Muscle-Ligament Interactions at the Human Knee(Ⅱ)
— Model Sensitivity Analysis —

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In part I of this paper, a three-dimensional model of the human knee joint was incorporated into a detailed human musculoskeletal lower extremity model. The model was used to determine the muscle, ligament, and articular contact forces transmitted at the knee as humans extend/flex in an isometric state. Part Ⅱ investigates the sensitivity of the model calculations to changes in the parameters which describe the mechanical behavior of cartilage and the ligaments of the knee. The ligament function in the real knee was most sensitive to changes in ligament reference lengths or strains, less sensitive to changes in cartilage stiffness, and least sensitive to changes in ligament stiffness.

Key Words: Sensitivity, Knee Model, Musculoskeletal Model, Knee Extension

1. Introduction

Model calculations of the human knee joint obtained in part Ⅰ of this paper depend greatly on the parameters used to describe the geometry of the bones and the geometrical and mechanical properties of the muscles, ligaments, and cartilage. The shapes of the femur, tibia, and patella in the model were based on geometric data obtained from 23 cadaveric knees. As such, the articulating surfaces of the model knee represent those of an average-sized joint. Differences in the sizes of the bones account in part for differences in the moment arms of the muscles spanning the joint. Bone geometry is therefore important, as it contributes in part to the variation in the torque developed by a muscle about a joint.

The calculations obtained from the model are also determined by the parameters which describe the geometrical and mechanical properties of the muscles, ligaments, and cartilage. For example, the net extensor torque developed about the knee depends not only on the shape of the bones, but also on the force-length properties and origin and insertion sites of the quadriceps muscles. The geometry of the bones and the origin and insertion sites of the muscles determine the variation of the knee-extensor moment arm with knee flexion angle. The force-length properties of the muscles, on the other hand, determine the variation in muscle force with changes in the configuration of the joint.

In this paper, a sensitivity study is undertaken to determine how changes in the parameters describing the mechanical properties of cartilage and the ligaments of the knee alter the model calculations reported in a companion paper (part Ⅰ by Kim, 1998). In particular, we examine the sensitivity of the calculated ligament and joint-reaction forces at the knee to changes in the mechanical properties of the model ligaments and cartilage.

2. Sensitivity to Changes in Cartilage Stiffness

Blankevoort et al. (1991) developed a three-dimensional model of the knee and used it to investigate the effects of cartilage stiffness on the relative displacements of the bones. These researchers increased the elastic modulus of cartilage in the model by a factor of 4, and calculated the resulting changes in the displacements of the tibia relative to the femur for passive flexion

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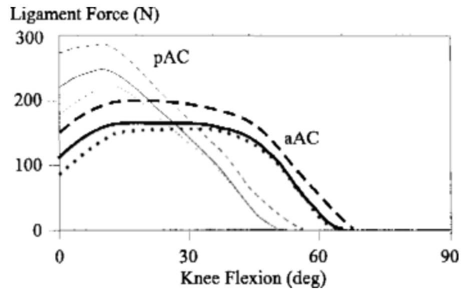
-extension movements of the knee. They also decreased the elastic modulus of cartilage by a factor of 2, and repeated the calculations. The nominal value of the elastic modulus of cartilage in Blankevoort's model was 5 MPa, which is the same as the value assumed in the present study. The results indicated that decreasing cartilage stiffness causes the ligaments to relax, which increases the laxity of the knee, particularly in the axial direction. Thus, total rotational laxity of the model increased as the stiffness of cartilage decreased. Unforunately, Blankevoort et al. (1991) did not show how changes in cartilage stiffness affect calculations of ligament forces during passive flexion-extension of the knee.

