Mathematical modeling of the knee

Chapter 21

Mathematical Modeling of the Knee

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Introduction

A model is a representation of reality that emphasizes its most important characteristics. By concentrating on the essentials, the complex reality becomes surveyable, controllable, and comprehensible. Models are indispensable tools for understanding the knee joint, where subtle interplay between anatomic structures determines complicated motion patterns. The history of modeling human joints is as old as the study of these structures themselves. Most of the models used were based on analogies. Examples are the ball-in-socket model of the hip joint, the hinge model of the elbow joint, and the four-bar-linkage model of the knee joint.[1] The beauty of these models is that they relate the complex joint motions directly to relatively simple engineering mechanisms that are well understood; can actually be fabricated, manipulated, and tested; and, indeed, mimic some of the most essential kinematic characteristics of the real joints. The kinematic characteristics of these engineering mechanisms can also be described mathematically, resulting in a mathematical model of a physical analogy of a joint. Because the analogies in these examples are relatively simple, the mathematical models can be very precise, and can, in fact, replace the physical analogy. The mathematical model then has the advantages of flexibility and quantifiability but, because it is abstract, does not provide the possibilities for manipulations and visual confirmation, which are the assets of the physical model.

Because the analogy between the knee joint and a four-bar-linkage mechanism is limited to some of these essential kinematic characteristics, the applicability of this model for the understanding of the knee mechanism also is limited. This does not imply that there is something wrong with it. Limitation of analogy is the central asset of any model. Without such a limitation, the model would not be a model, but a reproduction as complex as reality itself, and as difficult to comprehend. The problem with the four-bar-linkage model is that, although it does provide answers to important questions of functional anatomy, its limitations prevent our answering some important questions about diagnosis and surgical reconstruction of knee disabilities. To answer many of these questions we must have quantitative understanding of the complex relationships between joint structures, forces, and motions that cannot be provided by simple models, nor by experiments alone.

Although there is no obvious mechanical analogy of the knee beyond the four-bar-linkage mechanism, the possibilities for mathematical modeling are not exhausted. In fact, owing to the development of computers, these possibilities have become almost limitless, or appear limitless at this point. These possibilities were first recognized about 15 years ago, in a few groups active in biomechanical analysis using finite element methods. In the finite element methods, a structure is divided in parts, the mechanical behavior of which can be mathematically described with reasonable accuracy. A computer is used to solve the interrelationships between the elements (parts) to evaluate the mechanical behavior of the structure as a whole. This approach implied that the geometric and material properties of the relevant substructures of the knee, such as ligaments, menisci, and articular surfaces, would be described mathematically in a computer program that could simulate their mechanical interrelationships. Although the mathematical equations that govern these interrelationships had not yet been developed, and the geometric and material properties of the substructures had not been determined, the groups concerned were optimistic about the time and efforts required to develop a realistic computer-simulation model. Once developed, this model (or numeric analogy) was expected to provide complete understanding of the complex knee mechanism and to be useful in solving an almost unlimited number of practical problems relative to criteria for diagnosis and treatment for knee disabilities.

Fifteen years later, it is evident that the difficulties to be overcome for the realization of these goals were grossly underestimated. The efforts were not appreciated all that much by orthopaedists in clinical practice or in research, and many groups have abandoned this approach. Although the original vision has not yet been realized, progress was made; a number of knee-joint computer-simulation models are now in working condition and are used to address practical, clinical problems.

In this chapter, I discuss the philosophy and the principles on which these mathematical models are based and present a brief historical overview, followed by a discourse on experimental verification. The applicabilities of the models are discussed relative to their characteristics and limitations. Finally, a critical assessment concludes that these models must actually be used and further developed, proven by corroborating experimental data, and fueled by clinical interest.

The Characteristics of Mathematical Knee Models

Mathematical knee model is a generic term. Persons not familiar with the mathematics by which models are defined may well be confused about their similarities and differences. The literature will not be helpful, because although models are actually characterized by their practical limitations, authors usually emphasize the beauty of the mathematics and the models' potential applicabilities. To start from a common ground, I will introduce a conceptual model of the passive knee joint as a mechanical control mechanism, activated by forces. The model is limited to the tibiofemoral articulation, although this is a practical, rather than a principle choice. According to the conceptual model (Fig. 1), knee motions are caused by forces that are either internal or external. The relationship between forces and motion, signified by conversions in Figure 1, can be described by the Newtonian dynamic equilibrium equations. The external forces, applied to the tibia and femur, may be caused by gravity, by accelerations of the bones, and by muscles or outside manipulation. This conceptual model assumes that these external forces have independent, given (variable) characteristics. Because the model does not account for the relationships between knee motions and external forces through the nervous system, it is called a model of the passive knee joint. The internal forces, however, do depend on the motions, via the geometry and mechanical properties of the ligaments, capsular structures, menisci, and articular surfaces. Hence, a number of feedback loops in the force-motion relationship are apparent (Fig. 1).

This conceptual model is a generic one, relative to which the characteristics of all actual mathematical models can be explained. These characteristics depend on the limitations, simplifications, and assumptions on which a model is based. They may relate to the formulation of the dynamic equilibrium equations used in the model, or to the precision and degree of complexity by which the anatomic structures are mathematically described.

