Anatomy and biomechanics of the anterior cruciate ligament : a three-dimensional problem

Anatomy and Biomechanics of the Anterior Cruciate Ligament: A Three-Dimensional Problem

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The cruciate ligaments perform the contrasting functions of permitting motion of the articular surfaces on the one hand, and restraining their motion on the other by offering resistance to certain forces. The excessive restraint of mobility leads to functional disability and unphysiologic loading of the ligaments, whereas deficient restraint leads to instability. The anterior cruciate ligament (ACL) plays a critical role in the performance of this task. This role is determined entirely by the anatomic configuration of the ligament attachments and the mechanical properties of the ligament itself. In turn, the mechanical properties of the ACL depend on its three-dimensional collagenous structure. These interdependencies are of clinical importance. In knee laxity tests, for example, an attempt is made to assess the function of the ACL and diagnose the severity of lesions on the basis of observed or elicited joint motion. In reconstructions of the ACL, an attempt is made to repair the lesion to the degree that normal motion restraint is reestablished. In both cases the geometry of the ACL attachments and the mechanical properties of the ligament affect the overall mechanical behavior of the knee joint in a manner that can be utilized diagnostically and therapeutically. However, the situation is complicated by the fact that the ACL does not exert this effect in isolation but in concert with the articular geometry and the other ligaments of the knee. Thus, the relationship between the morphology and function of the ACL operates within a complex system of structures that influence one another.

For the scientist faced with a system as complex as the knee, the primary task is to create order out of apparent chaos. One way to do this is to develop a "model," i.e., a simplified description or reflection of a complex reality which makes certain aspects of this reality comprehensible and manageable. A familiar example is the "four-bar linkage" (Fig. 1), which models the cruciate ligaments as rigid links having mobile attachments to the femur and tibia (Strasser 1917; Menschik 1974; Huson 1974; Müller 1982). Basic assumptions in this model are that the cruciate ligaments experience little or no strain during knee motion and that they are represented mechanically by linear elements in one plane. The correctness of these assumptions is less important than the usefulness of the model, i.e., whether the model enhances our understanding of the relationship between form and function, and whether it offers an explanation for the phenomena observed. And indeed, the model accomplishes this reasonably well (Müller 1982). It explains the shape of the articular surfaces (at least in reasonable approximation on the sagittal plane), and it explains why the rotational axis for knee flexion, which the model represents as the point where the bars intersect, is not rigidly attached to the tibia or femur. It also offers an explanation for the resistance

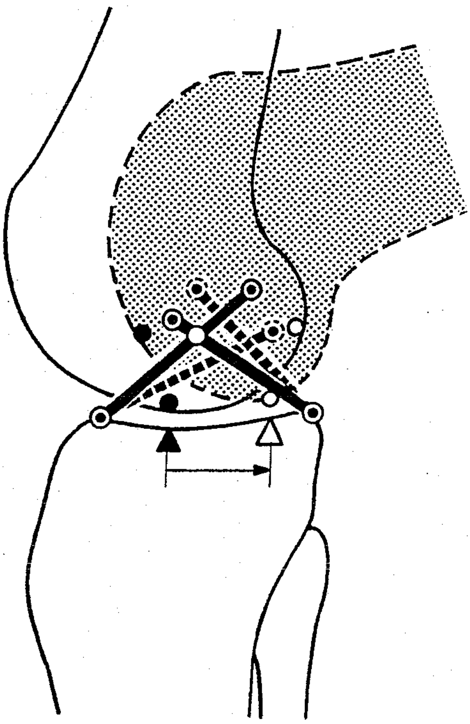


Fig. 1. Model of the knee as a four-bar linkage in the sagittal plane. The anterior and PCLs are represented by rigid bars that are connected to the tibia and femur by hinges. In all degrees of flexion, the instant center of rotation coincides with the point of intersection of the cruciate ligament bars. The constraining action of the cruciate ligaments causes the femur to roll backward on the tibia with increasing flexion. (From Müller 1982)

that the anterior and posterior cruciate ligaments offer especially to anterior and posterior drawer motion, and it explains the posterior translation of the femur relative to the tibia during flexion. However, the model does not describe the phenomenon of knee laxity, the role of the collateral ligaments, or internal-external rotation of the knee in the transverse plane. If we are to comprehend and describe the functional role of the ACL in greater detail, we must approach it as a three-dimensional problem. This requires that we first investigate the physiologic mobility of the knee joint in three dimensions.

Freedom of Motion of the Knee

The passive freedom of motion of the knee is investigated by mounting amputated or cadaveric knee joint specimens in an apparatus that permits the joint to be moved incrementally through designated flexion angles while leaving other tibial motions with respect to the femur unconstrained (Fig. 2) (Blankevoort et al. 1988). This apparatus can exert axial forces, AP forces, or torques (torsional moments) upon the joint (Fig. 3). The three-dimensional motions that are elicited by the applied load are analyzed using techniques of roentgen stereophotogrammetric analysis (RSA) (Selvik 1974; Huiskes et al. 1985b; de Lange et al. 1985; Blankevoort et al. 1988). For this purpose the bones are marked with radiopaque tantalum pellets, and radiographs are taken from two directions in each joint position (Fig. 4). Digitized and computer analyzed, these films can yield very precise data on the position of the tibia with respect to the femur in relation to a given starting position (unloaded extension). Multiple sequential positions serve to simulate a joint movement.

These experiments (Blankevoort et al. 1988) have shown that joint motion in one plane or in one direction is almost invariably associated with motion in another plane or direction. That is, translational motions in the three anatomic directions (proximal-distal, anterior-posterior, medial-lateral) tend to be coupled with rotational motions in the three anatomic planes (sagittal, transverse, frontal). The experiments further show that the degree of coupling depends greatly on the external load. An example is shown in Fig. 5. When the joint is flexed in the unloaded condition, practically no tibial rotation occurs. But if a (small) axial force is applied to the joint, internal tibial rotation does occur during flexion (Fig. 5a). A (small) anterior force on the tibia likewise leads to internal rotation during flexion, while a posterior force leads to external rotation (Fig. 5b). Small internal and external torques exerted on the tibia lead to internal and external rotation, respectively, during knee flexion (Fig. 5c).

This load dependence of the kinematic coupling between flexion and tibial rotation is caused by an extremely low ligamentous restraint to rotation (Fig. 6). We can conceptualize this lack of restraint as a freedom of rotatory motion whose excursion limits are defined by threshold torques of, say, +/−3 Nm (Fig. 6). The knee, then, can be modeled as a mechanism with two degrees of freedom, flexion and tibial rotation, within the limits defined by internal and external torques of 3 Nm (see Fig. 5c).

This (qualitative) kinematic model provides an explanation for the observed sensitivity to external

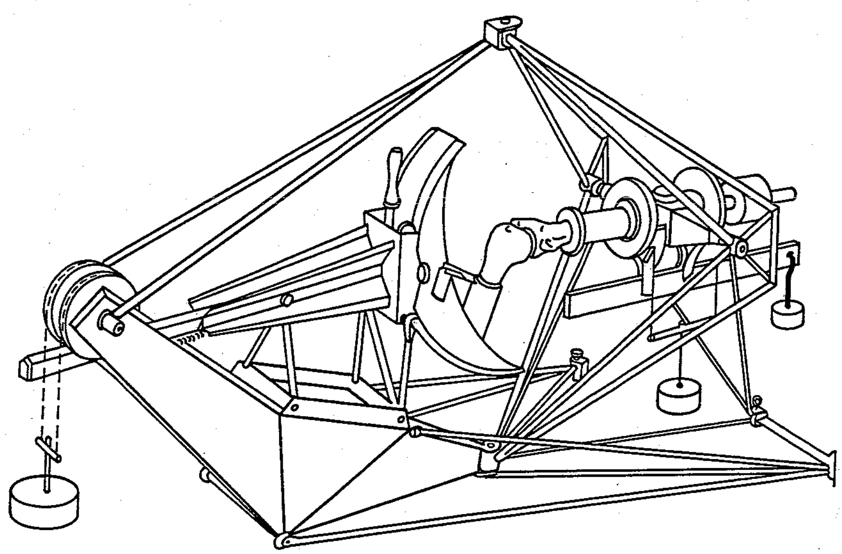


Fig. 2. Motion and loading apparatus for the quasistatic kinematic testing of the knee joint, as used by Blankevoort et al. (1988). (From Blankevoort et al. 1988)

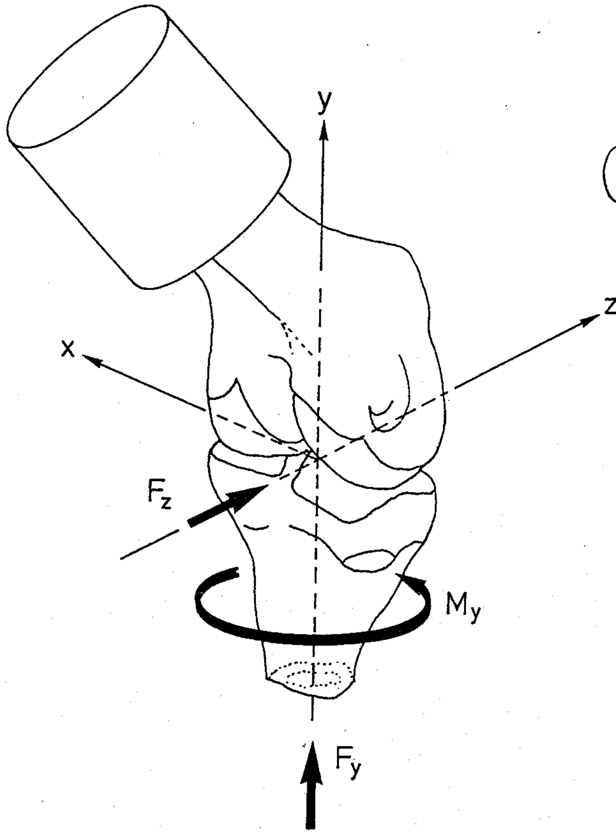


Fig. 3. Schematic representation of the loads that can be applied to a knee specimen with the motion and loading apparatus (see Fig. 2). The coordinate system is fixed to the tibia, with F_z representing the a.p. force, F_y the axial force, and M_y the torque. (From Blankevoort et al. 1988)

