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ACL function in the normal knee

Función de LCA en la rodilla normal

Summary

A biomechanical description of the anteroexternal cruciate ligament of the knee in normal conditions is made.

Key words: Function. Mechanics. Anteroexternal cruciate ligament. Knee joint. Cruciate ligaments.

Resumen

Se describe la biomecánica del ligamento cruzado anteroexterno de la rodilla en condiciones normales.

Palabras clave: Función mecánica. Ligamento cruzado anteroexterno (LCA). Articulación de la rodilla. Ligamentos cruzados.

Introduction

The ligaments of the knee have a number of apparently quite distinct mechanical functions. They define the motion limits of the bones on each other. They stretch and transmit tensile forces to resist movement outside those limits, pulling the articular surfaces together into compression. Within the motion limits, they control and guide the movements of the bones on each other. They also work in partnership with the muscles and articular surfaces in transmitting the loads of activity across the joint.

All knee structures interact geometrically and mechanically with each other; to explain the function of one, it is necessary to consider its interactions with the others. In this chapter, both cruciate ligaments will be discussed, although the functional role of the anterior cruciate ligament (ACL) will be given greater emphasis. After a summary of the roles of the cruciates in limiting joint movements, a simple geometric and mechanical model of the knee, implemented on a computer, will be presented and used to explain other important, yet more subtle, cruciate functions. The model is based on ligament functional architecture and represents a series of hypotheses about the way the natual joint works; comparison between model predictions and experimental observations on human specimens tests the validity of these hypotheses. The simplest model treats the ligaments as inextensible straight lines. This model is sufficient to explain how the ligaments control the

rolling movements of the bones on each other during flexion and extension. It illustrates the relationship between the geometry of the cruciate ligaments and the geometry of the articular surfaces of the tibio-femoral joint. An elaboration of the model treats the ligaments as ordered arrays of fibers. This development explains the changing shapes of the ligaments and the patterns of strain induced in the ligaments during passive flexion/extension. It throws light on the concept of isometry and its limitations. Finally, the model is used to show how ligament fibers are progressively recruited to bear load and how the effective ligament cross-sectional area increases to resist increasing antero-posterior (a-p) movements of the bones.

These models will be further developed in chapter II.9 to investigate interactions between ligaments and muscles during activity. The possible proprioceptive functions of the ligaments will not be discussed.

Restraints to joint motion

The ligaments limit the movement of the bones upon each other and define the range of passive movement[5]. In the simplest terms, the ACL resists anterior subluxation of the tibia; the posterior cruciate ligament (PCL) resists posterior subluxation; the medial collateral ligament (MCL) resists abduction; the lateral collateral ligament (LCL) resists adduction; the posterior capsule resists hyperextension. This description is obviously far too simple. The ligaments act together in groups to limit these motions. For instance, all the ligaments limit distraction of the joint. SHOEMAKER and DANIEL[37] have summarised the large body of experimental evidence defining the primary and

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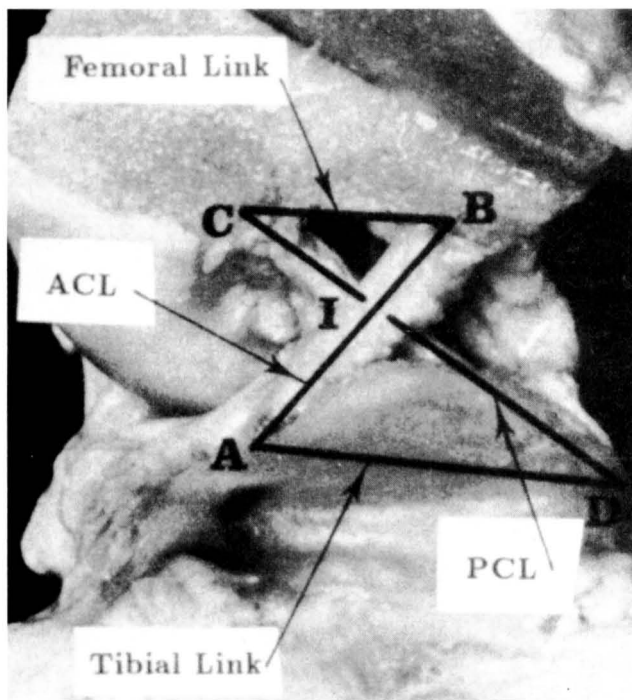Parks Road, Oxford OX1, 3PJ, England.

FIG. 1.—A human knee with the lateral femoral condyle removed, exposing the cruciate ligaments. Superimposed is a diagram of a four-bar linkage comprising the anterior cruciate ligament AB, the posterior cruciate ligament CD, the femoral link CB joining the ligament attachment points on the femur, and the tibial link AD joining their attachments on the tibia.
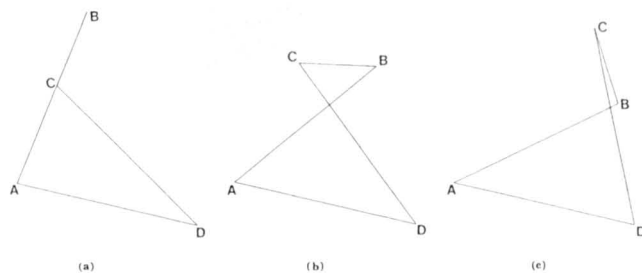

FIG. 2.—The cruciate linkage ABCD drawn by the computer at full extension and at 70° and 140° flexion. Between (a) and (c), the femoral link CB rotates through 140° relative to the tibial link AD, and the cruciate ligaments AB and CD rotate through 40° about their tibial attachments A and D and through 100° about their femoral attachments B and C.

secondary ligamentous restraints to various movements. GROOD (Chapter II.6) presents a similar review. The restraining functions of the cruciates are noted briefly below.

The ACL

The major function of the ACL is to resist anterior displacement of the tibia with respect to the femur at all flexion positions[8, 11, 12, 16, 20, 29, 35]. It has been reported that the antero-medial part of the ACL is most important in flexion, while the postero-lateral part contributes most to stability in extension[3, 17]. The ACL also helps to resist hyperextension[17, 20]. It contributes to the rotational stability of the knee at full extension for both internal and external tibial rotation[12, 17, 20]. In flexion, it has been noted to resist internal rotation only[34], and to resist both directions of rotation, depending on the part of the ligament considered[17]. The ACL resists medial displacement of the tibia[34], but it plays only a small role in maintaining the varus/valgus stability of the knee[21].

The PCL

The major function of the PCL is to resist posterior displacement of the tibia[8, 11, 12, 16, 17, 20, 29, 34]. It also resists late-

ral movement of the tibia, but it contributes only minimally to rotational stability[12, 20, 21] and to varus/valgus stability[12, 21, 34].

Guides to joint motion

The cruciate ligaments contain fibers which rotate about their points of origin and insertion on the bones to allow and to control movement within the motion limits of the knee. These actions can be explained by reference to a theoretical model of the joint.