Two-Dimensional (2-D) Models

A 2-D model assumes that the motion and loading characteristics of the knee can be represented in one plane. Hence, for a model describing the knee mechanisms in the sagittal plane, internal-external rotation, varus-valgus rotation, and mediolateral translation are neglected. An example is the four-bar-linkage model[1,2] extended by O'Connor and associates[3] in a mathematical sense by including elastic, instead of rigid ligaments. Although simple, 2-D models provide insight in the generic relationship between condylar shape and ligament properties.

Quasi-Static Versus Dynamic Models

The knee joint is a dynamic system in the sense that the external forces and the motions vary in time. Because variation of motion

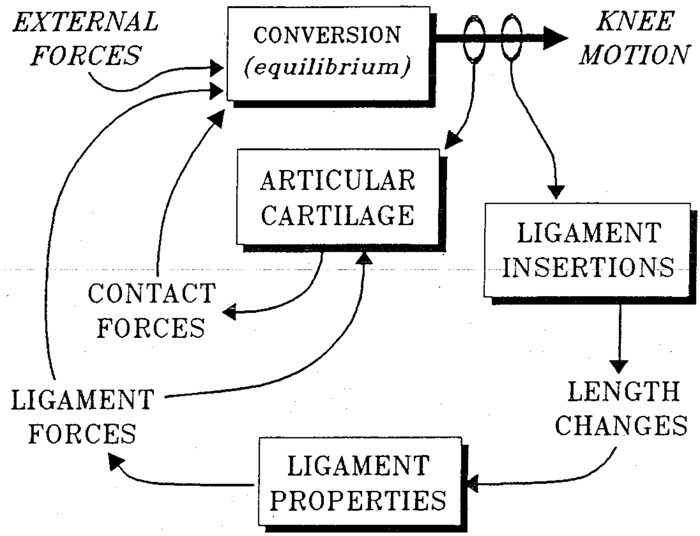

Fig. 1 *Conceptual model of the knee joint as a mechanical system. Knee motion occurs as a result of dynamic equilibrium between external and internal forces. The latter depend on the motion characteristics, fed-back via the anatomic knee-joint structures. (Reproduced with permission from Blankevoort L:* Passive Motion Characteristics of the Human Knee Joint, *thesis. The Netherlands, University of Nijmegen, 1991.)*

implies accelerations and decelerations, the dynamic equations of force equilibrium (Fig. 1) have terms representing the forces caused by the mass inertia of the elements involved, such as the bones. If the mass of an element is relatively small, or if its accelerations and decelerations are relatively small, the contributions of these inertia forces are also relatively small and could, potentially, be neglected. This underlying assumption of quasi-static (sometimes called kinetic) knee-joint models does not imply that forces and motions are constant; it only means that their variations occur relatively slowly, in the sense that the inertia forces are negligible relative to the other external forces. In practice it implies that quasi-static models cannot be used to study the effects of impact forces. The force and motion variations involved in normal walking can be considered as being relatively slow, as probably, can those involved in running. There is no clear dividing line; the faster the motion variations occur, the less precise a quasi-static model becomes.

Most of the knee-joint models hitherto developed were quasi-static ones. Moeinzadeh and Engin[4] developed the mathematical formulation for a three-dimensional (3-D) dynamic model; however, actual calculations were presented relative only to a 2-D model.

Elastic Versus Time-Dependent Properties

The equations of dynamic equilibrium also contain terms that account for the force variations resulting from time-dependent behavior of the joint structures (viscoelasticity). For instance, these terms describe the loading-rate dependency in the responses of ligaments and articular cartilage. These effects usually are neglected in mathematical knee-joint models; the only model known to me in which they are not neglected is that of Moeinzadeh and Engin.[4] The characterization dynamic often is assumed to include this time-dependent behavior, whereas quasi-static is assumed to imply that properties are purely elastic and, thus, time independent.

Contrary to the above difference between a quasi-static and a dynamic model, there is very little explicit justification for neglecting the time-dependent effects of the knee-joint structures. Results of experiments with cartilage and ligaments indicate that under normal, physiologic conditions, time-dependent behavior does play a role.[5,6] Hence, the response of the joint to external forces is loading-rate dependent. As long as the loading rate is relatively slow, such as in diagnostic manipulation, the rate dependency in the motion characteristics is probably of minor consequence.[5] A much more important effect is that of preconditioning of the ligaments; because the ligaments are stretched after a period of rest, they are much stiffer after they have been stretched (or preconditioned) a number of times consecutively.[5] When the knee is loaded in a knee-laxity tester, its response to the first loading cycle is significantly different from responses to the consecutive cycles.[7]

Inverse Versus Direct Dynamics Models

In mechanics, forces are causes and motions are effects. Without force, there is no motion. According to the conceptual model of Figure 1, the knee system is activated by external forces, which produce motions. The kind of motions (or joint positions) that will actually result depends on the characteristics of the external forces, but also on the internal forces, which again depend on the motions. In fact, the motion characteristics will be such as to satisfy dynamic (or quasi-static) equilibrium between internal and external forces. In direct dynamics models, this process is stimulated. Hence, in the mathematical solution procedure, those motions or positions are sought that satisfy the equilibrium between external and internal forces, just as it occurs in reality. The mathematical complexity is introduced because of the nonlinear feedback loop in the system. As a result, the relationship between forces and motions is not explicit.