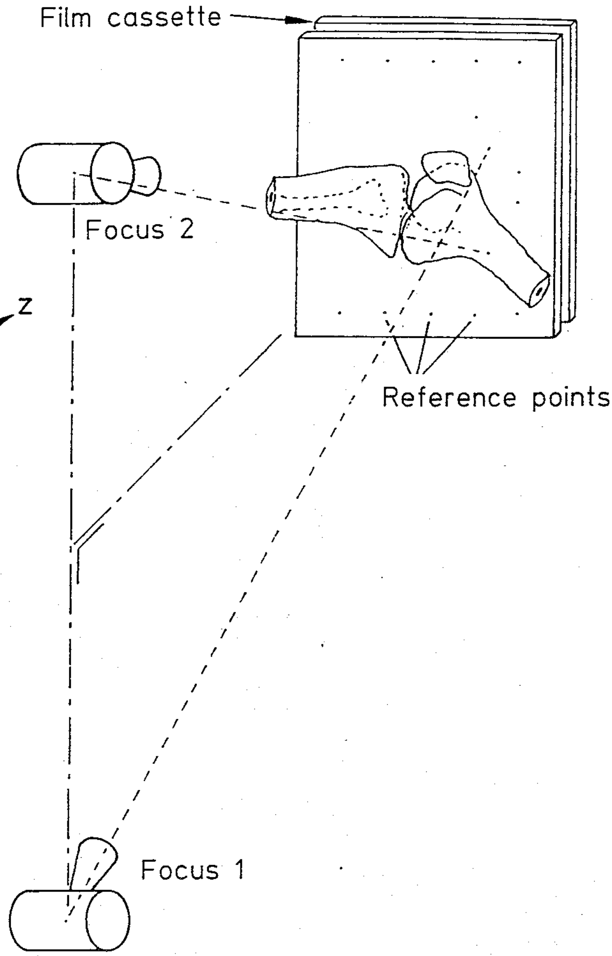


Fig. 4. Schematic representation of the experimental setup for roentgen stereophotogrammetric analysis (RSA). The knee specimen is positioned in front of a reference plate and X-ray cassette. In each joint position the specimen, with its affixed tantalum markers, is X-rayed by two roentgen tubes, and a calibration process is used to determine the spatial position of the X-ray focal spots and the spatial relationship between the tantalum reference markers and the laboratory coordinate system. The spatial positions of the markers can be reconstructed following digitization of the films. A series of successive joint positions simulate a movement, whose kinematic parameters can be determined from the positions of the markers. (From Blankevoort et al. 1988)

loads (Fig. 5 a,b): An axial force produces a small, internally directed torque that results from the shape of the articular surfaces, and which causes the tibia to rotate toward the internal limit of rotation (screw motion). Forces in the a. p. and p. a. direction likewise produce small torques that result in internal or external rotation. Torques are also exerted by the quadriceps muscle group during walking and running (Fig. 7), and this, combined with the axial force of the foot strike, produces internal rotation. As the envelope of rotatory motion becomes smaller with decreasing flexion, the joint approaches its internal limits (Fig. 7), ultimately resulting in a combination of extension and external rotation. This "screw home" phenomenon, then, is not just caused by passive joint properties but occurs exclusively in conjunction with an external load. The phenomenon does not occur during extension of the unloaded joint (see Fig. 5 a).

Thus, the three-dimensional kinematic model of the knee as a mechanism with two degrees of freedom implies that the actual motion pathway within the envelope of motion depends on the external load. This further implies that the joint axes are not uniquely defined. As Fig. 8 demonstrates, pure flexion without tibial rotation occurs about a horizontal axis, while pure rotation without flexion occurs about a vertical axis. If the actual motion is a combination of flexion and rotation, its axis will be oblique to the horizontal and frontal planes.

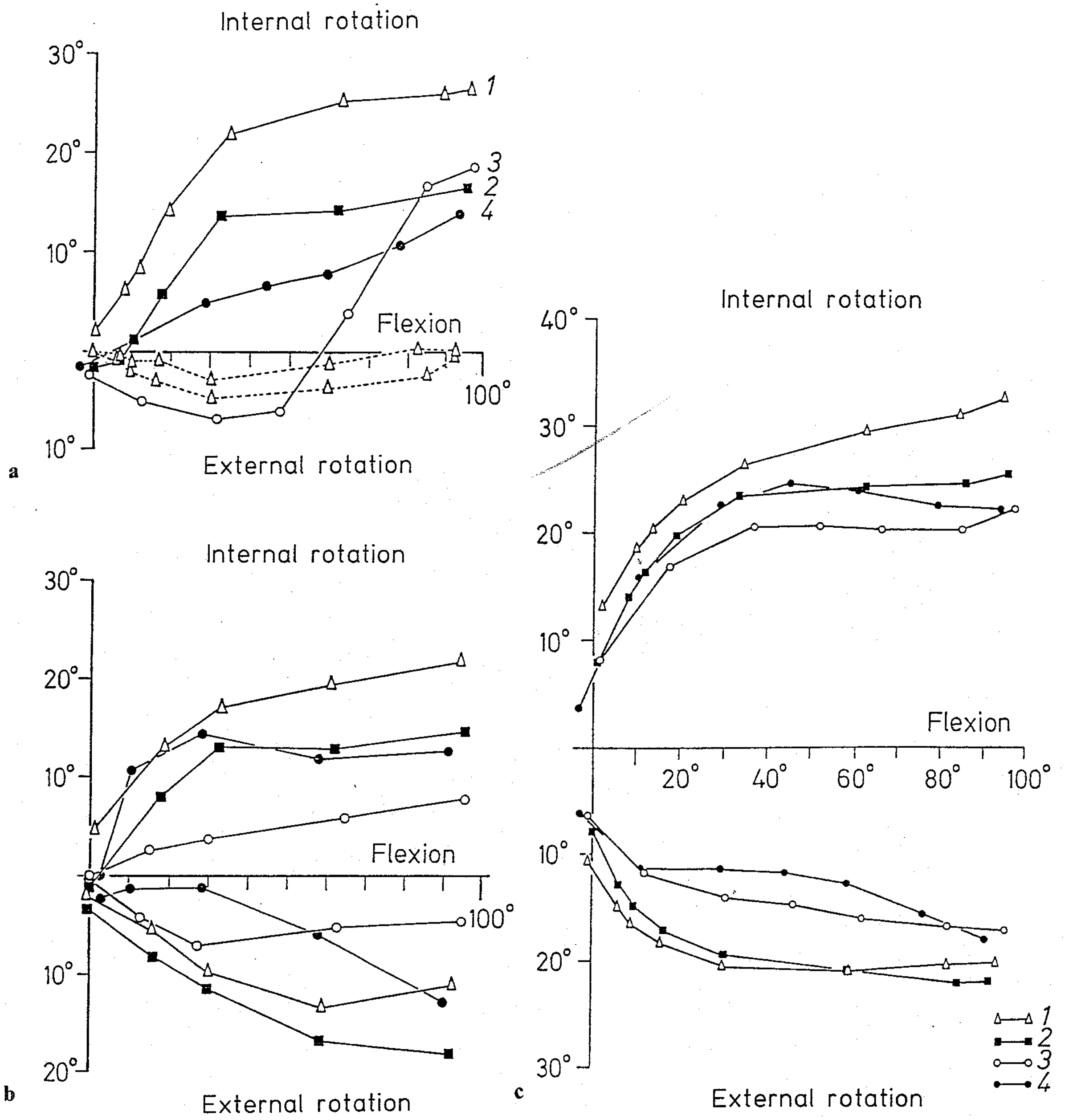


Fig. 5a–c. Tibial rotation versus flexion angle for various external loads in 4 knee joint specimens. **a** *Dashed lines:* 2 motion sequences in unloaded specimens. (The other 3 specimens showed similar responses.) *Solid lines:* motion pattern in 4 different specimens with an imposed axial load F_y of 300 N. **b** The motion pattern with an anterior load F_z of 30 N shows internal tibial rotation during flexion, while the motion pattern with a posterior load F_z of -30 N shows external rotation (4 different specimens). **c** Internal rotation occurs with a torque M_y of 3 Nm, external rotation with a torque M_y of -3 Nm (4 different specimens). If the knee is viewed as having 2 degrees of freedom, flexion and tibial rotation, and the excursion limit of tibial rotation is defined as +/− 3 Nm, then the inner and outer curves describe the envelope of knee motion. (From Blankevoort et al. 1988)