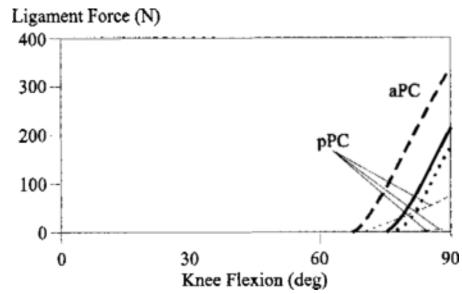
Loch et al. (1992) used a three-dimensional model of the knee to investigate how changes in cartilage stiffness affect the forces transmitted to the ligaments when an 89 N anterior drawer was applied to the tibia with the knee at full extension. Their calculations indicated that changes in cartilage stiffness can change the load sharing between the ACL and the other ligaments of the knee. In particular, they found that stiffening of the joint contact by increasing cartilage stiffness leads to an increase in ACL graft force.

To examine the effects of cartilage stiffness on model response, we changed the nominal value of the elastic modulus of cartilage in the model by as much as a factor of 20. Specifically, we increased the stiffness of cartilage by a factor of 20, and decreased it by a factor of 5. We then simulated maximum isometric knee extension in the model, and calculated the ligament and joint-contact forces for this task.

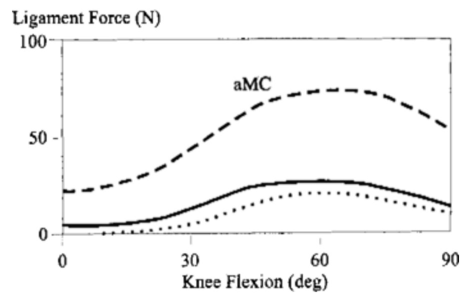
The net extensor torque developed by the model knee is relatively insensitive to changes in cartilage stiffness (not shown). The corresponding changes in the resultant force acting at the tibiofemoral joint are also small. There are considerable changes, however, in the calculated values of the ligament forces. In general, and in agreement with the results reported by Blankevoort et al. (1991), decreasing the stiffness of cartilage in the model leads to a decrease in the magnitude of the forces transmitted to the cruciate and collateral ligments of the knee(Fig. 1).



(a) Forces in the separate bundles of the model ACL (aAC, heavy lines; pAC, light lines).



(b) Forces in the separate bundles of the model PCL (aPC, heavy lines; pPC, light lines).



(c) Forces developed in the anterior bundle of the superficial model MCL (aMC).

Fig. 1 Knee-ligament forces calculated for maximum isometric extension of the model knee for the following three values of cartilage stiffness: 1MPa (dotted line), 5 MPa (solid line), and 100 MPa (dashed line). Increasing the stiffness of cartilage increases the cruciate and collateral ligament forces significantly.

Changes in cartilage compliance alter ligament forces, but they do not change the angle at which peak ligament force occurs. Independent of the value of cartilage stiffness used in the model, the peak force in the aAC occurs at about 15° of knee flexion, and that in the pAC occurs at around 10° (Fig. 1(a)). Interestingly, the change in the magnitude of ACL force does not scale proportionally with a change in the stiffness of cartilage as the angle of knee flexion changes. For example, at small angles of flexion, increasing cartilage stiffness by a factor of 5 (from 1 MPa to 5 MPa) produces an increase of around 30 N in the forces borne by the aAC and pAC. As the angle of knee flexion increases, however, increasing the stiffness of cartilage from 1 MPa to 5 MPa has almost no effect on the magnitude of ligament force.

The above trend is also evident when the stiffness of cartilage in the model is increased by a factor of 20, from 5 MPa to 100 MPa. At small angles of knee flexion, relatively large increases in ligament forces are produced by a twenty-fold increase in cartilage stiffness. At full extension, for example, the force in the aAC increases by about 40 N, while that in the pAC increases by as much as 70 N higher. As the angle of flexion increases, however, a twenty-fold increase in cartilage stiffness produces a much smaller (no more than 25 N) increase in aAC and pAC forces.

This trend is also reflected in the behavior of the model MCL (Fig. 1(c)). Increasing cartilage stiffness from 1 MPa to 5 MPa increased the force in the aMC almost uniformly over the entire range of knee flexion. A twenty-fold increase in cartilage stiffness, however, produced a nonuniform increase in MCL force as the angle of knee flexion was changed. These results suggest that changes in cartilage stiffness affect not only the magnitude of the forces transmitted to the cruciate and collateral ligaments, but also the distribution of force within the ligament. This is because the relationship between the change in ligament force and the change in cartilage stiffness is nonlinear for any given angle of knee flexion.