Reversing causes and effects in the mathematical analysis produces inverse dynamics models. In that case, the motion characteristics are given quantities, and the forces are determined. To determine the internal forces in this way is relatively simple, at least mathematically, because a particular given position of the tibia relative to the femur uniquely determines, for example, the amount of elongation in a ligament or the amount of compressive strain in the articular cartilage. Therefore, assuming that the force/length relationships (the elastic characteristics) of the knee-joint structures are known, the internal forces in each structure can be directly calcu-

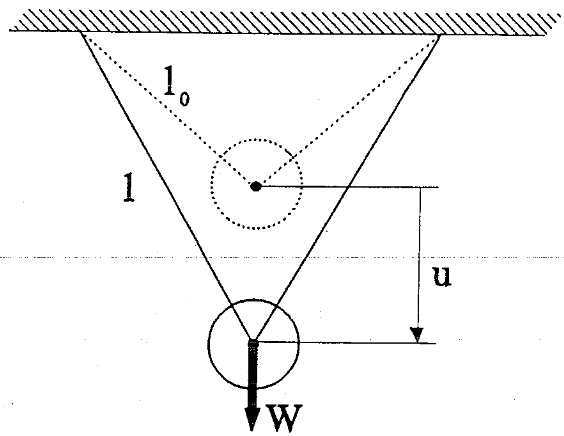

Fig. 2 *Illustration of the difference between a direct and an inverse dynamics model, using a ball hung from the ceiling by two equal rubber bands.*

lated from its deformation, which is uniquely and explicitly determined by the relative positions of the bones.

The difference between a direct and an inverse dynamics model may be illustrated by an example (Fig. 2): Assume a ball is hung from the ceiling by two rubber bands of equal thickness. The nonlinear force/elongation characteristics of the bands are given, as well as the position of the ball in which each band is straight but unloaded (the unloaded reference position). The variables of interest are the weight of the ball, the forces in the bands, and the ultimate, equilibrium position of the ball when it hangs free, relative to the initial, unloaded reference position. In a direct dynamics problem, the weight of the ball (W) is given, while the forces in the bands must be determined. In an inverse dynamic problem, the equilibrium position of the ball (u) is given, while the forces in the bands must be determined. Few readers will be able to solve the first problem, and no one will solve it easily. All readers who can read a nonlinear force/elongation graph and know Pythagoras' theorem will be able to solve the second problem.

Understanding the differences between these two approaches is important, because it separates the two main classes of knee-joint models in the present literature. Examples of inverse dynamics models are those of Crowninshield and associates[8] and Grood and Hefzy,[9] who modeled the 3-D ligament configuration around the knee to determine the ligaments' relative contribution to knee stiffness, using prescribed motion characteristics as input. A similar approach was taken by Walker and associates,[10] who included the average 3-D shape of the articular surfaces in a geometric model to determine ligament-length changes and contact areas for given motion patterns. Because no forces were determined in this model, and constitutive properties of the ligaments were not considered, it was rightfully named a computer-graphics model.[11]

Direct dynamics models were reported by Wismans and associates,[12,13] Andriacchi and associates,[14] Moeinzadeh and Engin,[4] Essinger and associates,[15] and Blankevoort and associates.[16-18] Except for the model of Moeinzadeh and Engin,[4] discussed above, all these models are 3-D and quasi-static. These models are similar in the sense that they describe the 3-D shape of the articular surfaces mathematically and represent the ligaments and capsule structures with one or more line elements (or springs) with particular elastic properties (Fig. 3). In all cases, a particular flexion angle and an arbitrary system of external forces must be prescribed. Therefore, knee function is simulated as a series of flexion steps. After each step, the equilibrium positions of the bones are determined from the quasi-static equilibrium equations and the equations describing the contact conditions between the condyles. Andriacchi and associates[14] used an indirect finite element approach for the mathematical solution strategy. Wismans and associates[12,13] introduced an iterative solution procedure based on the Newton-Raphson method, which is more computer efficient and solves the equations directly. The same approach was followed by Blankevoort and associates.[16-18] Essinger and associates[15] used the principle of minimum potential energy, which has an effect similar to that of the direct iterative procedures.

Because the direct dynamics models are intended to describe the knee mechanism as it works in reality (Fig. 1), they may be rightfully called computer simulation models. Instead of external forces, motions (3 rotations and 3 translations) may also be prescribed in these models. Hence, these direct dynamics models also can be used as inverse ones. The opposite is not possible.

Representation of Articular Shape and Contact

Direct dynamics models must always include descriptions of articular shape and contact conditions. Huiskes and associates[19] developed an effective and accurate method to measure and describe articular geometry, using stereophotogrammetry and polynomial interpolation. This method was later enhanced, both in accuracy and efficiency, by Ateshian and associates.[20] It was used by Blankevoort and associates[16] to measure four knee-joint specimens, each of which was analyzed individually. Essinger and associates[15] used femoral and tibial condylar shapes from 10 knee-joint specimens, determined from sagittal and frontal sections.[10] Wismans and associates[12] measured the articular surfaces of knee-joint specimens with a dial gauge and also used polynomial interpolation. Andriacchi and associates[14] took geometric data from the literature.