The four-bar cruciate linkage

The cruciate ligaments and the bones form a mechanical linkage which guides the movements of the bones upon each other. A simplified version of this mechanism is the four-bar cruciate linkage ABCD shown in figure 1[33, 34]. Two links, AB and CD, represent the ACL and PCL, and two links, BC and AD, join their points of attachment on the femur and on the tibia. Changes in the flexion angle of the joint result in equal changes in the angle between the femoral link BC and the tibial link AD. Figure 2 shows the linkage at extension, 70° and 140° flexion, with the tibial link fixed. The shape of the linkage changes as the ligament links rotate about their attachment points. The three diagrams of figure 2 are shown superimposed in figure 3. With the femur flexing and extending on a fixed tibia (fig. 3 A), the femoral attachments B and C of the cruciates rotate on circular arcs about their tibial attachments A and D, respectively, while the ligament links AB and CD remain isometric. Similarly, in figure 3 B, with the tibia flexing and extending on a fixed femur, the tibial attachments A and D of the ligament links rotate isometrically about their femoral attachments B and C, respectively.

The instantaneous center of the linkage

The instantaneous center of the linkage lies at the point I (figs. 1 and 3) at which the ligament links cross. The fle-
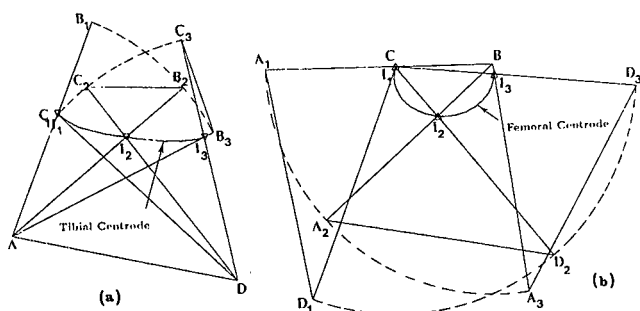
FIG. 3.—The cruciate linkage drawn by computer with the tibial link fixed (a) and with the femoral link fixed (b). The relative positions of the links are the same in the pair of diagrams for each of the corresponding three configurations. In (a), the femoral attachments of the ligaments move on circular arcs about their tibial attachments, and in (b), the tibial attachments move on circular arcs about the femoral attachments. The curves marked «tibial centrode» and «femoral centrode» are the tracks of the flexion axis of the joint on the tibia and femur, respectively, and are drawn through successive intersections of the cruciates.

xion axis of the joint, about which the bones flex and extend, passes through I. Because the geometry of the linkage changes during flexion and extension, the instantaneous center moves backwards and forwards relative to the two bones. It moves relative to the tibia along the curve marked «tibial centrode» (fig. 3 A) and relative to the femur along the curve marked «femoral centrode» (fig. 3 B).

Relation between ligament geometry and articular surface shape

The shapes of the articular surfaces of the bones must be compatible with the changing geometry of the cruciate linkage. They must allow passive flexion and extension while maintaining the isometricity of the cruciate links. If one arbitrarily chooses the shape of one of the articular surfaces, what should be the compatible shape of the other surface?

An alternative statement of the problem is to ask where a compatible femoral surface should touch a specified tibial surface. The theoretical solution is illustrated in figure 4, where the shape of the tibial surface is chosen to be flat and the shape of the femoral surface is sought. If the position on the tibia of the contact point can be found, its position relative to the femoral link BC can be calculated. When this is done over the flexion range, the shape of the compatible femoral condyle is obtained. In figure 4, contact must occur at the point F where the perpendicular to the tibial plateau passes through the instantaneous center I. Since all points on the femur rotate in circles centered at I, only the point F on the femur can move tangential to the tibial plateau while maintaining the isometry of the ligament links. Contact at F_1 or F_2 would cause, respectively, either interpenetration or separation of the bones and tightening or slackening of the cruciate links. The shape of the femoral surface shown in figure 5 was calculated on this basis.
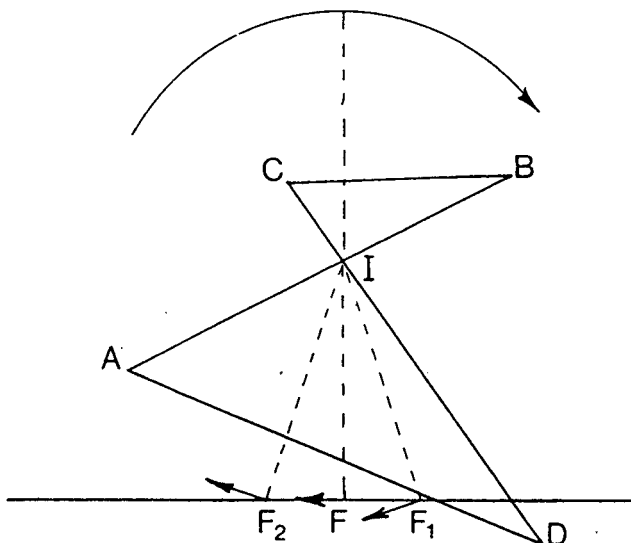

FIG. 4.—Demonstration that contact between a flat tibial plateau occurs at the point F where the common normal to the articular surfaces at their point of contact passes through the flexion axis at I. With the surface of the tibia fixed, all points on the femur move on circular arcs centered on the flexion axis. Contact between the bones at F_1 would lead to interpenetration and at F_2 to separation of the bones. Contact at F allows the bones to slide upon each other without interpenetration or separation.

Similar methods have been used[32, 33] to calculate the shapes of femoral surfaces compatible with concave and convex tibial plateaux. The calculated shapes have been shown to be very similar to the medial and lateral femoral condyles of the human knee. Detailed examination of the curvature of the calculated surface shapes shows them to be nearly circular in the sagittal plane, consistent with the conclusion of KUROSAWA et al.[25] that the femoral condyles of the human knee are spherical.
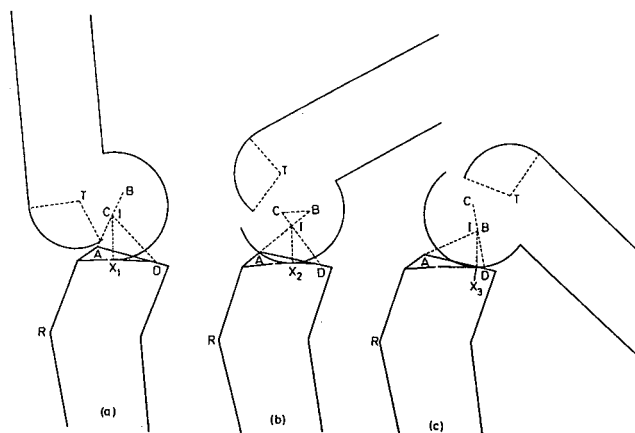

FIG. 5.—Model bones and cruciate linkage shown in three flexion positions. $ABCD$ = four-bar cruciate linkage, I = instantaneous center, T = center of trochlear facet of femur, R = tibial tubercle, X_1, X_2, X_3 = tibio-femoral contact points.