It is interesting to note that the tibiofemoral-joint force and knee-ligament forces obtained for a cartilage stiffness of 100 MPa are very similar to those reported by Shelburne (1996), who assumed rigid contact between the bones. The peak resultant force in the ACL calculated by Shelburne was around 500 N, whereas that given in Fig. 1(a) for a cartilage stiffness of 100 MPa is around 480 N (add dashed lines for aAC and pAC). Also, the peak resultant force predicted by Shelburne for the PCL was around 320 N at 90° of flexion, whereas Fig. 1(b) indicates a value of 400 N (add dashed lines for aPC and pPC). These results imply that a stiffness value of 100 MPa for cartilage approximates the condition of rigid contact between the femur and tibia.

The model calculations indicate that the articular contact forces at the knee are relatively insensitive to changes in cartilage stiffness (not shown). This supports the hypothesis that cartilage does not act to alter the magnitude of the resultant force transmitted through the joint. Instead, this soft tissue acts to decrease the contact stresses at the joint. If the bones were perfectly rigid, the contact area would be a point. In this case, the stresses created by an applied load at the point of contact would are infinitely high. In the real knee, however, the presence of cartilage allows the joint to have some compliance. Therefore, when muscles develop force and press the bones together, cartilage deforms, and, by so doing, increases the contact area between the bones. For a given force transmitted through the joint, an increase in contact area *decreases* the stresses within the layer of cartilage.

3. Sensitivity to Changes in Ligament Stiffness

Loch et al. (1992) investigated the effects of ligament stiffness on the forces transmitted to the knee ligaments. Not surprisingly, they found that stiffer ligaments lead to higher graft forces. To study the sensitivity of ligament and joint-contact forces to changes in knee-ligament stiffness, we altered the values of the elastic moduli of the model ACL and MCL by 50%. For each condition, cruciate and collateral ligament forces were then calculated for a static quadriceps leg raise. We chose to simulate the quadriceps leg raise rather than maximum isometric knee extension because ligament forces are much smaller in the former task (see part I by Kim, 1998). As a result, the sensitivity of model response to changes in the properties of the ligaments will be more apparent during the quadriceps leg raise.

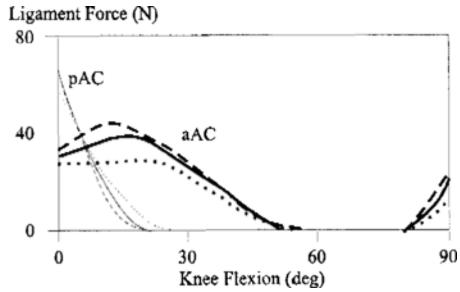
3.1 Sensitivity to ACL stiffness

Increasing or decreasing the stiffnesses of the aAC and pAC by 50% has practically no effect on the force applied by the quadriceps (not shown). The forces borne by the separate bundles of the PCL and MCL also appear to be relatively insensitive to changes in the stiffnesses of the separate bundles of the ACL (Fig. 2(b)-(c)). The forces borne by the ACL, however, are noticeably different (Fig. 2(a)).

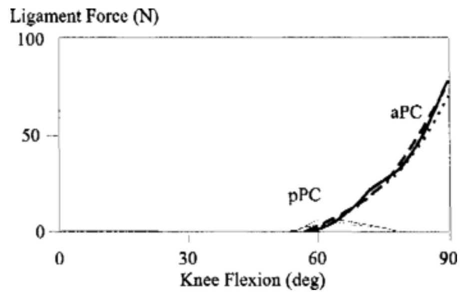
In general, changes in ACL stiffness alter the forces borne by the aAC much more so than the forces borne by the pAC. A 50% increase or decrease in the stiffnesses of the model aAC and pAC produce peak changes of around 10 N in the force borne by the aAC. The corresponding change in pAC force, however, is less than 5 N. Interestingly, a 50% increase in ACL stiffness causes the force in the aAC to increase for nearly all angles of knee flexion, whereas the force in the pAC actually decreases at some knee angles. These results imply that changes in the stiffness of the ligaments change not only the magnitude of ligament force, but also the distribution of force within the cruciate and collateral ligaments.