Of the four computer-simulation models reported in the literature, only Andriacchi and associates[14] included a representation of the

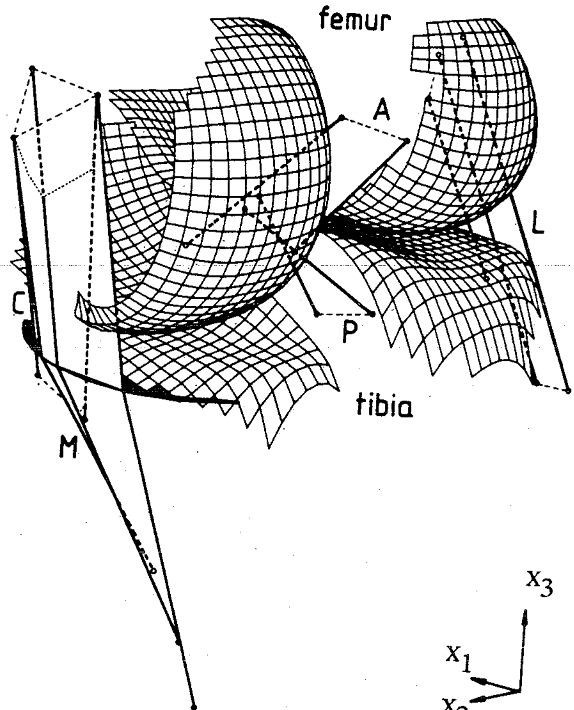

Fig. 3 *Graphic representation of the computer-simulation model of Blankevoort and associates seen from medioposterior with the knee in extension. (Reproduced with permission from Blankevoort L, Huiskes R: Ligament bone interaction in a three dimensional model of the knee.* J Biomech Eng *1991;113:263-269.)*

menisci. They did this by using a shear beam element in the FE-model, the precise characteristics and justification of which are not entirely clear to me. Certainly, modeling the meniscus remains an unsolved problem. The contact elements used in the model of Andriacchi and associates[14] were nonlinear elastic. Wismans and associates[12] assumed rigid joint surfaces; hence, contact points rather than contact areas were represented in the model. Essinger and associates[15] used a rigid femoral surface against a deformable tibial one; the cartilage of the tibia was assumed to be linear elastic. Blankevoort and associates[16] included thin, nonlinear elastic layers on both femur and tibia to represent the articular cartilage, assuming uniaxial stress within the layers. None of the models included surface friction, and in all cases the cartilage properties (stiffness values) were estimated based on the literature.

In fact, articular cartilage is a biphasic material, displaying complex nonlinear and time-dependent mechanical properties.[6] Blankevoort and associates[16] investigated the consequences of a simplified deformable contact formulation in their model by varying cartilage stiffness values in both linear and nonlinear elastic models. The

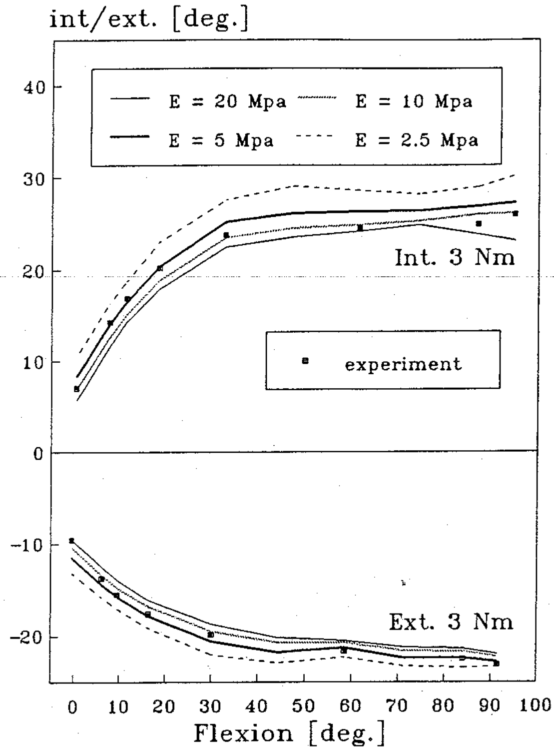

Fig. 4 *Internal and external rotation laxity for torques of ±3 N·m as functions of flexion, as determined with the computer simulation model.[16] In the model, the elastic modulus of cartilage was varied between 2.5 and 20 MPa. The effects on the laxity are relatively small. Also shown are the experimental findings for the same knee-joint specimen from which the model geometry was derived. (Reproduced with permission from Blankevoort L, Kuiper JH, Huiskes R, et al: Articular contact in a three-dimensional model of the knee.* J Biomechanics *1991;24:1019-1031.)*

effects on the knee-motion patterns, ligament forces, and contact forces were very small for moderate axial loading conditions (Fig. 4). It must be expected, however, that larger axial forces, such as occur during walking, will have more significant effects, making the simplified representations used in the models less precise in that case. In addition, the simplified models would not be very precise for the investigation of the local stress distribution in the cartilage. Of course, the lack of menisci in the model also reduces precision in this respect.

Representation of Ligament Function

In all knee-joint models presently known, the ligaments are represented by one or more line elements that run from tibial to femoral insertions (Fig. 3). These line elements are assigned particular elastic properties in the sense of force-elongation characteristics. Wismans and associates,[12,13] Essinger and associates,[15] and Blankevoort and associates[16-18] assumed nonlinear elastic force-elongation characteristics, whereby the parameter values describing the stiffness characteristics were estimated from the literature. Andriacchi and associates[14] assumed linear elastic properties. The number of line elements used per ligament varied from one to four in the different models, whereas in some cases capsule structures also were taken into account by additional line elements.

In fact, the knee-joint ligaments are complex 3-D structures of which different parts are tensed in the various knee-joint motions.[2] The choice for one, two, or more line elements per ligament seems to have been made rather arbitrarily; at least it was not based on rigorous analyses of the ligaments themselves. The ligament properties are nonlinear and time-dependent.[5] The latter aspect is neglected in all models. Another important parameter to be included in the models is the prestrain existing in the ligaments, which depend on the unloaded (or zero-force) length, about which, in fact, nothing is known. Andriacchi and associates[14] did not include prestrain. Wismans and associates[12] and Essinger and associates[15] selected prestrain values on rather arbitrary grounds. Blankevoort and Huiskes[18] assess prestrain value from comparisons between experimental and model results.