Movements of the bones

The rolling as well as sliding of the femoral condyles on the tibial plateaux was first described by WEBER and WEBER in 1836[39]. An elaboration of the computer model shown in figure 5 demonstrates this movement. The contact point between the model bones follows the movement of the instantaneous center, shown in figure 3, and moves backwards during flexion and forwards during extension. Because the articulation lies distal to the instaneous center, the femur not only rolls backwards but slides forwards on the tibia during flexion, *vice versa* during extension. It has been shown previously[32, 33] that the movements of the model femur on the tibia during flexion/extension agree well with measurements on human specimens in the laboratory.

Conclusions

The four-bar linkage model presented in this section highlights the role of the cruciates in guiding flexion of the knee, and it relates ligament anatomy to some important aspects of knee function. The results suggest that each human cruciate ligament has one point within its tibial attachment area which remains a constant distance from a corresponding point within its femoral attachment area during passive flexion/extension of the knee. If these points are joined by an actual ligament fiber (the «neutral fiber»[7] or «guiding bundle»[18]), then that fiber remains isometric during passive knee movements. This idea is used as the basis for an elaboration on the four-bar linkage model in which the ligaments are treated as ordered arrays of fibers.

Modelling of the ligaments

Modelling the ligaments as straight inextensible lines revealed the relationship between ligament geometry and the shapes and movements of the bones in the sagittal plane; however, such a simple model cannot explain the functional subtleties of the ligaments, *e.g.* the patterns of strain which develop within the ligaments during passive movements and during load transmission. To do this, it is necessary to consider the ligaments as arrays of fibers and to calculate the geometrical and mechanical relationships between the fibers.

When the knee is entirely unloaded, it is assumed that ligament fibers are either slack or just tight, transmitting no force. Passive movement of the joint, in the absence of loads, can then be considered as a movement from one state of neutral equilibrium to another.

Geometric representation of ligament fiber architecture involves a description of the bony attachment areas of the ligaments and of the relationships between the femoral and tibial attachment positions of each fiber, *i.e.* the fiber mapping. It is also necessary to specify the position of the neutral fiber within each cruciate ligament, corresponding to the ligament link of the four-bar linkage described above, and to specify the zero-tension reference position at which all other fibers within each ligament are just tight. Although the cruciate and collateral ligaments have been mo-
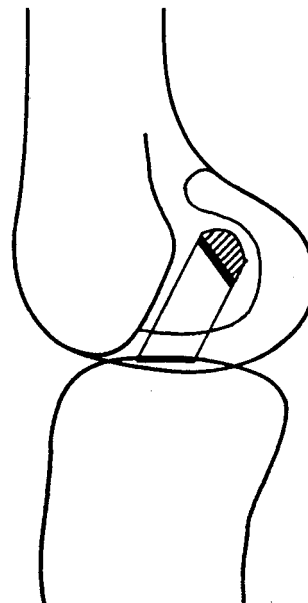

FIG. 6.—The tibial and femoral attachment areas of the ACL. Heavy lines represent the model ligament attachment lines. Light lines connecting the attachment areas represent the model anterior and posterior fibers.

delled previously using this method[42], only its development for the ACL is described in this section.

Attachment lines

A sagittal view of the femoral and tibial attachment areas of the ACL is shown in figure 6. The attachments are modelled as straight line segments, included as heavy black lines in the figure.

Fiber mapping

A fiber mapping between the tibial and femoral attachment lines must be specified. Because the human ACL has been described as having a flat band-like appearance in the sagittal plane at full extension of the knee[20, 36] with consistently matched origins and insertions[18, 31], a mapping with individual model fibers arising and inserting at the same proportional distance along each bony attachment line was used. In extension, the model ACL therefore consists of a sheet of fibers, all more or less parallel. The model fibers are defined as straight lines connecting points on the tibia and femur. In reality, slack fibers tend to buckle into curved shapes while tight fibers remain straight.

The neutral fiber and the zero-tension reference position

In developing four-bar linkage theory to analyze ligaments of finite size, the positions of the neutral fibers wi-

thin the cruciate ligaments must be chosen. Based on a radiographic study of cruciate ligament fiber lengths, FUSS[18] concluded that the neutral fiber in the human ACL originates most anteriorly within the femoral attachment area in extension and inserts most anteriorly within the tibial attachment area. The model ACL in extension should lie entirely behind its neutral fiber and hence behind the instantaneous center of the joint. All but the anterior fiber will then slacken with initial flexion. Therefore, the anterior fiber of the model ACL was chosen as a neutral fiber in the four-bar linkage model, and the position of extension was used as the zero-tension reference point for calculating model ACL fiber strain.

Model parameters

The outcome of the calculations depends on the choice of model parameters: the locations of the ligament bony attachment lines, the zero-tension reference positions, and the lengths of the links in the four-bar cruciate mechanism. The results for only one choice of parameters are reported here. They are based mainly on anatomical measurements reported by VAN DIJK et al.[13, 14, 26] A detailed description of the parameters and arguments leading to their choice have been reported elsewhere[42].

ACL shape changes during passive flexion

Figure 7 shows the model femur, tibia, and cruciate ligaments drawn in three flexion positions. As flexion progresses and corresponding tibial and femoral attachment sites rotate relative to each other, the fiber orientations and the overall shapes of the ligaments change. By 120° flexion, some fibers are crossed. The changing shapes of the model ligaments in figure 7 are quite similar to those of the human ligaments as seen from the side and described by
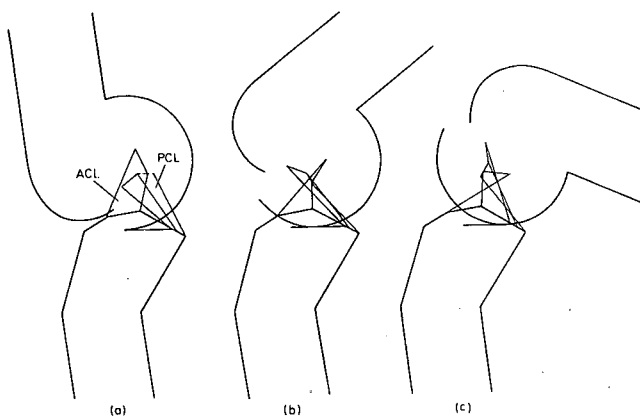
BRANTIGAN and VOSHELL[8], GIRGIS et al.[20], VAN DIJK et al.[13, 14, 26] and O'BRIEN et al.[31]

A number of investigators have described separate and distinct fiber bundles within the body of each cruciate ligament[4, 23, 24, 30]. The present work suggests that such bundles are manifest because of the relative movements of the attachment areas of the ligaments during flexion/extension. For instance, the ACL in extension is modelled by a sheet of fibers all more or less parallel (fig. 7 A). At 120° flexion (fig. 7 C), the same fibers are crossed because of relative rotation of the attachment areas. Fibers attached at the front and the back of the tibial attachment area are now pointing in very different directions. Although the anterior and posterior halves of the human ACL resemble distinct bands in the flexed knee, the same fibers are more or less parallel in the sagittal plane in the extended knee.