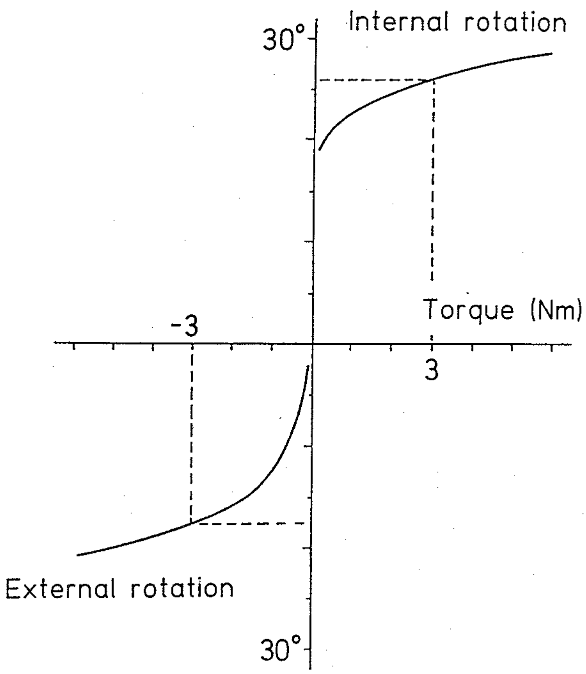


Fig. 6. Internal and external rotation versus torque at a fixed flexion angle of 25° (result for a representative specimen). Threshold torques of +/−3 Nm are arbitrarily defined as the limits of excursion for tibial rotation. (From Blankevoort et al. 1988)

The accuracy of the RSA method makes it possible to calculate the joint axes for specific flexion movements (Blankevoort et al. 1990). One result of this approach is shown in Fig. 9, where the internal and external limits of the motion are defined in terms of +/−3 Nm of torque. We find that the axes of the motion pathways are indeed different for both of these loads. The initially oblique course of the axes in the frontal plane reflects the strong internal or external rotation occurring for that pathway. But the final axes (at approximately 40°–90° of flexion) are almost horizontal, indicating that the motion consists of almost pure flexion. The oblique course of the axes in the transverse plane means that flexion and tibial rotation are coupled with some degree of valgus or varus rotation. The visible posterior displacement of the axis in this plane is comparable to the corresponding posterior shift of the instant center in the anatomic four-bar model and the "roll-back" of the femur upon the tibia. On examining the results, we are struck by the fact that over both (outermost) motion pathways along the limits of the motion envelope, the axes always pass through the region between the femoral attachments of the cruciate ligaments. Thus, while they do not pass through the intersection of "bars" as in the two-dimensional anatomic four-bar linkage model, they still appear to be determined largely by the configuration of the cruciate ligaments.

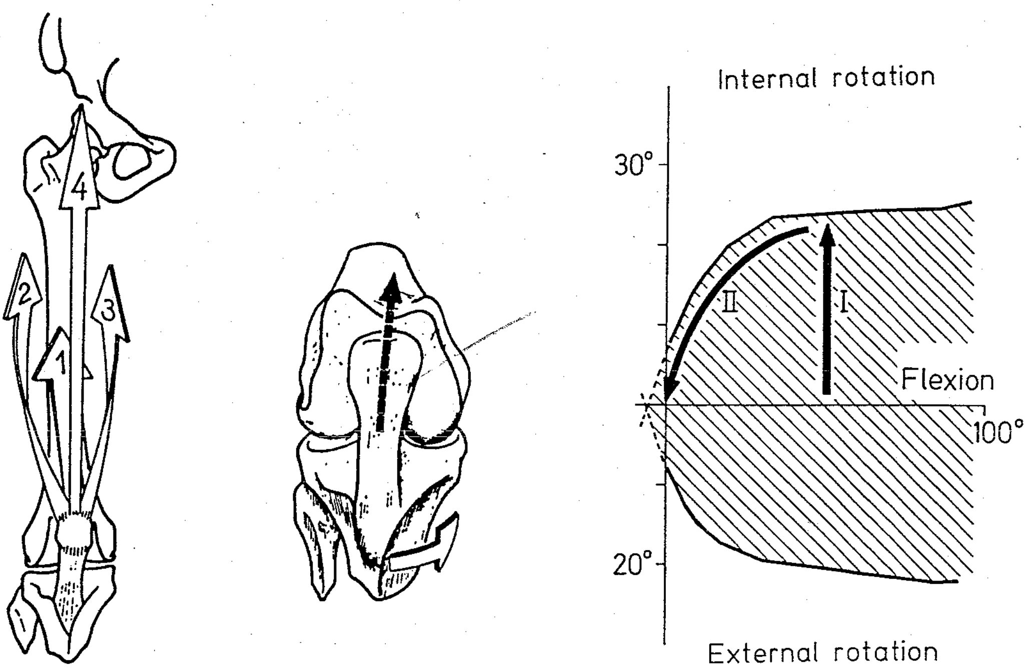


Fig. 7. The quadriceps muscles exert a small internal torque on the tibia, resulting in internal rotation (I). When the knee is extended, quadriceps contraction directs the motion pathway along the internal limits of the envelope of motion. As the envelope of motion dwindles, the knee is forced into external rotation (II). This can account for the "screw home" phenomenon, or compulsory external rotation of the tibia in terminal extension

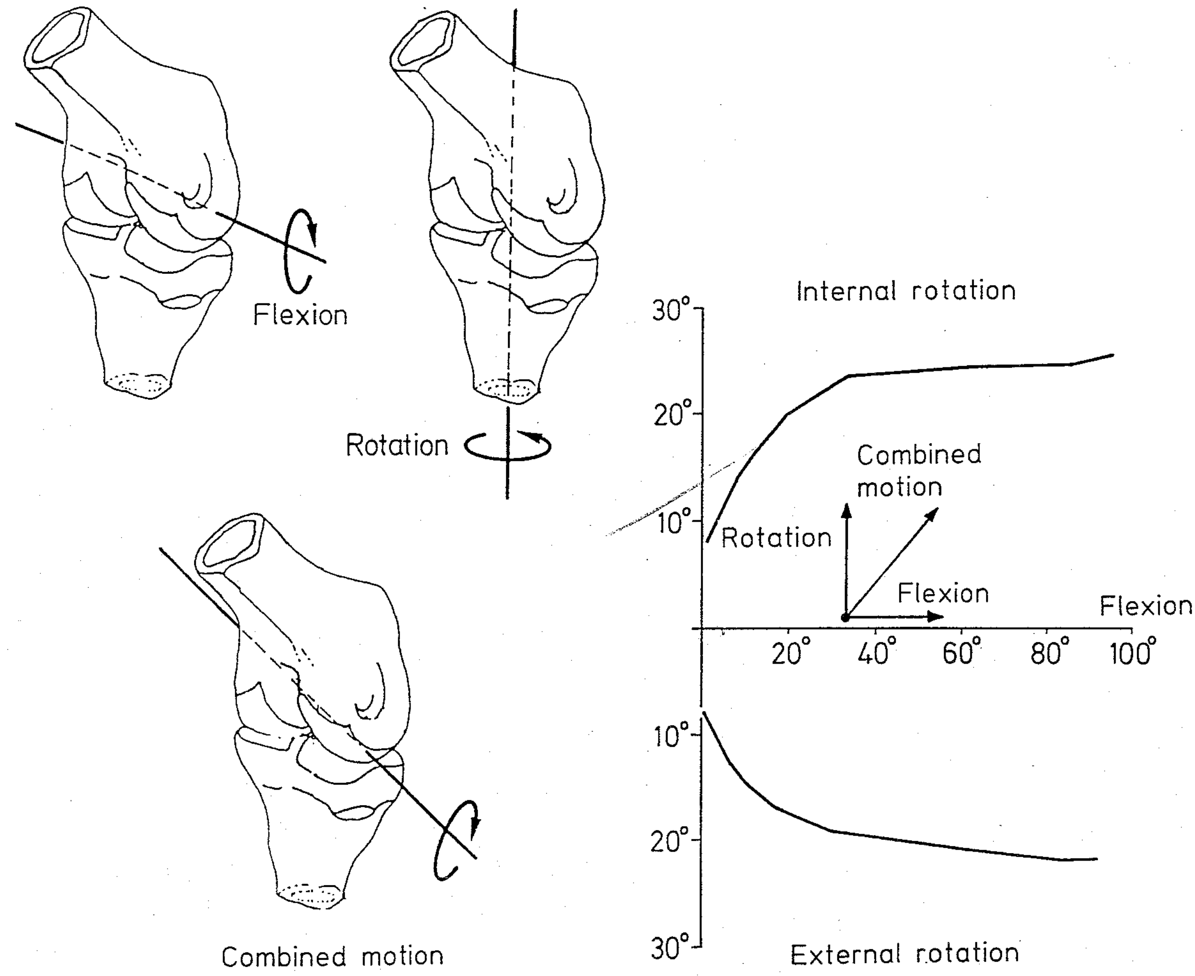


Fig. 8. Schematic representation of the axes of knee motion (helical axes) *(left)* for various motions within the envelope of knee motion *(right)*. The instantaneous axes are not uniquely defined. Pure flexion generally occurs about horizontally directed axes, and pure tibial rotation about vertically directed axes. Coupled flexion and rotation have axes directed obliquely to the anatomic planes

In summary, we can regard the knee joint as a mechanism with two independent degrees of freedom, flexion and tibial rotation, within a limited envelope of motion. During each combination of motions and along each motion pathway within this envelope, small translations (in 3 directions) coupled with flexion and rotation occur in addition to varus-valgus rotation. For motions along the limits of the envelope of motion, flexion likewise is coupled with tibial rotation. This is the coupling for which the ligaments, acting in concert with articular geometry, perform their specific task. By extending to the limits of the envelope of motion, the ligaments can exert forces that offer resistance to external loads and constrain further motion. The coupling, then, results from an internal balance of forces to which each ligament makes its own contribution. To isolate and analyze the role of a single ligament in this mechanism, force analyses must be performed. Clearly, our results show that we are dealing with an essentially three-dimensional problem that is made even more complex by the irregular geometric configuration of the anatomic structures.

