones and the ligament forces were determined (Mommersteeg *et al.*, 1995b).

To determine the bundle insertion-site geometry (iii), 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 perspex support. The relative positions of the ligament insertions were determined by RSA. Ligament bundles were identified using the fiber orientations as a guide. In the ACL seven bundles were identified, a posterior, a posterolateral, a posteromedial, a central, an anterolateral, an anteromedial and an anterior bundle, while in the PCL six bundles were identified, a posterior or reinforcing bundle, a posterolateral, a posteromedial, an anterolateral, an anteromedial and an anterior bundle. The contour lines of the insertion sites of these bundles were measured with a 6-d.o.f. digitizer (3Space Isotrak, Polhemus Navigation Sciences, Colchester, VT, U.S.A.; Mommersteeg *et al.*, 1995c).

The force-length relationships of the fiber bundles (iv) were determined by combining the series of tensile

tests (ii) with a model for each ligament (Mommersteeg *et al.*, 1996). In this model the bundles were represented by line elements of which the three-dimensional coordinates of the insertion sites were the geometric centers of the insertion site perimeters as measured (iii). The ligament forces were described as functions 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, 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. The ligament forces were calculated in this model as follows.

The length patterns of the line elements are determined completely by the coordinates of their insertion sites, as determined in (iii), and the three-dimensional kinematics of the femoral bone relative to the tibial one represented by the translation vector $\mathbf{D}$ and the rotation matrix $\mathbf{R}$, as determined in (ii). The length L_j of a line element j is calculated from the tibial insertion point $\mathbf{X}_{tj}$ relative to the tibial coordinate system and the femoral insertion point $\mathbf{x}_{fj}$ relative to the femoral coordinate system, according to

$$L_j = |\mathbf{X}_{tj} - \mathbf{D} - \mathbf{R}\mathbf{x}_{fj}|. \quad (1)$$

The tensile behavior of each line element is assumed to be nonlinear elastic: 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 (Grood and Hefzy, 1982; Elden, 1968) as

$$\begin{aligned} F_j &= k_j \varepsilon_j^2, \quad \varepsilon_j > 0, \\ F_j &= 0, \quad \varepsilon_j \leq 0, \end{aligned} \quad (2)$$

in which k_j is a stiffness parameter and ε_j is the strain in line element j , which is calculated from its actual length L_j , as calculated in (1), and its zero force length L_{0j} according to

$$\varepsilon_j = (L_j - L_{0j})/L_{0j}. \quad (3)$$

The bundle force vector $\mathbf{F}_j$ acting on the tibia is expressed by

$$\mathbf{F}_j = F_j \mathbf{v}_j, \quad (4)$$

where $\mathbf{v}_j$ is the unit vector pointing from the tibial to the femoral insertion site of line element j . For the purpose of this model we assume no mutual interaction between the bundles. Under this condition, the total ligament force $\mathbf{F}_l$ in each relative position of the insertion sites is found by summing the individual bundle forces $\mathbf{F}_j$ according to

$$\mathbf{F}_l = \sum \mathbf{F}_j. \quad (5)$$

The differences between the ligament forces measured and the corresponding values predicted by the model were minimized by adjusting the model parameters k_j and L_{0j} , using a modification of the Levenberg-Marquardt algorithm, applying the general least-squares solver LMDIF (from MINPACK; Moré *et al.*, 1980). This optimization process resulted in the identification of

values for k_j and L_{0j} of each line element, and thus in their force-length relationships.

To determine the bundle forces *in situ* (v), the ligament multiple-bundle representations as defined in (iv) were implemented in a whole knee-joint model (Blankevoort, 1991). The three-dimensional motion characteristics of the knee-joint determined in (i) were used as input for the model. The lengths of the line elements were calculated from equation (1) for different positions of the knee. From these lengths and the parameters k and L_0 (iv), the forces in the line elements were calculated from equations (2)–(5). The restraining function of a ligament (bundle) was defined as the difference in the posterior–anterior component of the bundle force between the load cases with an axial force only and those with an additional anterior–posterior force, expressed as a percentage of the externally applied load.