3.2 Sensitivity to MCL stiffness

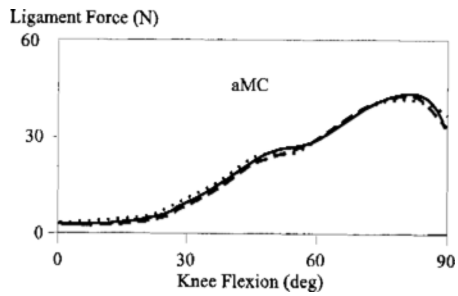
Increasing or decreasing the stiffnesses of the aMC by 50% has no effect on the applied quadriceps force (not shown). There was also relatively little effect on the forces borne by the separate bundles of the ACL and PCL (Fig. 3(a)-(b)). However, the forces produced in the aMC are significantly different when the stiffness of this ligament is altered (Fig. 3(c)). A 50% increase or decrease in the stiffness of the aMC causes ligament force to change by as much as 20%. And, as expected, increasing the stiffness of the MCL produces higher forces in the ligament.



(A) Effect of ACL stiffness on the forces produced in the model ACL (aAC, heavy lines; pAC, light lines).

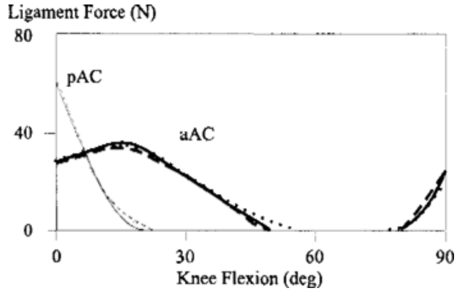


(B) Effect of ACL stiffness on the forces produced in the model PCL (aPC, heavy lines); pAC, light lines).

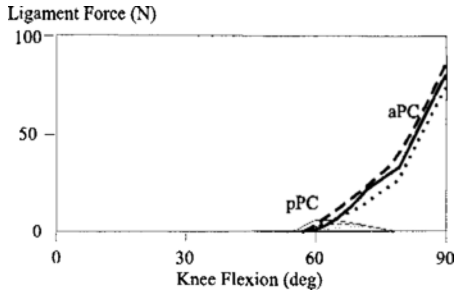


(C) Effect of ACL stiffness on the forces produced in the anterior bundle of the superficial model MCL (aMC).

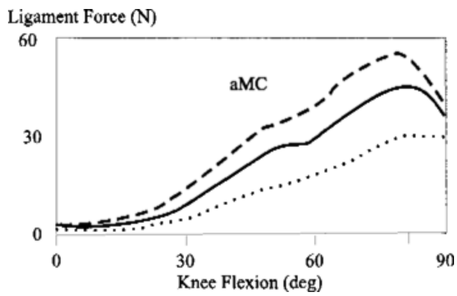
Fig. 2 Effects of ACL stiffness on the forces produced in the cruciate and collateral ligaments of the knee during the quadriceps leg raise simulation in the model. The stiffnesses of both bundles of the model ACL were changed. The stiffnesses of the aAC and the pAC were decreased by 50% (dotted lines) and increased by 50% (dashed liness).



(a) Effect of MCL stiffness on the forces produced in the model ACL (aAC, heavy lines; pAC, light lines).



(b) Effect of MCL stiffness on the forces produced in the model PCL (aPC, heavy lines;pAC, light lines).



(c) Effect of MCL stiffness on the forces produced in the anterior bundle of the superficial model MCL (aMC)

Fig. 3 Effects of MCL stiffness on the forces produced in the cruciate and collateral ligaments of the knee during the quadriceps leg raise simulation in the model. The stiffnesses of all three bundles of the superficial portion of the model MCL were changed. Specifically, the stiffnesses of the aMC, iMC and pMC were all decreased by 50% (dotted lines) and all increased by 50% (dashed lines).