A schematic representation of this problem is shown in Fig. 10. Knee motion occurs as the result of a combination of internal and external forces. The motion strains the ligaments, stretching them outside the envelope of motion. Depending on the mechanical and geometric properties of the ligaments, this strain gives rise to forces that in turn contribute directly or via the articular surfaces to the internal forces, and thus to the motions of the joint members. This means that we are dealing with a feedback control mechanism. Before proceeding to analyze this system in its

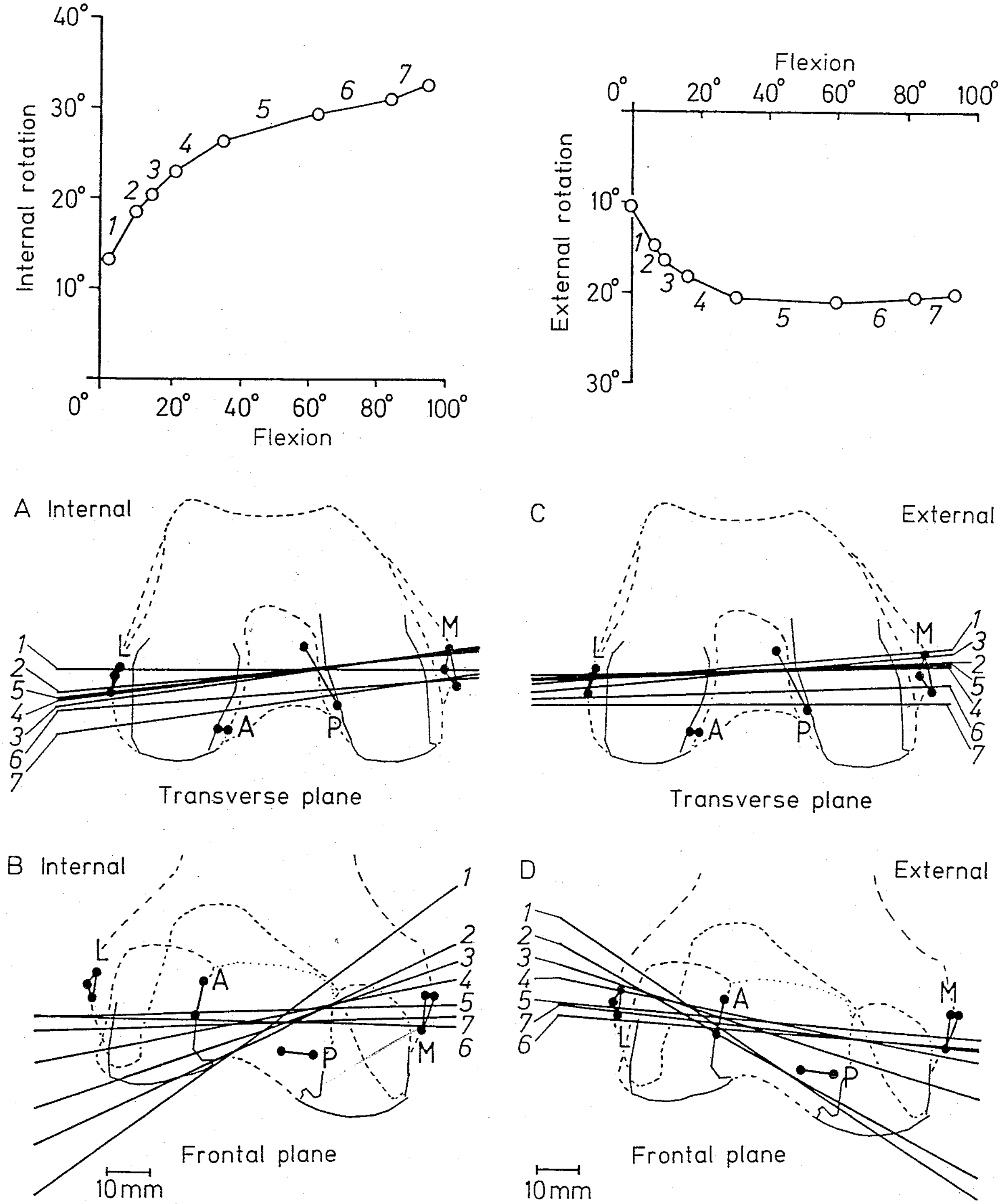


Fig. 9. Axes (helical axes) for finite motion steps along the limits of internal rotation *(left column)* and along the limit of external rotation *(right column)*. Each axis is shown projected onto two anatomic planes, the transverse and frontal, together with markers indicating anatomic structures, articular geometry, and ligament attachments (*A* ACL, *P* PCL, *L* lateral collateral ligament, *M* medial collateral ligament). Each axis is numbered according to the motion step in the graph

entirety, especially in terms of the role of the ACL, we shall first investigate the relationship between knee motion, typical ligament length changes, and ligament forces.

Mechanical Properties of the ACL

As Fig. 10 demonstrates, a ligament exerts its function by elongating. The force produced by this elongation depends on the mechanical properties of the ligament. These properties can be measured in vitro with a mechanical test rig (Noyes and Grood 1976; Butler et al. 1986; Woo et al. 1983; van Rens et al. 1986) that records displacement (strain) as a function of applied load. An example is shown in Fig. 11 (Meijer et al. 1987). In this experiment a bone-ligament-bone specimen was mounted such that the ligament was able to align in the direction of the applied force. The resulting stress-strain curve is nonlinear, especially in its initial portion where a very small force produces a relatively large elongation. In this region the ligament does not offer much resistance despite the relatively large strain. It is believed that this phenomenon relates to the unfolding and aligning of the collagen fibers (Butler et al. 1983). As loading proceeds, the strain resistance (the stiffness of the ligament) progressively increases until the curve becomes (almost) linear due to the increased number of taut collagen fibers and their elastic properties. As the load increases further, rupture begins to occur in isolated fibers and finally involves the whole ligament, represented by the "frayed" curve segment in Fig. 11.

It should be noted at this point that collagenous structures are viscoelastic in their behavior rather than purely elastic. One consequence of this is that the stress-strain curve is dependent on the loading rate. With rapid (impact) loading, the ligament exhibits greater stiffness than with a slowly incremental load.

As stated earlier, ligament stiffness and thus the amount of force exerted by the elongating ligament depends on the number of tense collagen fibers. This number is small in the initial stage (see first portion of curve in Fig. 11) but then gradually increases to the total maximum number present in the ligament, which in turn relates to the ligament thickness. The process whereby increasing numbers of fibers become tense is called "recruitment." It is essential to have a correct understanding of this process, because it determines the effect of a ligament, such as the ACL, on knee function for a given collagen density and given collagen properties. Recruitment is a complex process, especially in the ACL, which has a complicated three-dimensional structure in which the collagen fibers do not exhibit a strictly parallel arrangement. This is illustrated in Fig. 12, which shows the result of a new technique developed for the geometric measurement and numerical description of the three-dimensional collagenous structure of the knee ligaments (Meijer et al. 1989).

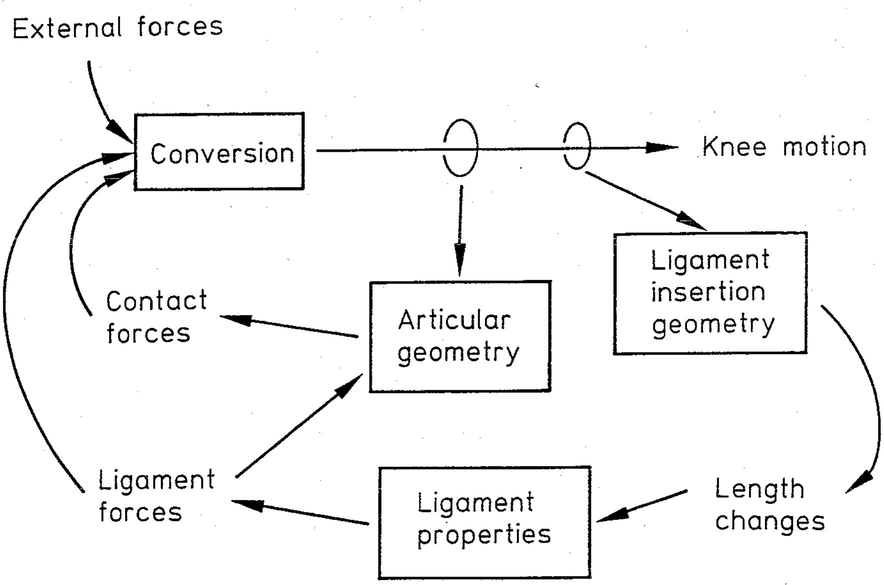


Fig. 10. Schematic representation of the mechanical interactions among the articular structures of the knee

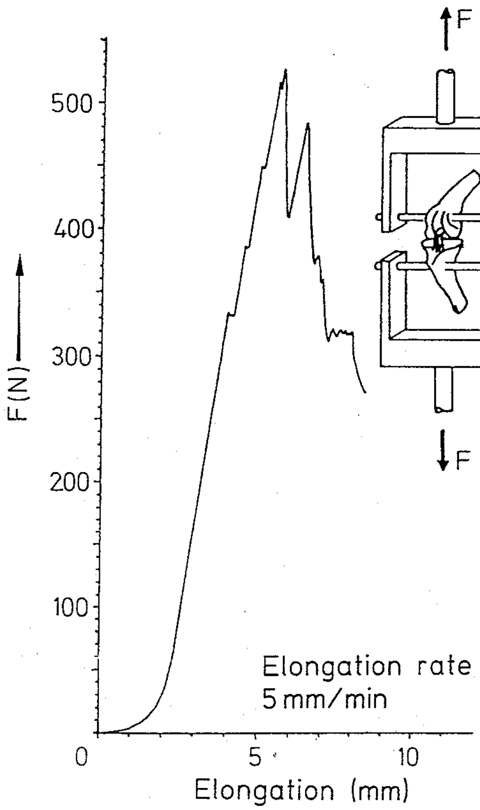


Fig. 11. Tensile test curve of an ACL in a canine knee (bone-ligament-bone specimen with drawing of experimental setup). The specimen was elongated at a prescribed rate, and the associated force was recorded. (From Meijer et al. 1987)