ACL fiber length changes during passive flexion

Figure 8 shows the calculated relative length changes of selected fibers of the model ACL in figure 7 plotted against flexion angle. Relative fiber length is defined as calculated fiber length divided by length at the zero-tension reference position. The distance apart of a pair of points representing the origin and insertion of a fiber on the femur and tibia is calculated and is assumed to represent fiber length.

Ligament fibers which lie behind the flexion axis stretch and those which lie in front slacken during passive flexion of the knee[2, 5]. The slackening and subsequent tightening of the central and posterior model ACL fibers with increasing flexion, as shown in figure 8, is a consequence of their movements from behind to in front of the flexion axis as the knee flexes. Our calculated fiber length changes are
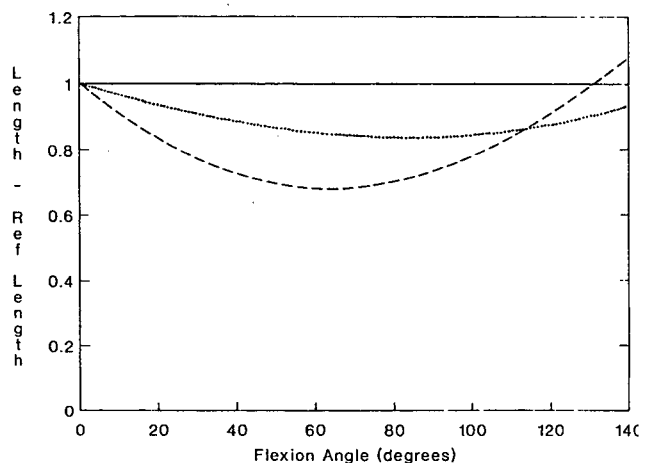
FIG. 7.—The model femur, tibia, and cruciate ligaments drawn in 0°, 60°, and 120° flexion. Both anterior and posterior parts of the PCL are shown.

FIG. 8.—The calculated relative length changes of the anterior, central, and posterior fibers of the ACL. Anterior fiber (solid line), central fiber (dotted line), posterior fiber (dashed line). Relative length equals unity at full extension, the zero-tension reference position.

comparable to those measured on human specimens by SAPEGA et al.[36] They found that fibers attached antero-medially on the femur in extension had the least deviation from isometry during flexion, although their measurements showed an initial slackening as flexion commenced. They also observed slackening followed by lengthening of the central and posterior fibers with flexion. AMIS and DAW-KINS[4] found similar patterns in central and posterior fibers. Their most anterior fiber showed initial slackening followed by tightening beyond the full extension length.

Fibers which pass through the flexion axis of the joint remain instantaneously isometric. Clearly, it is unlikely that a bundle of fibers occupying a finite area could all pass through a single point so that the concept of fiber isometricity can apply only to individual fibers. The clinical objective of implanting a graft or prosthesis so that all its fibers remain isometric during passive movements cannot be met.

Ligament fiber recruitment under load: Geometry

A knowledge of the forces which knee ligaments withstand under load is necessary for the design of grafts or prosthetic replacements for injured ligaments. Quantification of ligament forces *in vitro* is a difficult task, and only a few attempts at measurement have been reported[1, 27, 28]. Problems of calculating the distributions of strain and stress within individual ligaments and the sharing of load between different ligaments are statically indeterminate; force equilibrium conditions alone do not provide sufficient information. Geometrical analysis of the deformed configurations of the ligaments can supply the additional compatibility conditions necessary for solution of the statically indeterminate ligament-force problem, at least in the absence of muscle action.

In this and the following section, geometric compatibility is used to determine strain distributions and forces in the model ligaments produced by a-p tibial translation. This is essentially an analysis of ligament fiber recruitment, which occurs when ligament fibers made slack by passive flexion of the knee stretch and change their spatial orientations in order to resist applied loads. The concepts of fiber recruitment and of functional architecture as alluded to here have been developed mainly by physiologists studying skeletal muscle[19]. Their chief application to the study of ligaments has been in the modelling of the non-linear mechanical behaviour of ligaments, specifically in the relationship between fiber crimp and tissue stiffness[41]. The ideas of minimal fiber-bundle recruitment length and of recruitment probability have recently been put forth by BLANKEVOORT and colleagues[6] and have been applied by them to the results of an experimental study of cruciate ligament fiber bundle lengths.

The presente work is a development of an analysis by FITZPATRICK and O'CONNOR[15]. They considered each ligament as consisting of a single extensible line and calculated ligament strains and forces compatible with specified flexion angles and anterior tibial translations.

Geometric development

Figure 9 A shows a single fiber connecting femur and tibia. A forward movement x of the tibia relative to the femur displaces the fiber from its unloaded position, decreases its inclination with the tibial plateau and increases its length. From the geometry of the triangle in figure 9 A, the stretched length of the fiber *compatible with* a tibial displacement x can readily be calculated. Figure 9 C shows the anterior and posterior fibers of the model ACL (of fig. 7) and MCL[43] moved to new positions by successive anterior displacements of the tibia. The distribution of fiber elongation or shortening in these model ligaments compatible with any specified tibial displacement can be calculated[43].
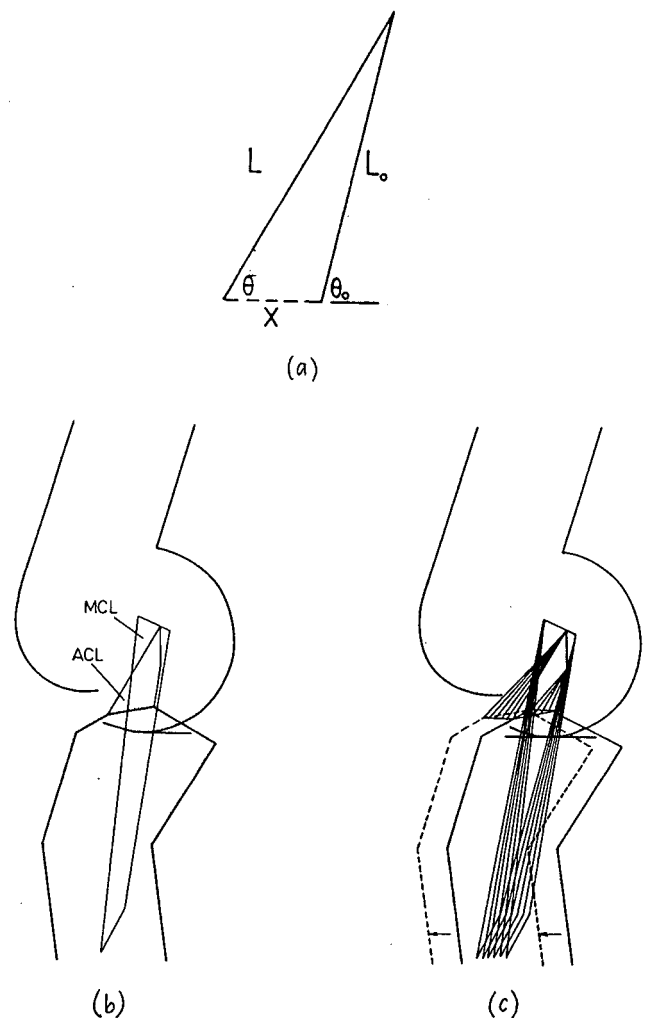