RESULTS

When a compressive force of 150 N is applied to the knee joint, the anterior cruciate ligament force decreases from about 130 N in extension of the joint to 0 N in 90° of flexion (Fig. 2(a)). RSA revealed that the compressive force induced valgus and internal rotation of the tibia, combined with medial and anterior displacements. When an anteriorly directed force of 100 N was added, the ACL forces increased for all flexion angles with about 92 N maximally at 15° of flexion. The percentage of the applied force restrained by the ACL varied from 70% at 15° of flexion to 30% at 90°. This force became increasingly directed anterior–posteriorly while the knee was flexed (Fig. 3(a)). The loads in the fiber bundles varied during knee flexion and shifted along the medial side of the ligament from posterior to anterior (Fig. 3(a)). Almost uniform loading of the fiber bundles occurred between 15 and 30° of flexion. At 0° of flexion, 2–3% of the ACL restraint was provided by the most anterior fibers of the ligament, while at 90° of flexion this percentage was 100% (Fig. 4(a)). For the most posterior bundle the reverse was true: at 0° of flexion the contribution to the total ACL restraint was 67%, while at 90° of flexion it was zero. The lateral fiber bundles were only slightly tensed at 15 and 30° of flexion.

Compressive loading of the joint resulted in minimal tension in the PCL in 0° of flexion (Fig. 2(b)). When a posterior force of 100 N was added, the PCL force increased for all flexion angles. The resultant PCL force difference was the lowest at 30° (13 N) and increased towards extension (86 N) as well as towards higher flexion angles (39 N). During knee flexion, the PCL force became directed increasingly proximodistally (Fig. 3(b)). The percentage of the applied force restrained directly by the PCL was maximal in extension (43%) and minimal at 30° of flexion (7%) (Fig. 2(b)). This restraint was provided completely by the posterior and posterolateral bundles (Fig. 4(b)). At higher flexion angles, the more anteriorly positioned bundles became the primary restraints, in particular the anteromedial bundles: at 60° of flexion, the contributions of the more anteriorly vs the more posteriorly positioned bundles to the total PCL restraint were 48 and 52%, respectively, while at 90° of flexion, they were 92 and 8%, respectively.

Fig. 2. Resultant forces (first column of stacked bars) as well as the anterior-posterior components of these forces (second column of stacked bars) of the ACL (a) and the PCL (b) at 0, 15, 30, 60 and 90° of flexion with an externally applied compressive force of 150 N (black and grey) in combination with anterior and posterior forces of 100 N, respectively (cross-hatched).

DISCUSSION

In this study it was shown that with an inverse dynamics approach it is feasible to obtain detailed three-dimensional load-distributions of knee ligaments simultaneously. The preliminary results for one knee-joint specimen confirmed the complexity of load transfer through the cruciate ligaments, whereby each part of a ligament plays a characteristic role in particular phases of knee motion. A gradual transition of forces is found from the posterior fiber bundles to the anterior ones. Hence, the contributions of the ligaments to knee restraint depend strongly on its flexion angle.

The methods used have some unique features. The forces in several bundles of more than one ligament can be determined simultaneously. Magnitudes, directions and points of application of these forces can be determined. The methods imply determination of the *in situ* ligament bundle lengths from kinematic data. The parametric characterization of ligament behavior in whole ligament tests, for different relative insertion orientations, provides the load-length curves of the bundles. *In situ* loads are obtained by simply comparing the bundle lengths with their load-length curves, in each relevant situation. The way in which the parameters describing the force-length relationships of the ligament bundles are identified tackles two problems common in ligament testing. First, ligament properties have been shown to depend on the alignment of the ligament in the testing machine (Woo *et al.*, 1991 and others). With the present method, the test orientations are varied and the variable ligament behavior is accounted for in the analysis. Sec-

ond, it offers a solution for the problem of the zero-load length determination of the ligament (bundles).