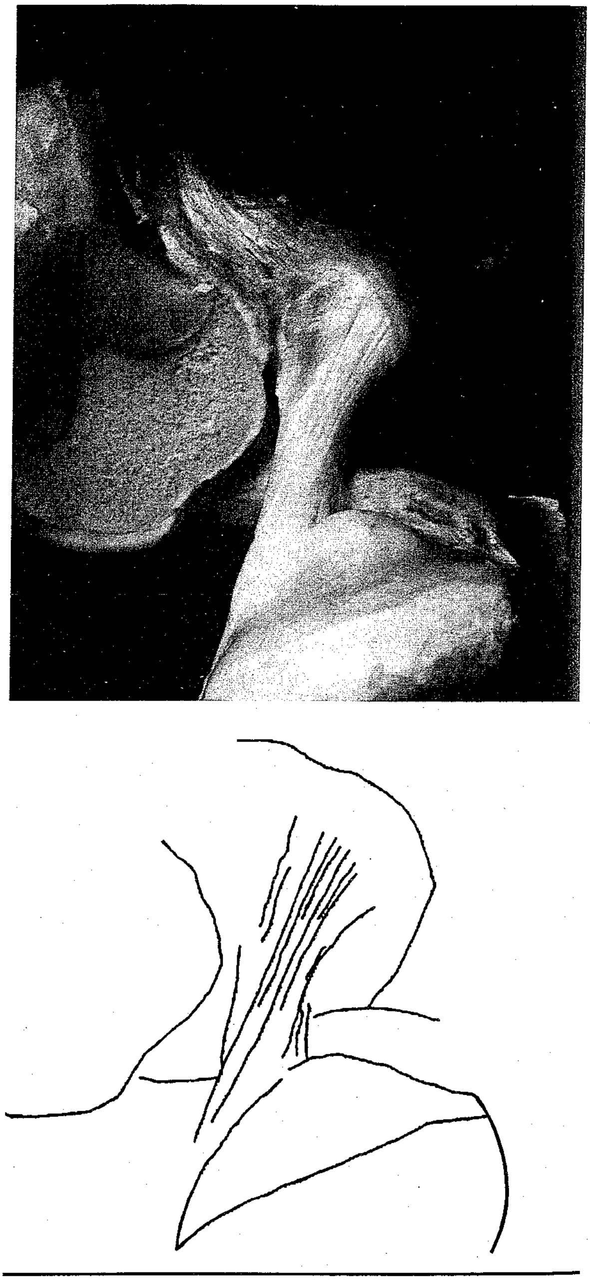


Fig. 12. Sample result of the technique of Meijer et al. (1988) for determining the three-dimensional fiber pattern of ligament bundles. Photographs taken from several directions are analyzed to reconstruct the three-dimensional course of the fiber bundles. *Above* is a photo from the series, from which the reconstruction *(below)* was prepared. (From Meijer et al. 1989)

Recruitment of the ACL

Recruitment is the process that determines the relationship between knee movements and typical ligamentous length changes (see Fig. 10), which in turn determine the forces that are developed and thus the restraining action of the ACL. To study this process, a technique was developed that permitted typical ligament length changes to be determined during motion measurements (van Dijk et al. 1979; van Dijk 1983; Blankevoort et al. 1991 a). Following the RSA experiments, the attachment sites of the cruciate ligaments were marked with small tantalum pellets (Fig. 13). The RSA technique was again used to measure these markers in the bone segments so that the distances between the individual pairs of markers could be reconstructed during the various motion experiments. In this case each of the cruciate ligaments was described by 2 lines (Fig. 13): the anterior portion of the ACL (A–B), the posterior portion of the ACL (C–D), and the anterior (E–F) and posterior (G–H) portions of the posterior cruciate ligament (PCL). Figure 14 shows an example of these mathematical reconstructions for 5 flexion angles in the frontal and sagittal projections (van Dijk 1983). These reconstructions convey a vivid impression of the changes in

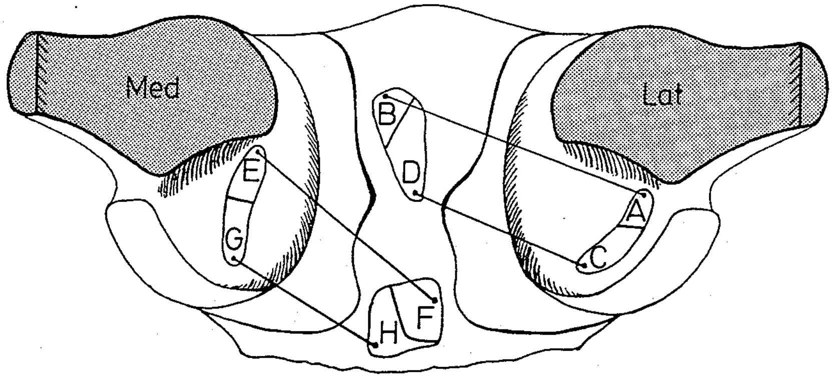


Fig. 13. Schematic representation of the technique for marking the cruciate ligament attachments to measure length changes in the cruciate ligament fiber bundles. Two fiber bundles in each cruciate ligament are identified, and their tibial and femoral attachments are marked with tantalum pellets. The spatial positions of the markers *(A–H)* are then determined using roentgen stereophotogrammetric analysis. With this technique, the distance between each pair of markers can be determined for each measured joint position (van Dijk 1983). See text for further explanation

Fig. 14. see p. 102

the configuration of these ligaments during knee flexion. The only interpretation of these findings is that each position of flexion is associated with the recruitment of different fiber bundles in the ligaments. Also, both projections demonstrate that the mechanical effects of the ligaments are not restricted to one plane. Figure 15 shows the associated length changes in the fiber bundles during flexion in relation to their length in the control state (0° flexion). We see that the anterior bundles of both cruciate ligaments elongate while the posterior bundles shorten. In terms of fiber recruitment, this means that the posterior portions of both ligaments make no functional contribution during flexion. This can also be demonstrated in anatomic specimens (Fig. 16a–c) (van Dijk 1983).

Besides flexion, tibial rotation also influences the recruitment of the fiber bundles. For each position of flexion, the ACL bundles were found to be elongated in internal rotation and shortened in external rotation. The same applies to the posterior bundle of the PCL, whereas the opposite is true for the anterior bundle (Blankevoort et al. 1991a). The length changes associated with tibial rotation, however, are not very large compared with the length changes in flexion.

Apparently the recruitment of the ACL fiber bundles within the envelope of motion depends chiefly on the position of joint flexion and is limited to the anterior portion of the ligament. It may be that the application of excessive anterior forces or internal torques would recruit more of the ligament fibers, but the degree to which this is true remains unknown.

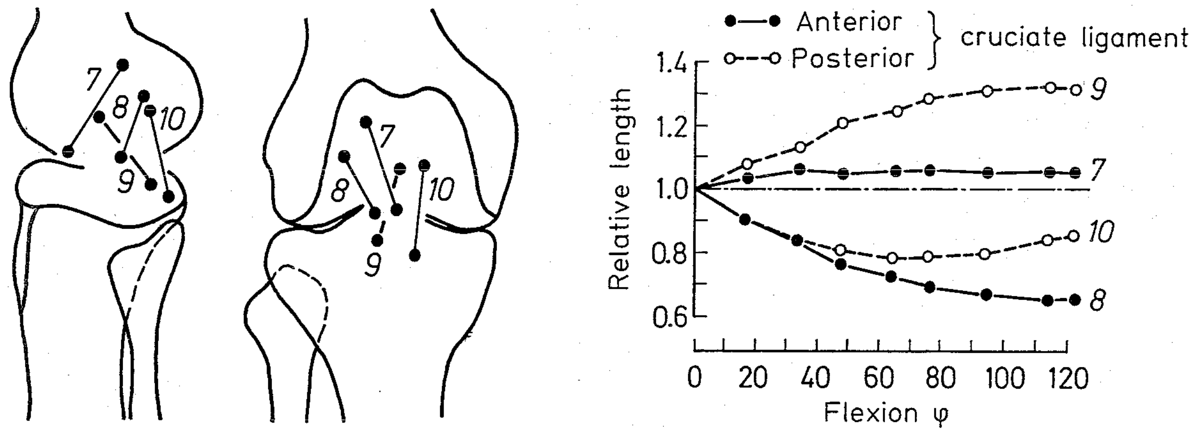


Fig. 15. Relative cruciate ligament length changes versus flexion angle in a single knee specimen. The distance between the ligament markers during flexion is related to the distance in the reference position of extension. *7*, Anterior bundle of ACL; *8*, posterior bundle of ACL; *9*, anterior bundle of PCL; *10*, posterior bundle of PCL. (From Huiskes et al. 1985b)