Finally, the medial collateral ligament, in particular, does not run straight from its tibial to its femoral insertion, but wraps around the tibial condylar edge, whereby mechanical ligament-bone interaction occurs. This is accounted for only in the model of Blankevoort and associates.[17] The effects of this phenomenon were investigated relative to a line-element representation without bone interaction. It turned out that it had a minor effect on the knee-motion parameters, but did cause the medial collateral ligament to counterbalance forced valgus rotations more effectively.[17]

Experimental Verification

The 3-D direct dynamics models have been verified experimentally only to a certain extent. Wismans and associates[12] and Andriacchi and associates[14] have compared knee stiffness values determined by their models to experimental results published in the literature.[21,22] Essinger and associates[15] used experimental data on articular geometry and ligament insertion geometry, measured in 10 knee-joint specimens.[10] Ten knees were modeled such that ligament stiffness parameters, prestrains, and articular stiffness were equal in each case. In each analysis, a flexion motion from 0 to 120 degrees was simulated by quadriceps pull. Internal-external rotations, varus-valgus rotations, translations in the three perpendicular directions, ligament length patterns, and tibiofemoral contact patterns as functions of flexion were determined and averaged for the series of 10 simulations. These average results were similar to average experimental results reported in the literature.[10,23,24]

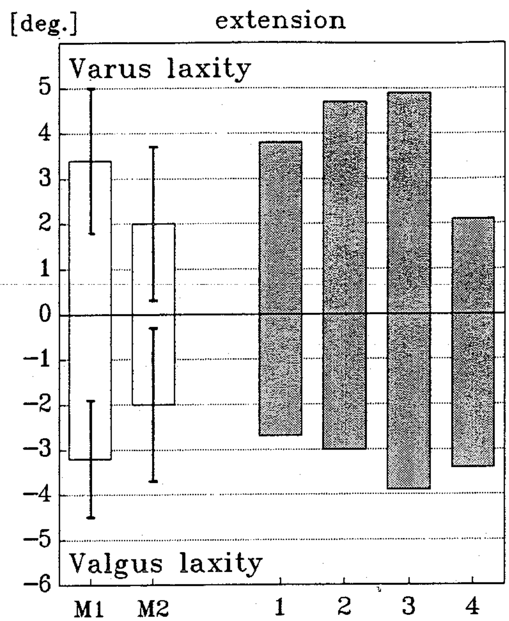

Fig. 5 *The varus and valgus laxities determined in computer-simulation models of four different knee-joint specimens*[18] *compared to in vitro experimental data* (M_1)[21,22] *and in vivo experimental data* (M_2).[26] *(Reproduced with permission from Blankevoort L:* Passive Motion Characteristics of the Human Knee Joint, *thesis. The Netherlands, University of Nijmegen, 1991.)*

Blankevoort and Huiskes[18] simulated individual experiments with four knee-joint specimens. In each case, the knee was moved through an internally and an externally rotated motion pathway, and internal-external rotations, varus-valgus rotations, and translation in the three perpendicular directions were measured as functions of flexion.[25] The knees were then dissected, and the articular surface and ligament insertion geometries were measured and used in four simulation studies. In each of these studies, the ligament stiffness parameters were equal, and their prestrain values were chosen from experimental ligament-length patterns. Experimental results and numerical predictions were compared for each knee specimen separately (Fig. 4). In addition, the four simulation models were subjected to anteroposterior-laxity tests in 20 and 90 degrees of flexion, and to varus-valgus-laxity tests in extension and 20 degrees of flexion. The results were compared with those obtained experimentally and reported in the literature (Fig. 5).[21,22,26] The similarities were quite satisfactory. Simulations were repeated with adjusted prestrain values per line element and per specimen, based on a numeric optimization procedure in which the individual prestrain values were optimized relative to requirements of minimal deviation between experimental and predicted motion patterns.[18] This improved the similarity with the experimental results. The articular contact-