FIG. 9.—(a) A single fiber connecting femur and tibia. In the unloaded state, the fiber has a length L_o and is inclined to the tibia at an angle L_o. When the tibia moves forward a distance x, the fiber stretches to a length ℓ and to an inclination θ compatible with the displacement x. (b, c) The model bones with ACL and MCL (anterior and posterior fibers only) drawn (b) in 25° flexion, and (c) with tibia translated anteriorly 10 mm in steps of 2 mm.
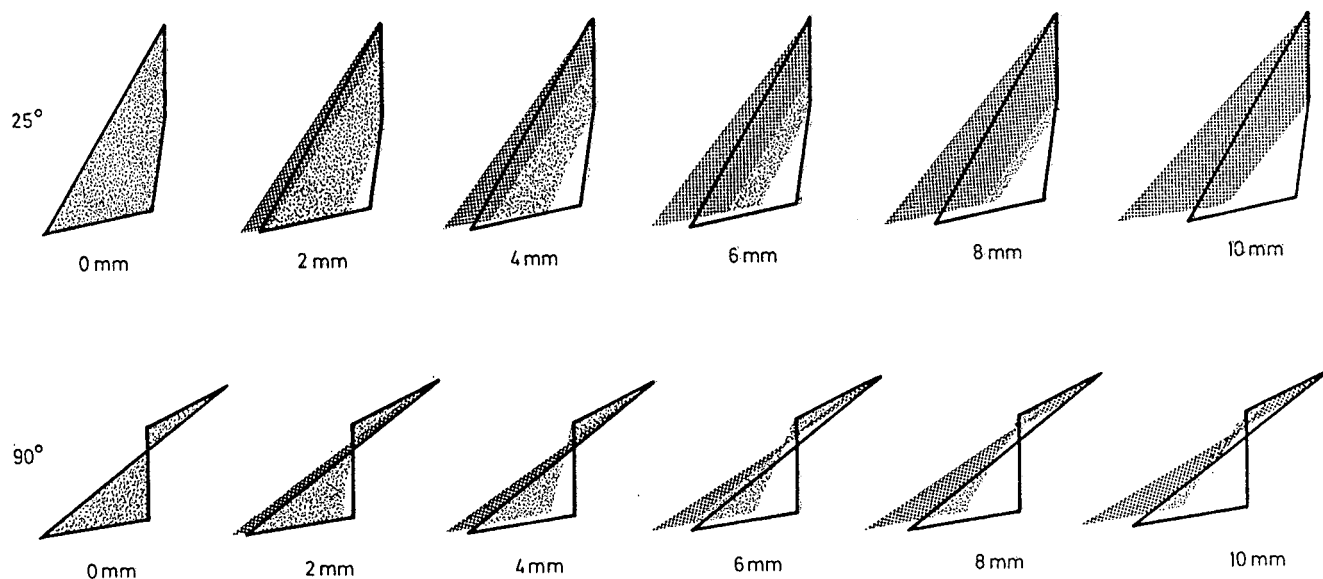
FIG. 10.—The model ACL at 25° and 90° flexion and after imposed tibial translations of 2-10 mm. Heavy black lines = shape of ACL as a result of passive flexion. Dark shading = part of ACL which is taut. Light shading = part of ACL which is slack.

Figure 10 shows diagrams of the model ACL at 25° and 90° of flexion and at 2 mm intervals of anterior tibial displacement. The shape of the ligament as a result of passive flexion is defined by heavy black lines. The dark shading indicates that part of the ligament which has been stretched beyond its zero-tension reference length. The lighter shading indicates that part of the ligament which is still slack. The shape of the ligament in the two positions of flexion and the directions of its fibers differ because of relative movement of its attachment areas on the bones.

In the unloaded state in both positions, the fiber inserting into the anterior point on the tibial attachment is the isometric fiber of the ligament coinciding with the ACL link in the four-bar linkage of figure 1. It rotates towards the tibial plateau under the control of the four-bar linkage of figure 2. The attachment line of the ligament on the femur at 90° has rotated through 65° relative to its orientation at 25°.

When the tibia is moved forward, all fibers stretch and their inclinations to the tibial plateau decrease. Fibers rendered slack by passive movement, as in figure 10, are *progressively recruited* to offer increasing resistance to tibial translation. The effective cross-sectional area of the ligament increases as fibers are successively recruited, and the mechanical stiffness of the ligament increases. An anterior translation of 8 mm is required to tighten all model ACL fibers at 25° but is not sufficient at 90°.

Fiber strain distributions

As a further illustration of fiber recruitment, figure 11 shows the calculated fiber lengths for successive inter-vals of anterior tibial translation for anterior, central and posterior fibers in the model ACL plotted against flexion angle. Figure 10 above was based on a series of such calculations at 25° and 90°. The model anterior fiber (fig. 11 A) which remains isometric during passive flexion, stretches as soon as tibial translation beings. The central and posterior fibers tighten with initial translation only when their lengths are near their zero-tension reference lengths in the positions of extension and full flexion. Larger displacements are required to tighten the central and posterior fibers at all other flexion angles. From 35°-100° flexion, even *10 mm* of anterior translation cannot tighten the model posterior fibers made slack by passive flexion. Similar fiber recruitment occurs in the model PCL, MCL, and LCL.