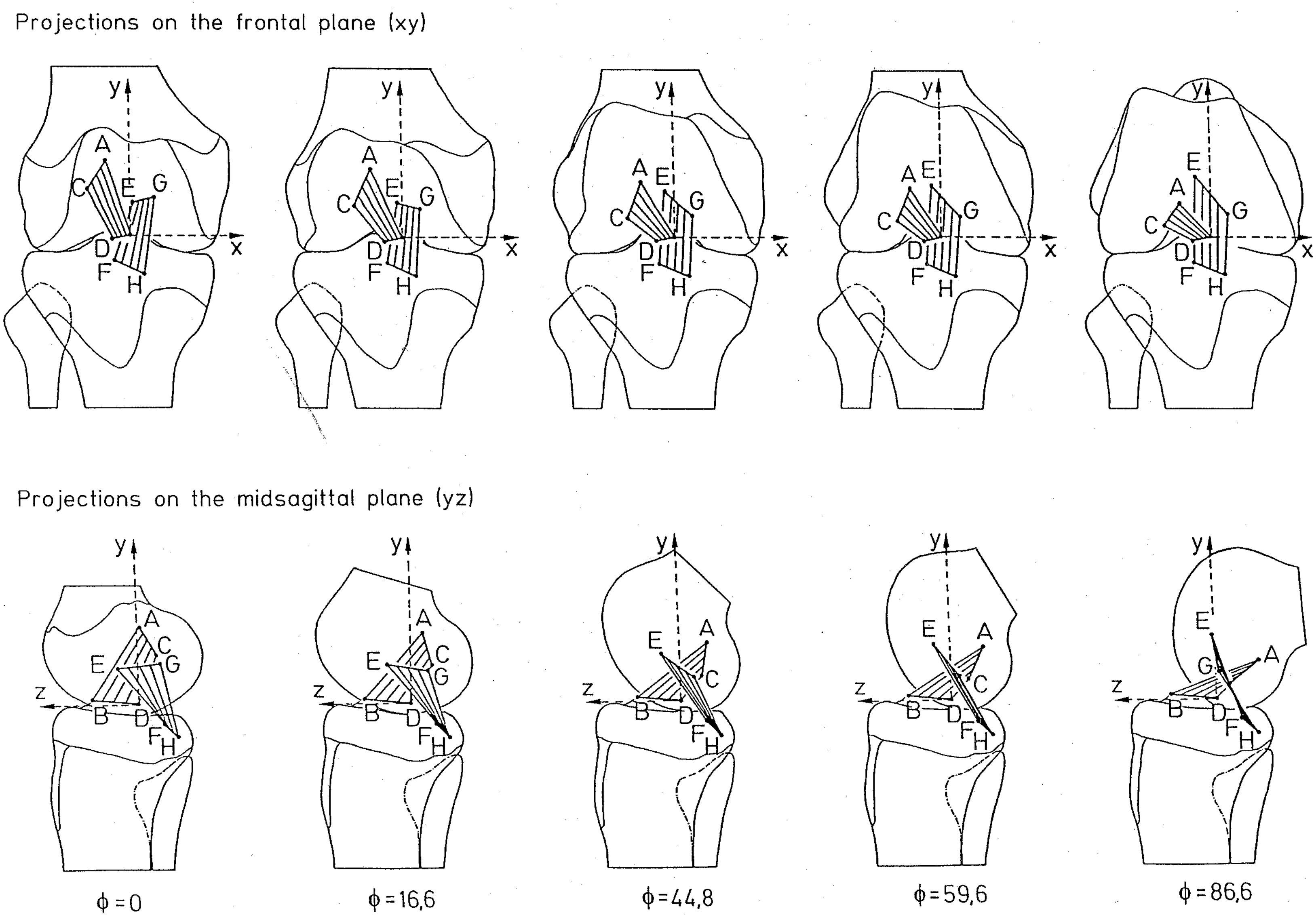
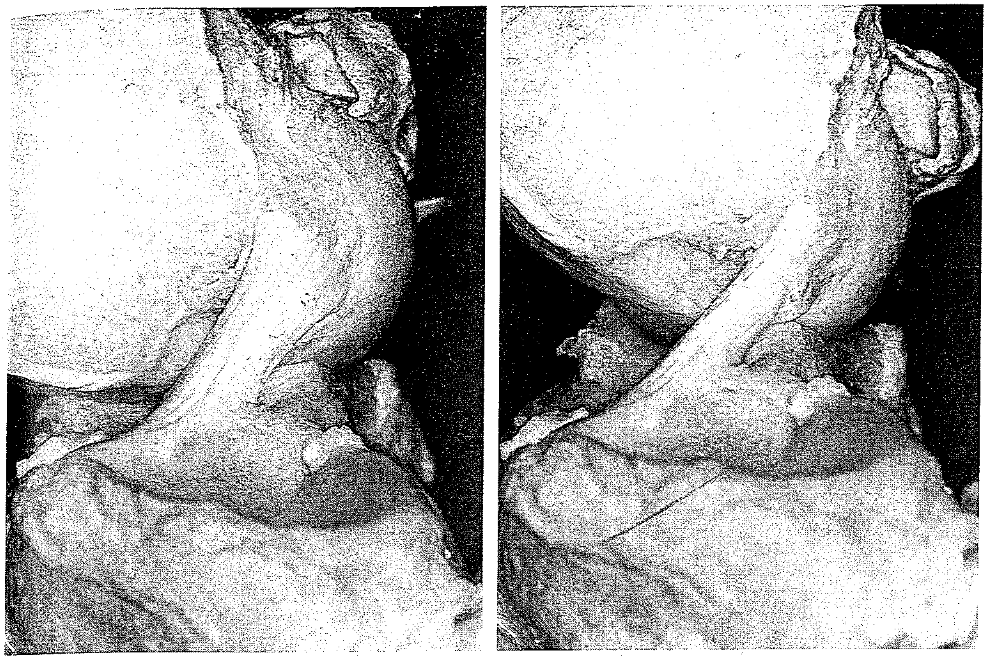


Fig. 14. Sample reconstructions of the spatial orientations of the cruciate ligaments for 5 knee positions, projected onto the frontal and sagittal planes. Right knee of a 16-year-old. φ = Flexion angle. (From van Dijk 1983)



a b

c

Fig. 16a–c. ACL, medial aspect. **a** Full extension. The posterior bundle is taut. **b** Moderate flexion. The posterior bundle is lax, as indicated by the folds in the posterior edge of the ligament. **c** In complete flexion the anterior bundle is very tense, while the femoral attachment of the posterior bundle is rotated in front of the attachment of the anterior bundle, and the posterior bundle is completely lax. (From van Dijk 1983)

Role of the ACL in the Play of Forces

The above technique for the investigation of fiber recruitment yields information on the portion of the ligament that can exert force in a given position, and it provides some indication of the direction of the force. The magnitude of this force remains unknown, however, because nothing is known about the relationship between force and length change in the recruited portion. Nor do we have definite information about the reference strain of the ligament. In other words, we can measure the *relative* length change but not the length change with respect to an absolute zero point (see Fig. 11). Thus, a force analysis is the only means available for understanding the system outlined in Fig. 10 and investigating the effect of the ACL on knee function. To perform a force analysis in a system as complex as the knee joint, it was necessary to develop computer simulation models (Wismans et al. 1980; Andriacchi et al. 1983; Essinger 1986; Blankevoort et al. 1991 b; Blankevoort and Huiskes 1991).
Computer simulation models essentially consist of systems of equations, based on mechanical laws, that incorporate the relevant properties of the individual joint structures. Some degree of schematization is required so that these properties can be mathematically described. The model used by us (Blankevoort et al. 1991 b; Blankevoort and Huiskes 1991) has a three-dimensional character and employs realistic repre-

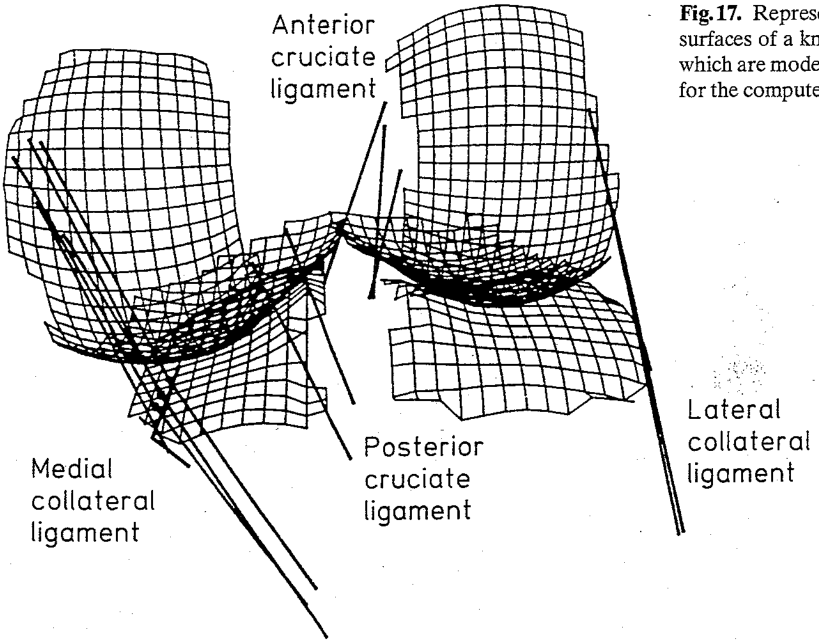


Fig. 17. Representation of the geometry of the articular surfaces of a knee specimen and of the knee ligaments, which are modeled by line elements. Both serve as input for the computer simulation model of the knee

sentations of the articular surface geometry of the femur and tibia (Fig. 17) as measured in a number of knee specimens using a specially developed stereophotogrammetric technique (Huiskes et al. 1985 a). The model allows for some deformability in the contact zone between the articular surfaces. The menisci are not represented in the model due to a lack of basic theoretical data. Experiments have shown, however, that the menisci have little effect on motion patterns in the intact knee joint (Huiskes et al. 1985 b). The ligaments are represented by a number of nonlinear, elastic line segments with the properties shown in Fig. 11. The ACL is modeled as consisting of 3 lines, the PCL as consisting of 2. The input data for the simulation model consist of the successive flexion angles and external load configuration. The following data are available as output: the successive three-dimensional equilibrium states of the joint (tibial rotation, varus-valgus rotation, and translation in the 3 directions), the locations of the femorotibial contact zones, the contact forces, the ligament strain values, and the ligament forces. Because this is a very complex model and some quantities (such as the stiffness curve of the modeled ligaments and their primary tension) are not precisely known, use of the model requires continuous checking against experimental findings. So far it has been found that the results of motion experiments can be simulated quite accurately with this model (Blankevoort et al. 1991 b). Of course, the model yields much more information than experiments, especially in terms of the forces that develop in the joint.