Another feature characteristic for the new method is that the multiple-line representations of the ligaments are based on actual measurements of the anatomy. In the past, ligaments were modeled by two or three arbitrary line elements, based on measurements of the ligament insertion sites' perimeters (Blankevoort, 1991; Van Dijk, 1983; Takai *et al.*, 1993). Blankevoort (1991) showed that two or three line elements do not suffice to stabilize the knee during its entire range of motion. Hence, the present method is expected to produce more precise and anatomically consistent information.

In comparison with the direct measuring approach, the method described here has the advantage that no external devices must be attached to the ligament which might cause impingement problems or alterations in the load-elongation behavior of the ligament. Furthermore, the joint remains intact during the kinematic experiment, preserving the anatomic relationships within the joint. In dissection studies, the removal of a supporting structure might influence the internal configuration of remaining knee-joint structures and, thus, their role in knee-joint mechanics. The ligaments provide direct restraints to knee-joint motion, but also indirect articular restraints caused by interaction between ligaments and the articular surfaces (Huiskes and Blankevoort, 1991). In addition there are interactions between a ligament and other knee structures, such as the menisci, the capsule and other ligaments which provide also indirect restraints. Using the present method, the direct individual ligament restraint is determined in isolation, while in dissection

Fig. 3. Medial views of the line element representations of the ACL (a) and the PCL (b) *in situ* at 0, 15, 30, 60 and 90° of flexion with an anterior and posterior force of 100 N, respectively. A loaded line element is presented by a continuous line and a nonloaded line element by a dotted line. The femoral insertion sites of these line elements are presented by square markers and the tibial insertion sites by round markers. Legends: Fiber bundles: 1. posterior, 2. posterolateral, 3. posteromedial, 4. central, 5. anterolateral, 6. anteromedial, 7. anterior.

studies the indirect effects are also included. The method is not limited with respect to the complexity of the knee loading configuration and the degrees of freedom of the test as is the case in dissection studies in which the maintenance of the path of motion before and after dissection of a structure is a prerequisite (Blomstrom *et al.*, 1993; Takai *et al.*, 1993; Vahey and Draganich, 1991).

However, our approach carries practical limitations. The experimental technique applied is extremely time consuming and technologically involved. Secondly, the ligaments are represented by a number of line elements which do not interact with each other. It was assumed that the effects of mechanical bundle-matrix and inter-bundle interactions are negligible. For the ACL it was demonstrated that longitudinal separation of the

anterior and posterior portions of the ACL influences the stiffness of the knee in anterior tibial loading only minimally (Blomstrom *et al.*, 1993; Takai *et al.*, 1993). However, it could be a factor which affects the accuracy of our results. Furthermore, it is assumed that the multi-line-element models can extrapolate information to other ligament orientations which occur *in situ*, but were not represented in the bench tests. Implicitly, it is assumed that the bundles are small enough to assume their force-length relationships independent of the relative orientations of the bones. Some of the uncertainties inherent to these limitations, however, have been removed by a global validation study of the ligament models, recently performed in our laboratory (Mommersteeg *et al.*, 1995a). Of course, these limitations are not inherent to our approach in general, because the multi line element

Fig. 4. The contributions of the ligament bundles to the ACL-restraint to anterior loading (a) and to the PCL-restraint to posterior loading (b) at flexion angles of 0, 15, 30, 60 and 90° of flexion. The black bars present fiber bundles at the posterior side of the ligaments. Toward the anterior side of the ligaments, the bars turn gradually into white bars.

models could also be replaced by more precise three-dimensional solid models in the future.

Another limitation not inherent to the method *per se*, but to the way in which it was applied here is related to the relevance of the results to the *in vivo* situation. To represent *in vivo* conditions, realistic loads must be represented in the knee tests. In our experimental design, this is possible only to a certain extent. In particular, the compressive knee-joint forces we applied were limited to about 150 N. The results are strongly dependent on the compressive force applied. The compressive force influences the distribution of restraints between the articular contact areas and the ligaments. With a compressive force, the contribution of the contact becomes higher (Hsieh and Walker, 1976). Under 925 N of tibio-femoral contact force, the resultant ACL force was decreased as much as 36% at 0° of flexion and 46% at 20° of flexion (Markolf *et al.* 1991).