Numerous experimental studies of knee ligament fiber isometry have been conducted. Some investigators have reported cruciate ligament isometry regions located outside the anatomic attachment areas of the ligaments[4]. Others have not been able to find truly isometric ligament fibers and have concluded that the concepts of isometricity and of the four-bar linkage model, which assumes isometricity, are not valid[4,6]. The present paper challenges these assertions. As defined here, the concept of isometricity applies only during passive, unloaded movement of the joint, a condition nearly impossible to achieve experimentally. Figures 10 & 11 show that even small anterior tibial translations result in stretching of the anterior fiber of the model ACL, the neutral fiber in the four-bar linkage. Thus, even small loads can stretch or slacken fibers, making the interpretation of fiber isometry studies difficult, especially when the conditions for holding and loading the knee specimens are defined only vaguely, as in the isometry study by SIDLES et al.[38]
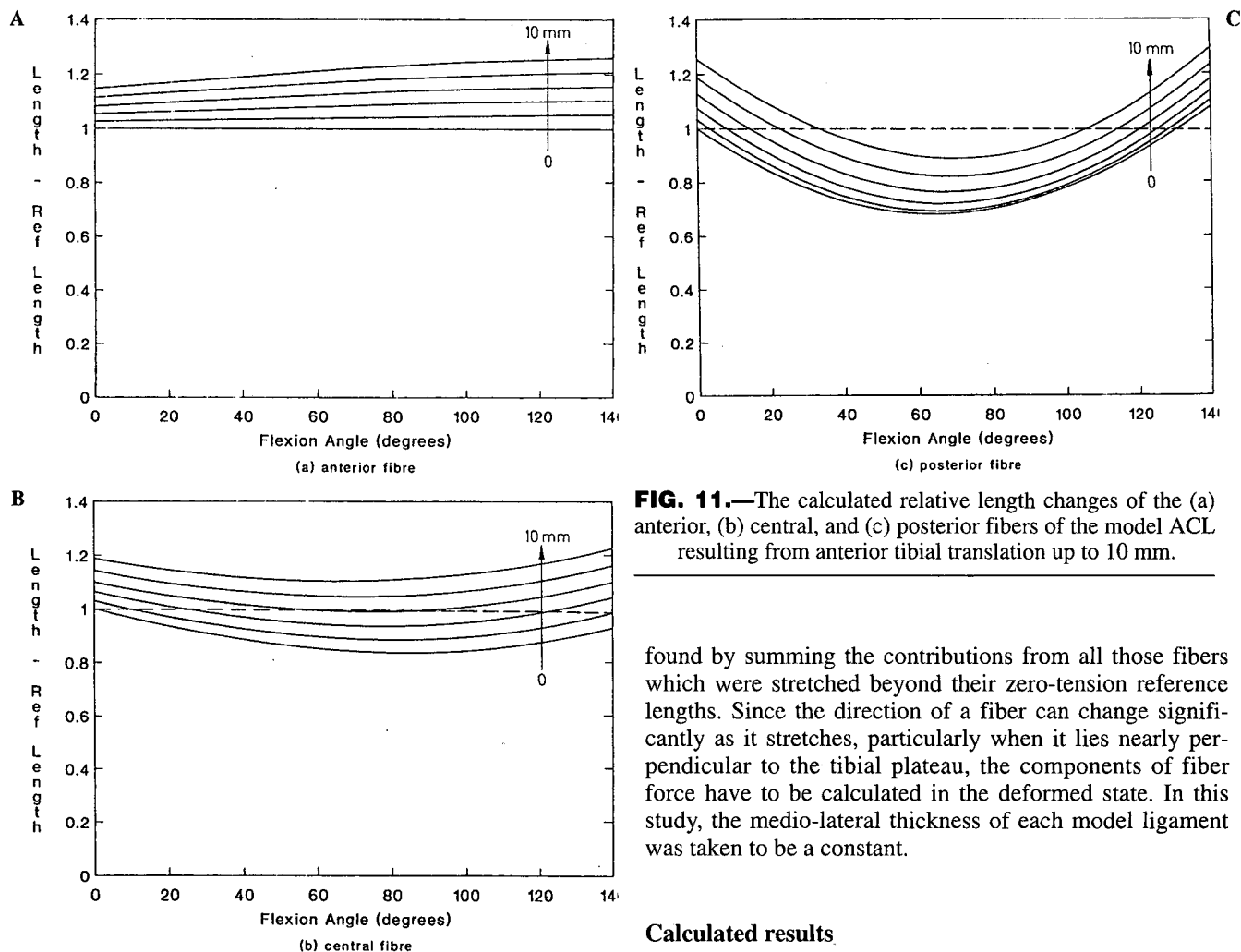
(a) anterior fibre

Flexion Angle (degrees)

Length - Ref Length

10 mm
0

B

(b) central fibre

Flexion Angle (degrees)

Length - Ref Length

10 mm
0

C

(c) posterior fibre

Flexion Angle (degrees)

Length - Ref Length

10 mm
0

FIG. 11.—The calculated relative length changes of the (a) anterior, (b) central, and (c) posterior fibers of the model ACL resulting from anterior tibial translation up to 10 mm.

found by summing the contributions from all those fibers which were stretched beyond their zero-tension reference lengths. Since the direction of a fiber can change significantly as it stretches, particularly when it lies nearly perpendicular to the tibial plateau, the components of fiber force have to be calculated in the deformed state. In this study, the medio-lateral thickness of each model ligament was taken to be a constant.

Ligament fiber recreuitment under load: Mechanics

Fiber stress distributions

For a specified tibial translation, fiber strain can be calculated. Because fiber extension can be significant, up to 20%, a non-linear elongation/strain relationship was used[42]. As a first approximation, a linear stress/strain relationship, with Young's modulus $E = 8000$ N/cm^2, was then used to deduce fiber stress

Ligament force magnitude

The total force in a ligament is the sum of the force contributions from individual fibers which are in tension. For any chosen tibial displacement, the fiber force acting on a small element of tibial attachment area was calculated by multiplying the fiber stress by the area element, taking account of the fiber direction. The components of ligament force parallel and perpendicular to the tibial plateau were

Calculated results

The sum of the horizontal components of the ligament forces must equal the magnitude of the horizontal force which has to be applied to the tibia to achieve the specified a-p displacement. Figure 12 is a series of graphs of applied a-p forces versus tibial displacement for flexion angles of 0°, 25° (the position in which the Lachman test is usually performed) and 90° (the position for the drawer test). The separate contributions of individual cruciate and collateral ligaments are labelled A, P, M, and L, while the curve labelled S represents the situation in which all ligaments are intact. The curves illustrate load-sharing between the ligaments. For anterior tibial displacement, the model ACL and MCL share the load, For posterior tibial displacement, the model PCL and LCL share the load at 0° and 25° flexion, while the PCL resists all the load at 90° flexion. Thus, the ACL and PCL are the primary restraints to anterior and posterior tibial displacement, respectively. The MCL and LCL are secondary restraints.