As an example, let us consider a simulation experiment in which the joint is flexed 20°, and internal and external torques are applied to the joint so that the tibia rotates from +30° to −30° (Blankevoort and Huiskes 1988). The model can calculate, among other things, the forces in the modeled ligaments, the contact forces, and the relative contribution of these forces to the equilibrium of forces and moments. Figure 18 shows sample results in which the calculated forces in the modeled ACL and PCL are plotted as a function of tibial rotation. These data clearly depict the recruitment of the ACL during internal rotation. The forces computed in the modeled collateral ligaments indicate that the lateral collateral ligament is a restraint only to external rotation, while the medial collateral ligament restrains both external and internal rotation. These results are consistent with the experimental findings of Ahmed et al. (1987), who estimated the ligament forces from measurements with "buckle transducers" (Lewis et al. 1982). But here again, the model provides additional information. The external torque is held in balance by the sum of the moments produced by the transverse components of the ligament forces and the moments produced by the transverse components of the contact forces. The contact forces in turn are produced indirectly by the tension of the ligaments, as shown in Fig. 10. The model can calculate the relative contributions of both these effects, the direct effect of ligament resistance and the indirect effect via the contact surfaces. The result, shown in Fig. 19, clearly indicates that the contributions of both effects are approximately equal, es-

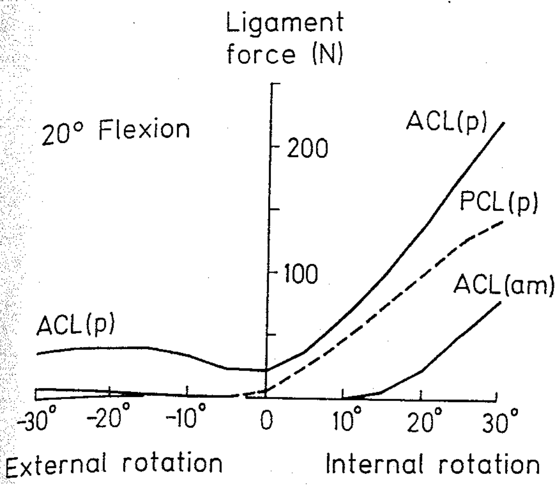


Fig. 18. Forces in the cruciate ligaments during internal and external rotation of the knee for a fixed flexion angle of 20°, determined by simulation with a mathematical knee model. *PCL (p)* posterior portion of PCL, *ACL (p)* posterior portion of ACL, *ACL (am)* anteromedial portion of ACL. (From Blankevoort and Huiskes 1988)

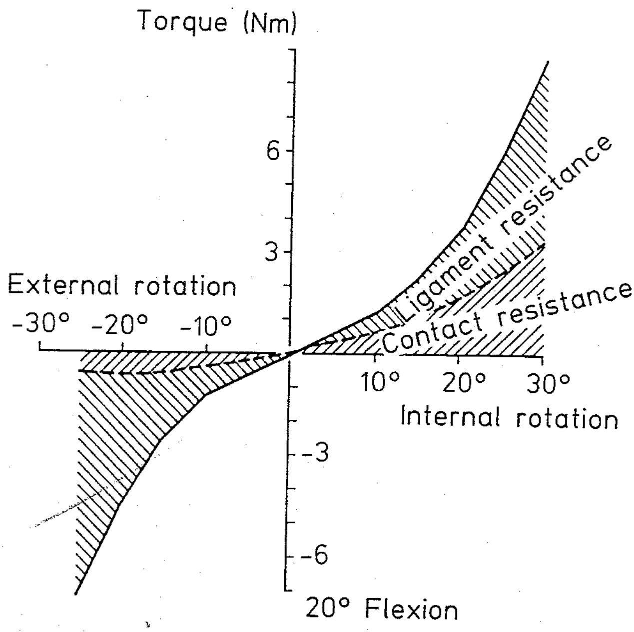


Fig. 19. Torque versus internal and external knee rotation for a fixed flexion angle of 20°, and distribution of the torque over 2 mechanisms of resistance, determined by simulation with a mathematical knee model. (From Blankevoort and Huiskes 1988)

pecially in internal rotation. Thus, through its indirect effect on articular contact forces, the ACL functions as a much more important restraint to internal rotation than one would expect from its anatomic configuration and the forces that develop in the ligament.

This analysis offers a prime example of the potential importance of simulation models, which can be used to investigate an almost limitless number of fundamental questions. A second example that further illustrates the complex role of the ACL concerns the problem of "isometric" cruciate ligament reconstructions. This simulation experiment (Blankevoort and Huiskes 1987) focused on the effect of nonisometric placement of the ligament, which was mathematically modeled by 2 nonlinear elastic lines. The joint was flexed in the unloaded condition, and all the foregoing kinematic and mechanical quantities were recomputed. Primary attention was given to the length changes in the anterior and posterior components of the ACL as the knee moved through flexion. As had been found in the experiments described above, the study showed that the anterior portion of the ACL elongates during flexion, while the posterior portion shortens (Fig. 20). Another simulation was run to learn how the lengths would change if the femoral attachment were placed 5 mm farther anteriorly or posteriorly. These results, also shown in Fig. 20, indicate that advancing the ligament anteriorly causes the ligament to undergo greater elongation, while moving the ligament posteriorly causes it to undergo greater shortening. These results, which agree with the experimental findings of Hefzy and Grood (1986) and Sidles et al. (1988), suggest that placing the ligament too far anteriorly in an ACL reconstruction could expose the ligament to excessive forces and restrict joint mobility. Placing the ligament too far posteriorly could lead to instability. Such a conclusion is frequently drawn in the literature, due in part to the fact that it is intuitively clear. It should be noted, however, that our analyses, like those of Hefzy and Grood (1986) and Sidles et al. (1988), were based on identical motion paths. Thus, while the analysis included the elongation imposed on the ligament by the knee motion, it did not take into account the feedback effect on the motion pathway due to the altered force pattern caused by transposing the ligament attachment (see Fig. 10).

This led us to perform a second experiment in which the actual alternative reconstructions were simulated. A separate simulation was run for each configuration. The results indicate that the length changes caused by the altered ACL attachments are less dramatic than one might have thought based on the al-

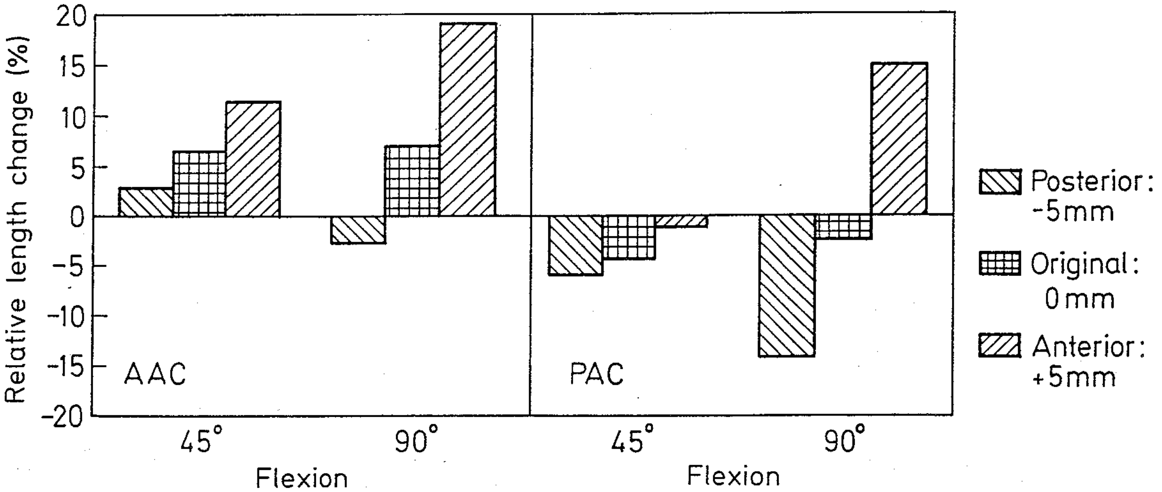


Fig. 20. Anticipated effect of AP transposition of the femoral attachment of the ACL on relative length changes in the modeled *AAC* (anterior part of ACL) and *PAC* (posterior part of ACL) for 2 identical positions of flexion in a simulated knee, where the transposition is not actually carried out. The relative length changes are shown relative to the length in the extended knee. (After Blankevoort and Huiskes 1987)