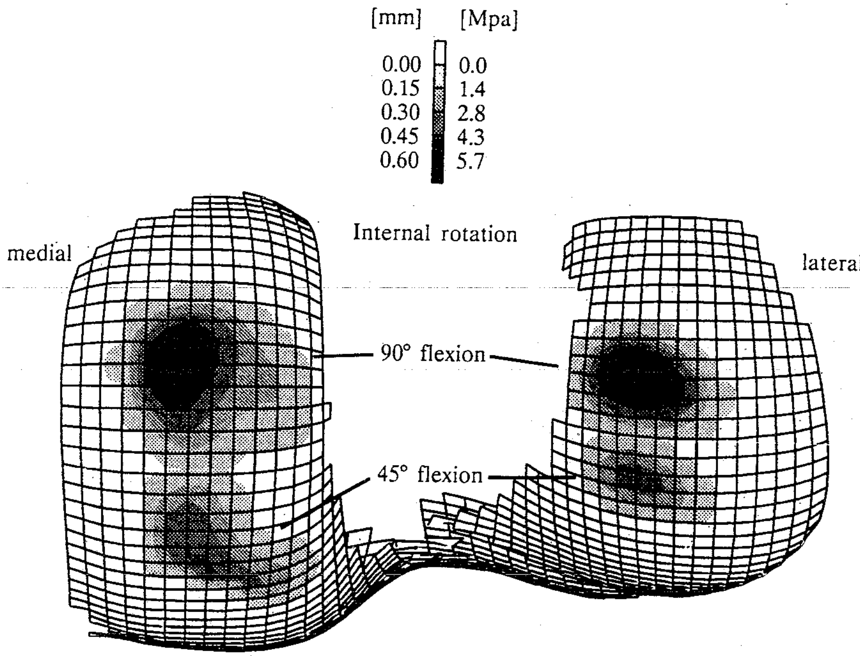

Fig. 6 *Contact-stress distribution on the femoral condyles, for 45 and 90 degrees of flexion during an internally rotated flexion pathway, as determined in the computer-simulation model. (Reproduced with permission from Blankevoort L, Kuiper JH, Huiskes R, et al: Articular contact in a three-dimensional model of the knee.* J Biomechanics *1991;24:1019-1031.)*

area locations, magnitudes, and contact-stress values obtained in the simulations (Fig. 6) were also compared with experimental results,[27,28] and, finally, the ligament forces predicted for internal and external rotation were compared with experimental values.[29]

Attempts at validation of the models have been directed predominantly at the general motion patterns and the general laxity patterns. In most cases, the model results were compared with average experimental data reported in the literature. Only Blankevoort and Huiskes[18] have compared experimental and numeric results for individual knee specimens, at least where 3-D motion patterns were concerned. In all cases, authors reported the results of confrontations with experimental data to be satisfactory. The question is, however, what is satisfactory. This question cannot be answered without considering the inherent limitations of the models and their uses.

First, the models are limited in their scopes; dynamic and time-dependent effects are neglected, limiting the validity of the models to relatively slow (quasi-static) loading rates. The simplified representation of articular cartilage, the neglect of friction, and the absence of the menisci restrict use of the models to moderate articular contact loads, that is, moderate axial knee forces. The absence of the menisci also implies that important secondary restraints to internal-external rotations and anteroposterior laxity are missing.[30] The same is true for the absence of some of the capsule structures. Thus, the models are probably less suitable to study the knee without the ligaments as primary stabilizers. Or, in a more general sense, the models are valid particularly for configurations as near to the normal knee as possible. Validation studies have not yet established precisely what slow loading rates, moderate forces, and 'as near to the normal knee as possible' mean in a quantitative sense.

Second, there are limitations to the degree of valid details. As noted, the simplified cartilage representation precludes an accurate, detailed analysis of local cartilage stresses, and the same is true for the ligaments. However, the detail to which, in a quantitative sense, the model results can be trusted, if only as reasonable approximations, is still uncertain.

Third, there is the problem of validation. If results of the simulation of an experiment are in excellent agreement with the results of the experiment itself, for example, the 3-D motion patterns for a particular loading case, it can be concluded only that the model, as a whole, behaves similarly to the knee-joint specimen as a whole. The agreement does not prove, for example, that all ligaments were modeled correctly. A deficiency in the model of one ligament may compensate for the deficiency in another. This problem has not yet been addressed sufficiently in validation studies.

An important question is what kind of a model is actually required: one of an average knee, of a particular knee, or of a typical knee. An average-knee model would not display only average 3-D motion patterns in one particular flexion-extension movement, as in the validation study of Essinger and associates.[15] It also would require average articular surface and ligament geometries, and average ligament and cartilage properties. And, of course, an average knee has two menisci. None of the four models discussed can meet these requirements, and none of the authors have really attempted to model a particular knee. Although Blankevoort and Huiskes[18] have included articular surface shapes and ligament geometries of particular knees and experimented with the same knees, their ligament stiffness parameters were average ones, taken from the literature, not from the particular knees. However, it is probably fair to say that, within the inherent limitations discussed above, the models of Essinger and associates[15] and Blankevoort and associates[16-18] have been validated as typical knee models in the sense that the differences in overall behavior between the models and an arbitrary knee joint are no more extensive than those between two arbitrary knee joints where it concerns geometric and mechanical properties and mechanical behavior.

Applications

Different applications of knee-model studies must be recognized. First, these studies can be conducted to investigate the models themselves. Such an objective may sound trivial or superfluous to some, but it is extremely important to assess the sensitivity in the behavior of a complex mathematical model to its particular characteristics or parameter values. Investigating the model points out, for example, if the results are very susceptible to a particular parameter, that is, if that parameter should be measured very precisely, or if the parameter's influence is only small and its precise assessment should be of lesser priority. Extensive sensitivity analyses of this kind were published by Wismans and associates[12,13] and by Blankevoort and associates[16-18] relative to ligament stiffness parameters (Fig. 4), prestrains in ligaments, amount and placement of line elements representing the ligaments, variations of articular-surface geometry, and articular cartilage properties.

A second objective is to obtain information about the mechanical interrelationships in the knee. Applications of this kind have been reported by Andriacchi and associates,[14] relative to the 3-D stiffness characteristics of the knee, and by Blankevoort,[31] relative to the contribution of the articular surfaces to resistance against internal-external rotation torques.

A third objective is the analysis of clinical problems. In this category, Essinger and associates[15] reported studies of contact-area locations, magnitudes, and contact stresses in different types of knee prostheses, and Blankevoort[31] applied his model to the problem of femoral placement and prestress of anterior cruciate ligament reconstructions.

For the last two classes of objectives, additional examples are briefly discussed.