4. Sensitivity to Changes in Ligament Reference Strains

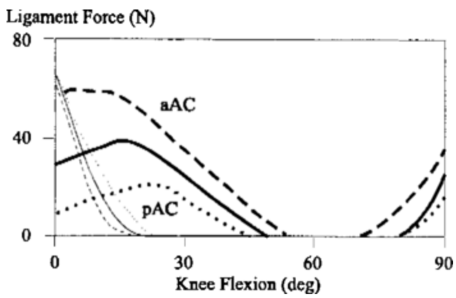
Blankevoort and Huiskes (1991) examined the sensitivity of bone movements to changes in the reference strains of the knee ligaments. They changed the reference strains of the MCL by 3% and calculated the corresponding change in the envelope of passive flexion in a model knee. Decreasing the reference strain of the model MCL was found to increase both the external rotation and valgus rotation of the tibia relative to the femur.

Loch et al. (1992) studied the effects of changing the reference lengths of the ligaments on the forces transmitted by the ligaments during an anterior drawer test. The model calculations showed that smaller values of ligament reference lengths produced higher values of ACL graft forces. Specifically, a 50% change in graft force was predicted by the model when the reference length of the ACL was changed by only 0.7 mm.

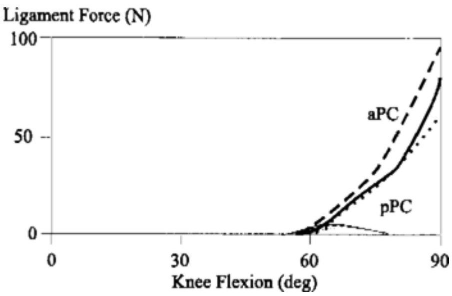
We used the model to study the sensitivity of knee-ligament and joint-contact forces in the model to changes in ligament reference strains. After changing the reference strains of the model aAC and pAC by 5%, the forces transmitted to the cruciate and collateral ligaments were calculated for the quadriceps leg raise.

4.1 Sensitivity to ACL reference strains

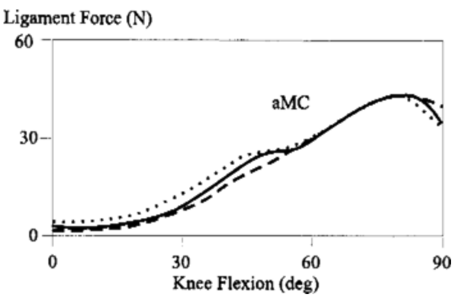
Increasing the reference strains of the model aAC and pAC by 5% produces an increase in aAC force only (Fig. 4). Decreasing the reference strains of the model ACL by 5% decreases the force in the aAC at all angles of knee flexion. The model also predicts a reciprocal relationship between ligament reference strains and the forces produced in the pAC. Specifically, pAC force in the model decreases when ligament reference strains are increased, and increases when the reference strains are decreased. This behavior was also evident when ligament stiffness in the model was changed (Fig. 4). In that case, increasing ligament stiffness caused the force in the model pAC to decrease, whereas decreasing ligament



(a) Effect of ACL reference strains on the forces produced in the model ACL (aAC, heavy lines; pAC, light lines).



(b) Effect of ACL reference strains on the forces produced in the model PCL (aPC, heavy lines; pAC, light lines).



(c) Effect of ACL reference strains on the forces produced in the anterior bundle of the superficial model MCL (aMC).

Fig. 4 Effects of ACL reference strains on the forces produced in the cruciate and collateral ligaments of the knee during the leg raise task. The reference strains of both bundles of the model ACL were changed. Specifically, the reference strains of the aAC and the pAC were decreased by 5% (dotted line) and also increased by 5% (dashed line).

stiffness caused an increase in ligament force (Fig. 4(a)). These results can be understood by considering the relationship between a change in ligament stiffness and a change in ligament reference strain. Increasing the reference strain of a ligament is equivalent to decreasing its reference length. Decreasing ligament reference length is equivalent to increasing ligament stiffness. Increasing the reference strain of a ligament is therefore equivalent to increasing the stiffness of the ligament. This explains why pAC forces in the model behave the same way for similar changes in both ligament stiffness and ligament reference strains.