PIZIALI et al.[34, 35] measured all applied forces and moments necessary to achieve pure a-p tibial translation in cadaveric knee specimens at full extension. Apart from the a-p force, they found that the coupled loads during a-p displacement were almost all quite small and were mainly directed medio-laterally. The dashed lines in figure 12 A are
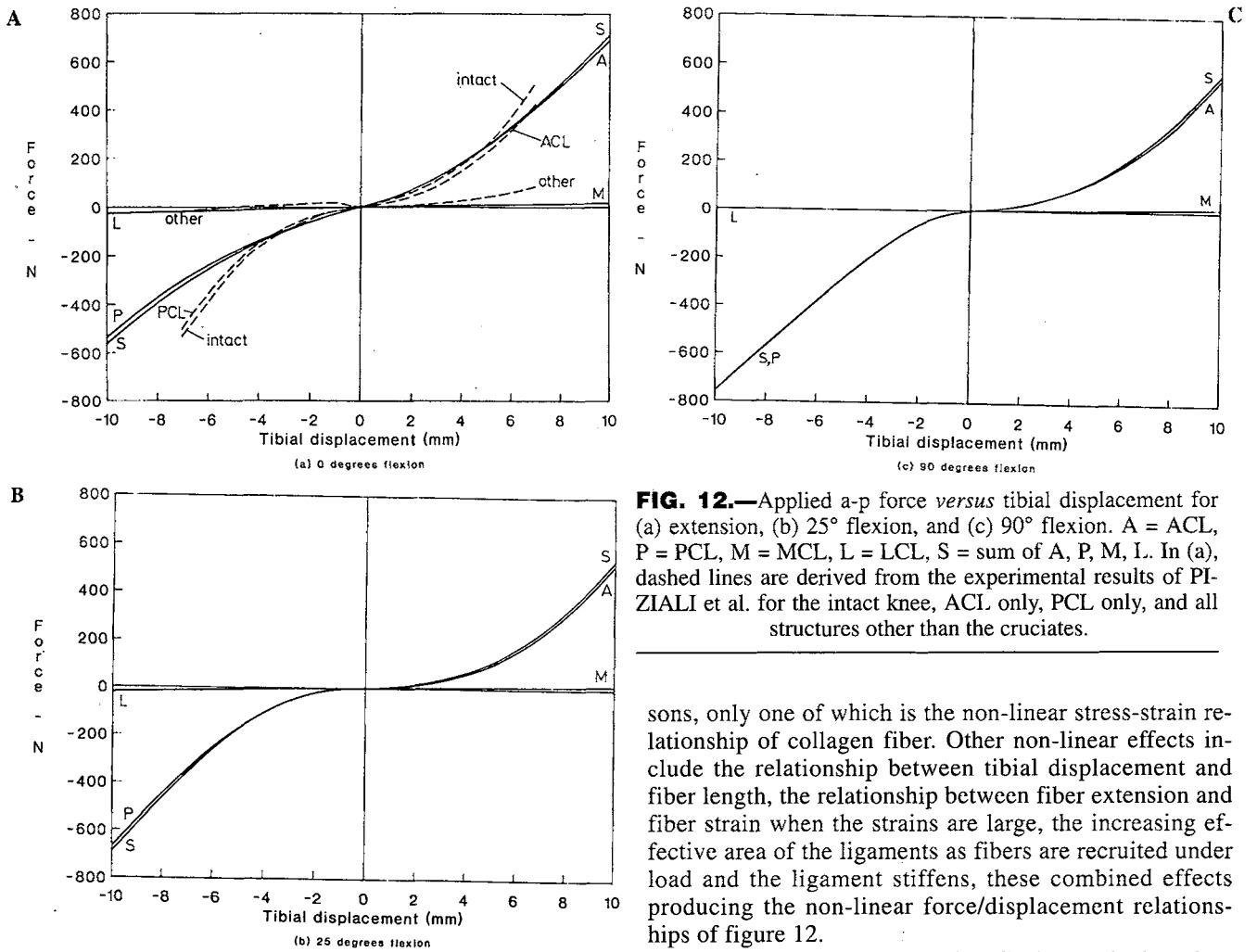
A

(a) 0 degrees flexion

B

(b) 25 degrees flexion

C

(c) 90 degrees flexion

FIG. 12.—Applied a-p force *versus* tibial displacement for (a) extension, (b) 25° flexion, and (c) 90° flexion. A = ACL, P = PCL, M = MCL, L = LCL, S = sum of A, P, M, L. In (a), dashed lines are derived from the experimental results of PI-ZIALI et al. for the intact knee, ACL only, PCL only, and all structures other than the cruciates.

derived from PIZIALI's load/displacement data[35]. The load sharing among the model ligaments is consistent with that of PIZIALI; however, the stiffnesses are not the same. The intact model knee is only slightly stiffer for anterior displacement than the experimental knees, but the model intact knee is less stiff than the experimental knees for posterior translation. When the cruciates are omitted from the calculation, simulating cruciate deficiency, the stiffness of the residual collateral structures is less than that of the experimental knees in anterior translation and very similar to the experimental knees for posterior translation. Inclusion in the model of the posterior oblique ligament, whose fibers are oriented to resist posterior tibial translation, or use of a higher value for the YOUNG's modulus would give a better match with PIZIALI's data for the posterior translations. Use of the non-linear stress-strain relationship shown in figure 13 A leads to a somewhat better fit with the experimental data, at least in this isolated case, as illustrated in figure 13 B. The calculated results, however, are not fundamentally changed. The non-linear relationship between anteroposterior force and anteroposterior displacement of figure 13 arises for a number of rea-

sons, only one of which is the non-linear stress-strain relationship of collagen fiber. Other non-linear effects include the relationship between tibial displacement and fiber length, the relationship between fiber extension and fiber strain when the strains are large, the increasing effective area of the ligaments as fibers are recruited under load and the ligament stiffens, these combined effects producing the non-linear force/displacement relationships of figure 12.

Figure 14 shows plotted against flexion angle the calculated total a-p tibial translation resulting from forces of 67 N, applied first anteriorly and then posteriorly for the model knee with both cruciates and collaterals intact. Translation is greatest at 35° flexion, where the ACL is both slack and poorly oriented to resist loads, and where the PCL is slack. Translation is least when one of the ligaments, although poorly oriented, is completely taut (ACL at extension, PCL at full flexion). Data from the work of GROOD and NOYES[22] are included in figure 14. The model adequately represents the general trends in a-p knee laxity, but it overestimates experimental translations at extension and from 45°-120° flexion. A better fit near extension would be obtained if the posterior capsule and posterior oblique MCL fibers were included in the model. The discrepancies at higher flexion angles are the result of a greated slackness in the model ACL fibers than in the human ACL fibers.