Articular Surface Contribution to Rotary Laxity

When the tibia is rotated relative to the femur, internal forces develop in ligaments and between condylar contact regions. For a particular torque applied to the tibia, it will rotate until the internal forces balance the applied torque. Ahmed and associates[29] performed tests of this kind with a large number of knee-joint specimens. For increasing torque values they used buckle transducers to measure the amount of internal and external rotation and the forces developed in the ligaments. They then estimated the contributions of the ligaments to the restraining torque to find that the ligament forces could nicely balance the torque when applied to the tibia towards external rotation, but not when applied towards internal rotation. These authors concluded that "restraining mechanisms [other than tension in the ligaments] must be involved in resisting internal rotation."[29]

This test was simulated with the computer-simulation models of four knee-joint specimens.[31] Other than during the experiments, the model could also determine the articular contact forces. These contact forces were found to increase to about 100 N laterally and 200 N medially when the tibia was externally rotated to 25 degrees, and to

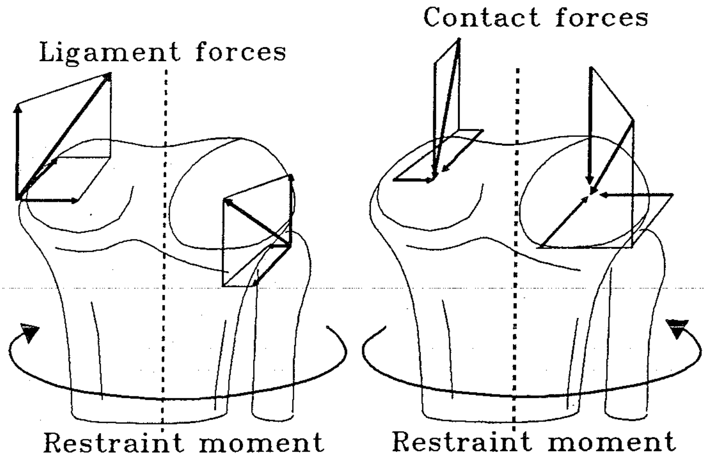

Fig. 7 *The ligament forces and the contact forces in the knee both have horizontal components, depending on the obliqueness of the ligaments and the inclinations in the plateaus. These horizontal force components contribute to the balance of the applied torque. (Reproduced with permission from Blankevoort L:* Passive Motion Characteristics of the Human Knee Joint, *thesis. The Netherlands, University of Nijmegen, 1991.)*

about 40 N laterally and 250 N medially when the tibia was internally rotated to 25 degrees. This increase, of course, is caused by the increase of ligament forces while the tibia rotates. As a result of the inclination of the tibial plateaus, these contact forces (perpendicular to the contact surfaces) have components in the horizontal plane that may contribute to resisting the applied torque (Fig. 7). Because of tibial articular geometry, this actually does occur in internal rotation, but does not in external rotation. Thus, it was found that, in internal rotation, the articular surfaces contribute an average of 50% to 85% (depending on the flexion angle) to the restraining torque (Fig. 8).

This is an example of the kind of information about biomechanical interrelationships in the knee-joint that is quite difficult to establish with experiments alone. The example also shows how experimental results can be better understood and generalized by using mathematical knee models.

Effects of Anterior Cruciate Ligament (ACL) Reconstruction on Knee Motion and Ligament Forces

It has often been suggested in the literature that the femoral and tibial insertions of an ACL reconstruction should be isometric, that is, that the graft length preferably should remain constant, at least by approximation, throughout the flexion range. Several authors have reported experimental studies in which isometric femoral insertion locations were determined in normal knee-joint speci-

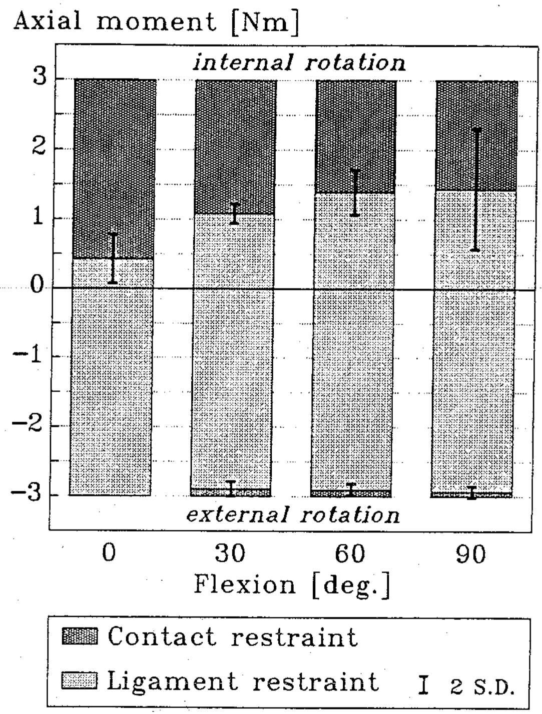

Fig. 8 *Contributions of the contact forces and the ligament forces to the balance of the applied torques of ±3 N·m. Averages and standard deviations were determined in the computer-simulation models based on the geometries of four knees. (Reproduced with permission from Blankevoort L:* Passive Motion Characteristics of the Human Knee Joint, *thesis. The Netherlands, University of Nijmegen, 1991.)*

mens.[32,33] The question is, however, will an isometric location still be isometric after the normal ACL is replaced by a graft? According to the scheme of Figure 1, this is very unlikely. Because of the interrelationships between internal forces, ligament properties and insertion locations, and motions, it is highly likely that the answer is negative.