The above results suggest that changes in the reference strains of the model ligaments produce changes not only in the magnitude of the forces transmitted to the ligaments, but also in the distribution of force within each of the ligaments. Increasing either the stiffness or reference strain of the model ACL leads to an increase in aAC bundle force only. The forces transmitted to the pAC and aMC decrease in both cases.

Changes in MCL force brought about by changes either in the stiffness or reference length of the ACL can be explained by changes in the relative displacements of the bones which meet at the knee. Changing the mechanical properties of the ACL causes a change in the position and orientation of the tibia relative to the femur, which alters the force transmitted to the MCL (Fig. 4(c)). In fact, increasing the reference strains of the ACL by 5% decreases the amount of internal rotation undergone by the model tibia as the knee flexes toward 90° (not shown). The change in axial rotation of the tibia is relatively small, however, which explains the correspondingly small change in MCL force.

The results of the sensitivity analysis suggest that the calculated values of ligament forces are much more sensitive to changes in ligament reference strains than to changes in ligament stiffness (compare aAC in Figs. 2 and 4). When the stiffnesses of the anterior and posterior bundles of the model ACL were changed by 50%, the forces transmitted to the aAC changed by no more than 10 N (Fig. 2(a)). However, when the reference strains of these ligaments were changed by only 5%, aAC forces in the model changed by as much as 20 N for the quadriceps leg raise task (Fig. 4

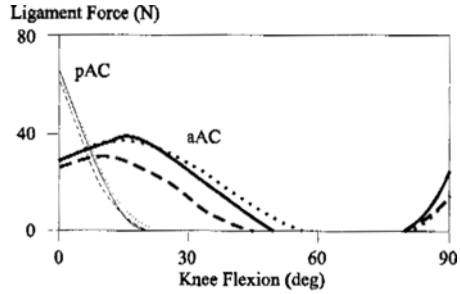
(a)). Larger changes also occurred in PCL and MCL forces when ligament reference strains in the model were changed (compare aPC, pPC, and aMC in Figs. 2(b)-(c) and 4(b)-(c)). These predictions of the model are consistent with calculations reported by Loch et al. (1992), who found that ACL graft forces were highly sensitive to changes in the ligament reference length. On the basis of their model and experimental results, these researchers concluded that the reference length of a ligament is the variable which causes large variations in graft forces during ligament reconstruction.

4.2 Sensitivity to MCL reference strains

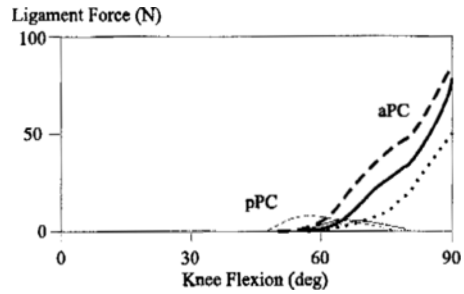
Changing the reference strain of the model aMC produces significant changes in the recruitment patterns and in the magnitudes of the forces borne by the collateral ligaments (Fig. 5(c)). Relatively small changes, however, occur in the forces borne by the cruciate ligaments during a quadriceps leg raise (Fig. 5(a)-(b)).

A 5% increase in the reference strain of the aMC causes the force in this ligament to increase by a factor of 3. Furthermore, the intermediate bundle of the MCL is now recruited at small angles of the knee, whereas previously it remained slack at all angles of flexion. Decreasing the reference strain of the aMC by 5% causes a substantial decrease in the aMC force, so much so that this bundle remains slack over a large portion of the joint range of motion.