There is general consensus in the literature that the ACL and the PCL are the primary structures resisting anterior and posterior tibial loading, respectively (Ahmed *et al.*, 1987; Blankevoort, 1991; Butler *et al.*, 1980;

Markolf *et al.*, 1991; Piziali *et al.*, 1980; Wascher *et al.*, 1993). This was once more confirmed in our approach. The actual force patterns we found in the PCL during posterior loading of the tibia, however, are not compatible with the results of Wascher *et al.* (1993). At a 50 N posterior tibial force, the mean PCL force was minimum at extension and increased gradually towards 58.4 N at 90° of flexion. The force patterns are, however, similar to those of Race and Amis (1993), who showed that the PCL force peaked twice, at extension and at 70° of flexion, for 2, 4 and 6 mm posterior tibial translations. Vahey and Draganich (1991) also found an increase in PCL force from 30 to 90° of flexion.

At a compressive force of 150 N only, the ACL forces decrease from 130 N in extension to 0 N at 90° of flexion. Markolf *et al.* (1991) measured considerable variation in the magnitudes of ACL force at passive knee extension. At 5° hyperextension, forces varied from 50 to 241 N and at 0° of flexion from 16 to 87 N. From 20° of flexion ACL forces were zero. When a 200 N quadriceps tendon pull was applied, resulting in compressive and anteriorly directed forces in the joint, the ACL forces became similar

to those found in the present study. With anterior tibial loading, the increases in ACL force were lower than the applied anterior tibial load for all flexion angles in the present study, with a maximal increase at 15° of flexion (92 N). This corresponds to the results of Vahey and Draganich (1991) and Ahmed *et al.* (1987). Takai *et al.* (1993) and Markolf *et al.* (1991), however, found that the resultant forces in the ACL were slightly higher than the anteriorly directed forces applied externally. Our finding that the loads in the ACL are more uniform at lower than at greater flexion angles is in agreement with Takai *et al.* (1993), who found a contribution of 70% of the anterior part of the ligament at 30° of flexion, which decreases towards extension (47%) and increases towards 90° flexion (93%). That the lateral bundle of the ACL does not play a significant role in resisting anterior tibial loading is in agreement with the findings of Hollis *et al.* (1988) and Blomstrom *et al.* (1993).

It appeared that relatively high forces during functional loading are carried by only a few fiber bundles. Therefore, the maximum stress that exists in a ligament cannot be determined by dividing the total force carried by the ligament by its cross-sectional area. Concentrations of the force in only a few fiber bundles of the ligament can result in (micro)failures at loads below those which would cause a ligament which is uniformly stressed to fail.

Data as obtained in the present study can be useful for the design of improved ligament reconstructions. The preliminary results suggest that the function of the ligaments, with these unique three-dimensional multibundle arrangements, may be difficult to reconstruct with a single-stranded structure. It may be more appropriate to reconstruct the ligaments with multiple strands, each with its own function at different flexion angles. There have been reports from studies indicating that the use of two-stranded reconstructions provided better results (Radford and Amis, 1990; Zaricznyj, 1987).

The information provided with an experimental approach as presented here is also useful for the evaluation of knee-ligament injuries, particularly for isolated lesions of fiber bundles. Two different tests are popular for the evaluation of anterior cruciate ligament lesions. In 90° of flexion the anterior drawer test is performed and in 15° of flexion the Lachman test. Our preliminary results suggest that during an anterior drawer test only the most anterior bundle is tested. An isolated lesion of this bundle could be detected in this way. With a Lachman test, the fiber bundles of the ACL are loaded more uniformly.

In conclusion, the approach presented here, involving inverse dynamics, whole knee-joint and ligament computer models, with associated experimental tests, is suitable to estimate detailed load distributions in the (knee-joint) ligaments simultaneously, in a variety of motion and loading conditions. Application to other joints is also possible. The information provided is unique and useful, albeit the method is elaborate and extremely time consuming. Evidently, where it concerns the present application to the cruciate ligaments, more knee specimens must be tested to verify the observed force distributions among the several fiber bundles during knee flexion.

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