Model ACL force, expressed as a proportion of the applied anterior force, is plotted versus flexion angle in figure 15. Included is a curve representing ligament force per unit anterior load based on the orientation of the inextensible neutral fiber of the four-bar linkage (rigid model). Curves are also drawn for anterior tibial displacements of 2-10 mm. The force ratio decreases with flexion as the
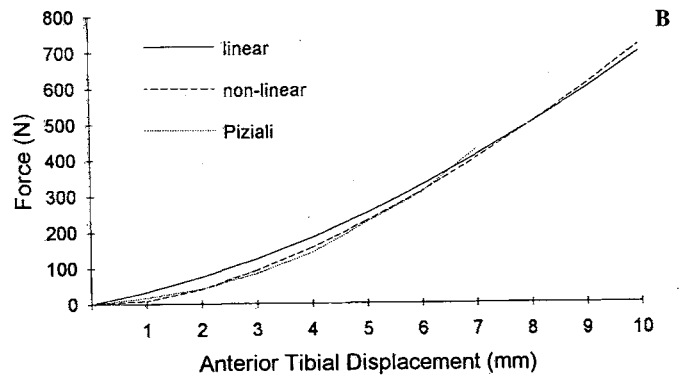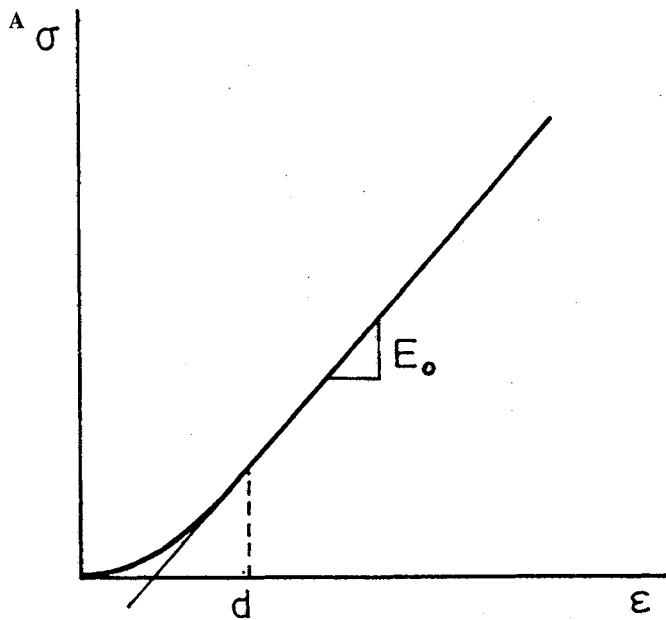
FIG. 13.—A: Non-linear stress/strain relationship used to calculate fiber stress. The curved region at the toe of the graph is a parabola, tangential at a strain of $d = 0.03$ to a straight region where the slope is $E_o = 9000$ N/cm^2. B: Applied anterior force *versus* tibial displacement at 0° flexion for the model ACL. Comparison of calculations using linear and non-linear stress/strain relationships with the experimental results of PI-ZIALI et al.

ACL becomes more horizontal and better oriented to resist the applied load. At any flexion angle, this effect is further accentuated when ligament fibers stretch, as indicated by the reduction in ligament force per unit load with increasing tibial translation.

Discussion

We have described the mechanisms whereby the ACL guides the movements of the femur on the tibia during flexion/extension and resists movement away from the positions dictated by those mechanisms. The apparently separate functions of the ligament, limiting movements of the bones and guiding movements within those limits, are in fact interrelated. Our models simplify the description of those relationships but contain the essential elements.

Treating the ligaments as inextensible straight lines explains the rolling and sliding movements of the femur on the tibia during flexion and extension and the relationships between ligament and articular surface geometry. Elastic ligament theory illustrates the strain patterns within ligaments during passive movements and progressive fiber recruitment under load. The simple two-dimensional model shows how geometric compatibility can be used to calcula-
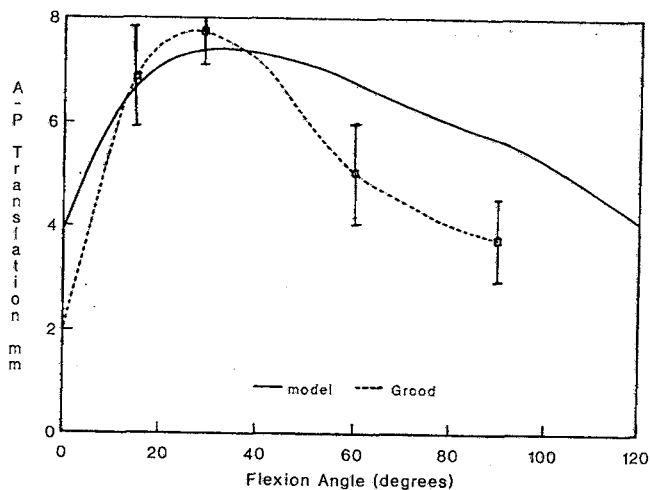

FIG. 14.—Total a-p tibial translation for applied a-p forces of 67 N plotted *versus* flexion angle for the model knee with both cruciates and collaterals intact. The dotted line is derived from the experiments of GROOD and NOYES figure 9-9.
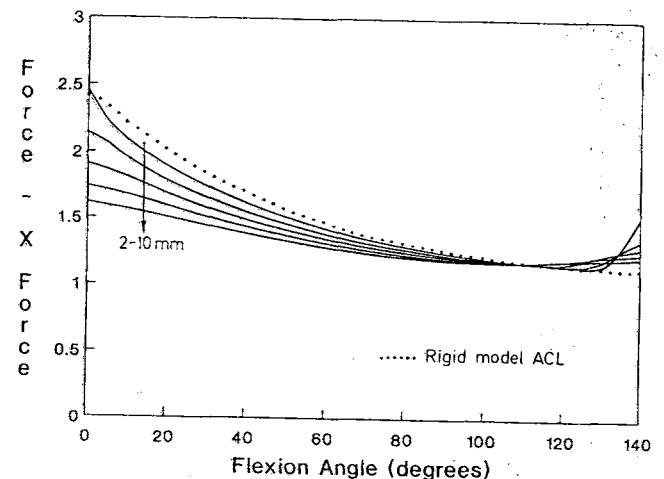

FIG. 15.—Model ACL force expressed as a proportion of applied anterior force (the horizontal component of total ligament force) *versus* flexion angle for 2-10 mm anterior tibial translation. For the dotted line, the ACL force direction is given by the ACL neutral fiber, calculated from four-bar linkage theory.

te the distributions of strain in all the ligaments of the joint at any position after specifying the sites of the neutral fibers within each cruciate ligament, the attachment lines of ligament fibers on the bones together with the mapping of fibers between these attachment lines, and the zero-tension reference position of all but the neutral fibers. The detailed results depend on the choices of all these parameters and a complete parametric analysis represents a formidable task, even for the two-dimensional model. The alternative is direct measurement of strain distributions within the human ligaments, itself a difficult task, even *in vitro*.

Limitations of the model

The theoretical models described here are capable of much further refinement and elaboration, but they contain most of the important elements. Quantification of their geometric aspects requires a knowledge only of high-school trigonometry. A three-dimensional model which requires more complex mapping functions for the fibers but which will calculate fiber strains compatible with medio-lateral translation, abduction, adduction, and long-axis tibial rotation, as well as a-p translation, is now being developed. In addition, the present analysis does not account for possible intra and inter-ligament variations in material properties[9, 10, 40] or cross-linking and other interactions between ligament fibers.

Clinical significance

The choice of model parameters necessary to achieve reasonable agreement between theory and observation defines the quantities which control the changing strain patterns within ligaments during passive flexion. These include identification of ligament bony attachment areas, choice of fiber mappings between corresponding areas on the two bones, localization of the neural fibers of the cruciate linkage within the finite cruciate ligaments, and specification of the zero-tension reference positions for all other fibers. When designing and implanting a ligament graft or prosthesis with the aim of reproducing the natural strain patterns, it is necessary not only to implant through the natural attachment areas but also to reproduce and maintain the natural mapping of the ligament fibers. If this could be achieved, it would then be necessary to render all fibers just tight at the appropriate zero-tension reference position. New prostheses which emulate these aspects of ligament structure may improve the rate of success of ligament replacement surgery.

Acknowledgements

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