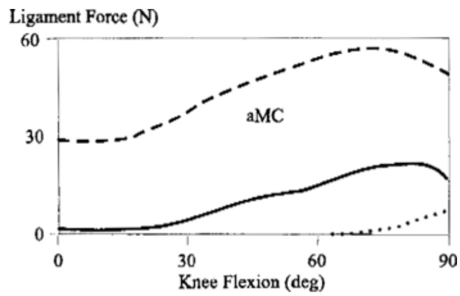
These effects on MCL forces are clearly visible in the axial rotational response of the model knee. Very large changes in the internal-external rotation of the tibia relative to the femur are brought by changes made to the reference strains of the MCL (not shown). A 5% increase in the reference strain of the aMC causes the internal rotation of the tibia to increase by as much as 10° as the knee approaches 90° of flexion. It is significant that such a change is still not able to produce a response of the model which exhibits the screw-home mechanism of the knee near full extension. Finally, a 5% decrease in the reference strain of the aMC decreases the internal rotation of the tibia by nearly 20°. These results indicate that the reference strains of the collateral ligaments have a



(a) Effect of MCL reference strains on the forces produced in the model ACL (aAC, heavy lines; pAC, light lines).



(b) Effect of MCL reference strains on the forces produced in the model PCL (aPC, heavy lines; pAC, light lines).



(c) Effect of MCL reference strains on the forces produced in the anterior bundle of the superficial model MCL (aMC)

Fig. 5 Effects of MCL reference strains on the forces produced in the cruciate and collateral ligaments of the knee during the quadriceps leg raise simulation in the model. The reference strains of all three bundles of the superficial portion of the model MCL were changed. Specifically, the reference strains of the aMC, iMC and pMC were all decreased by 5% (dotted lines) and also all increased by 5% (dashed lines).

profound effect on the total rotational laxity of the model knee. By comparison, changes to the properties of the collateral ligaments have relatively little effect on the anterior-posterior laxity of the knee (Fig. 10A-B).

5. Conclusion

There are many parameters in our knee model: parameters which specify the inertial properties of segments (i. e., length, mass, location of the center of mass, and moment of inertia), as well as those which define the mechanical and geometric properties of muscle (e. g., peak isometric strength, and optimal fiber length), tendon (e. g., cross-sectional area and slack length), ligament (e. g., slack length and modulus of elasticity), and cartilage (stiffness).

Many of these parameters have been carefully measured and reported in the literature. Nominal values for optimal muscle-fiber length, muscle cross-sectional area (and therefore peak isometric force), ligament and tendon stiffness, and the inertial properties of the body segments have all been measured and reported with a reasonable degree of consistency. However, to better understand the sensitivity of performance to small changes in the parameters of the model, we conducted a detailed sensitivity analysis.

Simulations of maximum isometric knee extension and quadriceps leg raising were used to explain the variation in cruciate ligament loading at the knee resulting from the interaction between the external forces applied to the bones, the action of the muscles, and the articular-contact forces transmitted through the joint. For a distal location of the restraining pad, the ACL is loaded from 70° of flexion to full extension during maximum isometric extension of the model knee. Quadriceps-strengthening exercises should therefore be avoided in this region of flexion if the ACL is to be protected from bearing load. The pattern of cruciate ligament loading at the knee is determined by the variation in the resultant shear force applied to the tibia. During maximum isometric extension, ACL forces in the region from full extension to 20° of flexion are governed by the force-length properties of the quadriceps muscles. At all angles beyond 20°, however, the pattern of ACL force is determined by the geometry of the articulating surfaces of the joint.

Through sensitivity analysis of the parameters used in the model knee, we found that the ligament function in the real knee was most sensitive to changes in ligament reference lengths or strains, less sensitive to changes in cartilage stiffness, and perhaps least sensitive to changes in ligament stiffness. Constraining movements of the bones to the sagittal plane also produced much higher forces in the cruciate and collateral ligaments than for the three-dimensional model, though the result was not shown in the paper. These constaints increased the anterior-posterior translation of the tibia relative to the femur, which then increased the forces transmitted to the ligaments.

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