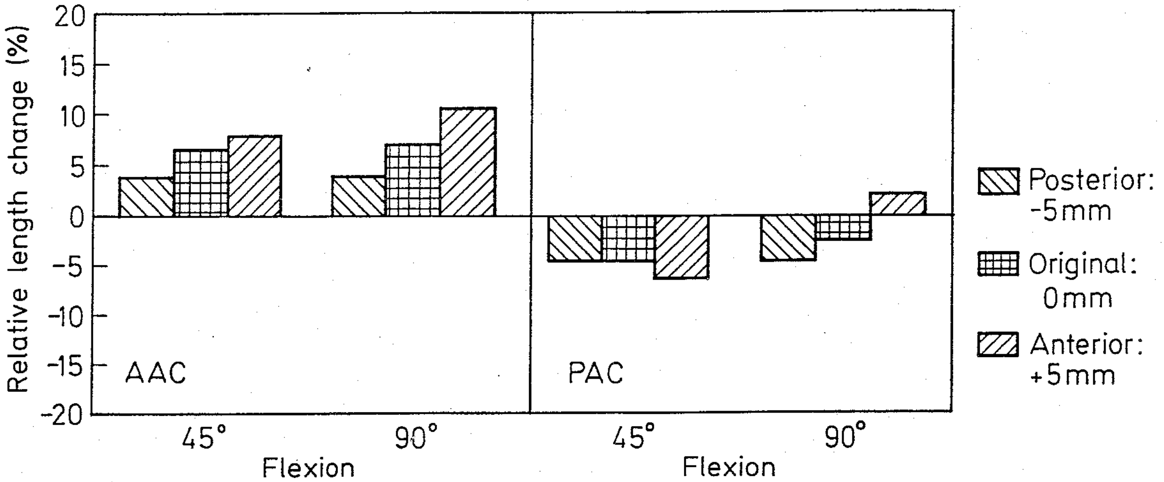


Fig. 21. Effect of an *actual* variation in the a.p. placement of the femoral attachment of the ACL on relative length changes in *AAC* (anterior part of ACL) and *PAC* (posterior part of ACL) for 2 identical positions of flexion in a simulated knee. (After Blankevoort and Huiskes 1987)

teration of typical ligament length changes noted in the previous experiment (Fig. 21). The reasons for this discrepancy can be read directly from the diagram in Fig. 10: A change of insertion geometry alters the pattern of forces via the mechanical properties of the ligament, which in turn alters the course of the movement. The change in the motion pattern chiefly affects a.p. translation (Fig. 22): An anterior femoral attachment of the ACL causes an anterior displacement of the femur with respect to the tibia – perhaps as much as 3.5 mm at large flexion angles. This results from the decreased length of the ligament, which holds the femur more anteriorly on the tibia during flexion. A posterior femoral attachment gives the femur a greater freedom of posterior motion during flexion, allowing the PCL to pull the femur backward. Of course, the altered motion pattern also affects the strain values in the PCL. These increase to a maximum of 3.5% relative to the initial situation. Thus, the system partially "corrects" for the effects of the altered biomechanical environment of the ACL by altering the pathways of joint motion, giving rise

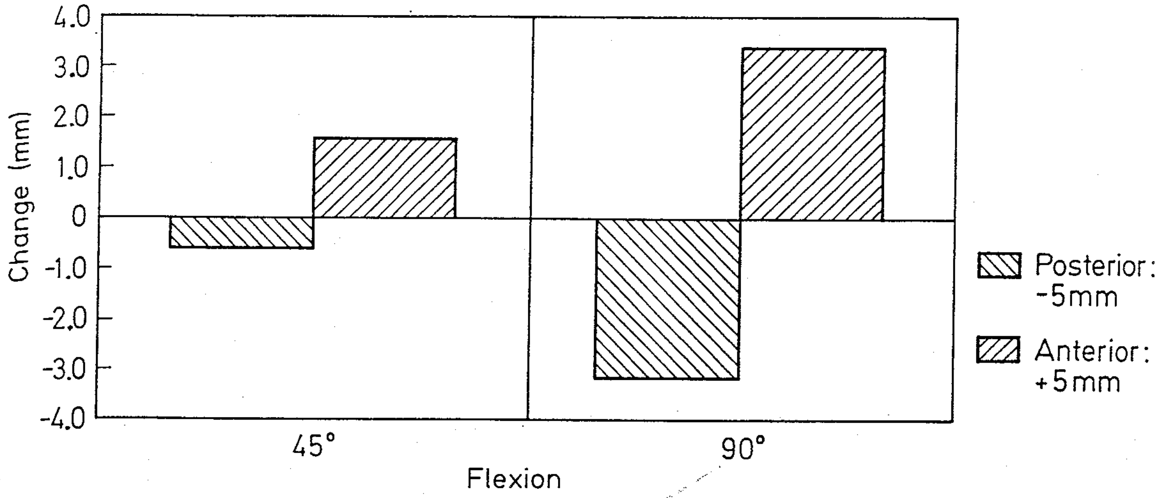


Fig. 22. Change in the a.p. position of the femur relative to the tibia caused by varying the a.p. placement of the femoral attachment of the ACL for 2 identical positions of flexion in a simulated knee. (After Blankevoort and Huiskes 1987)

to different strain values in the remaining ligaments.

This does not mean that a nonanatomic reconstruction of the ACL is desirable, because it could very well have adverse effects in terms of knee laxity and the stresses that develop in the ligament. We may conclude, though, that the sensitivity of the joint to deviations of ligament placement is less than is occasionally believed (and reported). Above all, however, this example gives us greater insight into the functional relationship between the ACL and the other anatomic structures of the knee. This means that the effects of cruciate ligament lesions and reconstructions are very difficult to appreciate in an isolated analysis. The use of simulation models is practically indispensable for an integral analysis.

Discussion and Conclusions

The functional analysis of the ACL is a three-dimensional problem. Two-dimensional anatomic models such as the four-bar linkage are useful for obtaining a general appreciation of the primary effects of motions in one anatomic plane. But for a more detailed analysis, the knee must be conceptualized as a three-dimensional system in which mechanical feedback plays an important role. Because the behavior of the knee and the role of the ligaments and articular surfaces happen within the context of a play of forces, analytical methods are necessary to comprehend and interpret the results of experiments. The computer simulation model discussed here is an example of this. Although it is not perfect, use of this model for the simulation of experiments offers greater insight into the relevant phenomena than can be gained from experimental results alone.

The knee joint derives its complexity from its complicated anatomic structure and mechanical properties as well as from its three-dimensional laxity and the kinematic coupling between its degrees of freedom. Conceptually, the joint can be regarded as a mechanism with 2 independent degrees of freedom. The ligaments restrain mobility and produce kinematic couplings in both a direct sense and an indirect sense via the articular surfaces. In the process, a ligament does not function as a unit but as a complex of collagen fibers that are recruited as required. This recruitment process has considerable clinical significance in both the diagnosis and repair of ligamentous injuries. Thus, a fiber bundle damaged by a tear is not recruited, and an incorrectly positioned fiber bundle will not be recruited at the appropriate time.

Drawer tests are unquestionably useful for the evaluation of the ACL, especially when they can be documented and quantified with the aid of an instrumented test device (Daniel et al. 1985; Edixhoven et al. 1987, 1989). An example of an instrumented drawer tester is shown in Fig. 23. However, these devices are based on the concept of the knee as a two-dimensional mechanism, and they are most effective for testing the fibers that must be recruited in selected positions of flexion and tibial rotation. For a more exacting diagnosis of partial tears of the ACL, additional test methods are required. Perhaps this could be accomplished by developing methods based on the recruitment mechanism described above and

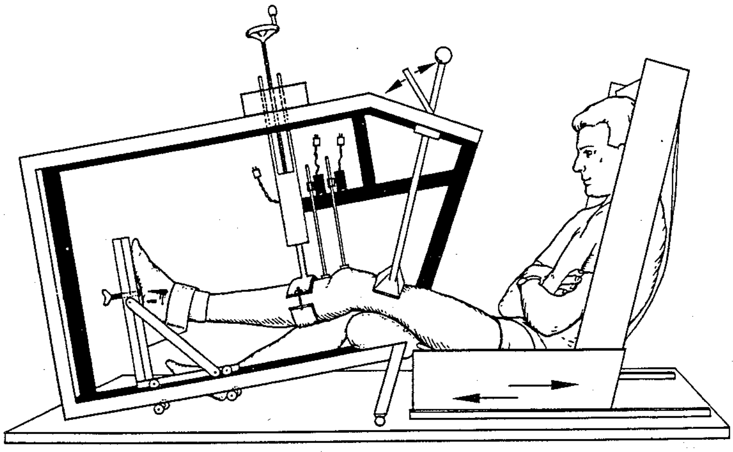


Fig. 23. Instrumented drawer tester of Edixhoven et al. (1987)

the indirect effect that the ACL exerts on the limits of knee motion via the articular surfaces. The function of the ligament should be approached as a three-dimensional problem, even in laxity tests.

One of the very few acknowledgments of this basic concept in the clinical diagnosis of ligamentous injuries is the pivot shift test (Galway et al. 1972), an essentially three-dimensional test that is difficult to perform and quantitate. The pivot shift test is an important component of a quantitative biomechanical analysis that respects the three-dimensional character of ligamentous function.

Even in discussions of "anatomic" reconstructions of the ACL, the two-dimensional four-bar linkage model is implicitly used as a reference. Often this disregards the fact that an alternative ligament results in an alternative recruitment pattern, which in turn alters the characteristics of joint motion through mechanical feedback within the joint as a whole. The extent of this process depends in turn on the condition of the PCL and collateral ligaments.

The finding that the function of the ACL is far more complex than is often assumed on the basis of simple anatomic models, while scientifically interesting, is of no immediate clinical benefit in itself. This chapter, then, is concerned more with defining the problem than offering a solution. Nevertheless, it would be very useful to approach the injured ACL as a three-dimensional problem in the development of diagnostic techniques, therapeutic procedures, and diagnostic and therapeutic instruments.

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