ACL replacement was simulated in the model of a particular knee-joint specimen,[31] in which 11 alternative femoral insertion locations were tried (Fig. 9). Beside the different insertion locations, all other parameters were kept as constant as possible by assuming a graft with the same force-elongation characteristics as the original ACL and by using a consistent amount of pre-tension. In the normal case and in each of the 11 reconstruction cases, a full flexion motion was simulated, whereby the graft tension was calculated. In addition, anterior laxity and internal-external rotation laxity tests were simulated for the knee in 20 and in 90 degrees of flexion, whereby the anterior displacement and the internal-external rotation were determined. It was also found that after the reconstruction, the relative anteroposterior position of the tibia relative to the femur

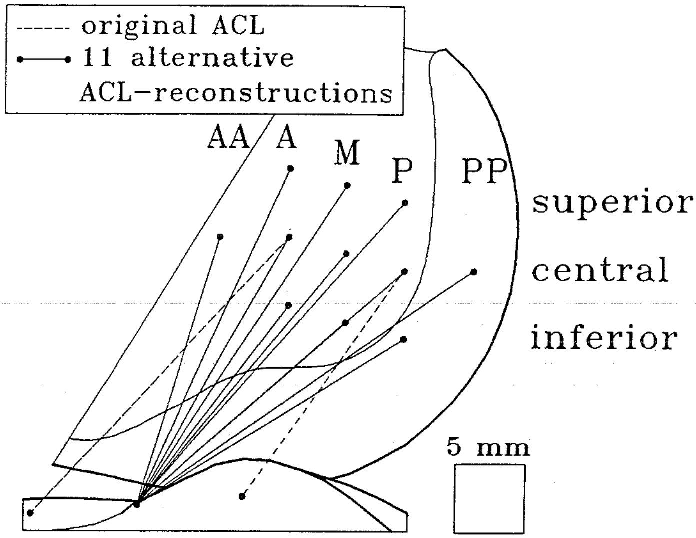

Fig. 9 *The 11 alternative femoral insertion locations tested in the simulation of ACL reconstruction. (Reproduced with permission from Blankevoort L:* Passive Motion Characteristics of the Human Knee Joint, *thesis. The Netherlands, University of Nijmegen, 1991.)*

changes in comparison with the normal situation. This shift, or anteroposterior position error, was determined as well, in 20 and 90 degrees of flexion (Fig. 10).

It is evident from the results that an isometric location will not remain isometric after normal ACL is replaced by a graft. The change in relative position of the femur and tibia indicates that the interrelationships of all the ligaments change as well. Thus, the motion characteristics are different after the reconstruction. The anteroposterior-laxity characteristics changed up to 36% relative to the normal situation in 20 degrees of flexion, and up to 93% in 90 degrees of flexion. For internal-external rotation laxity these changes were maximally 12% and 73%. Scoring the alternative insertion locations relative to normality of anteroposterior position (minimal anteroposterior error), anterior laxity, and internal-external laxity, and on minimal graft tension during flexion, it was concluded that locations A-central and A-superior (Fig. 9) provided the best options, and location AA-central the worst.

Applying a computer-simulation model is probably the only way in which this kind of information can be obtained. It would be almost impossible to perform the same experiments on knee-joint specimens because of variability in individual knee-joint specimens, the impossibility of performing multiple reconstructions in the same specimen, and the inaccessibility of the knee to a variety of measurement gauges. Of course, the simulation was performed relative to the geometry of a single knee-joint, and the simulation model is a simplified representation of reality. Therefore, it would be foolish

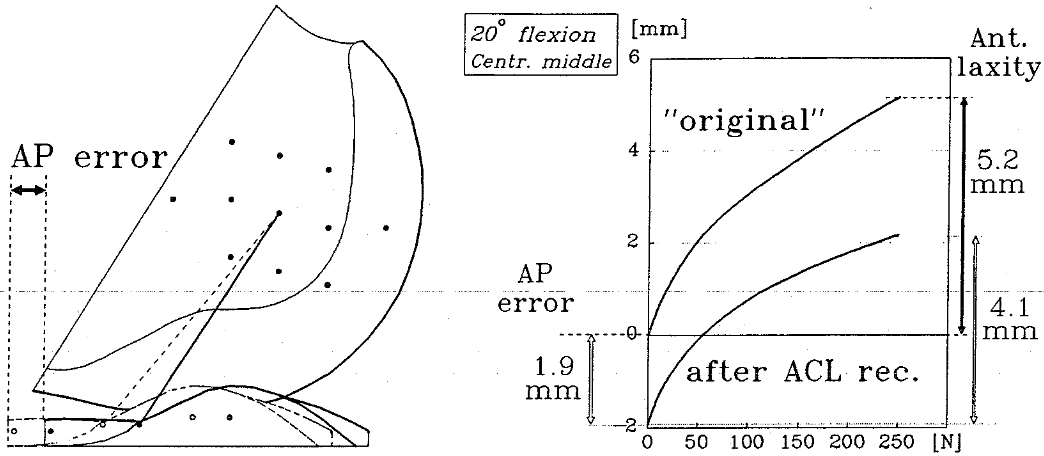

Fig. 10 *Left, If the ACL is replaced, an anteroposterior error usually occurs as a posterior shift of the tibia relative to femur.* ***Right,*** *A comparison of anterior laxity tests is shown between original and reconstructed cases. The anteroposterior error causes an adjustment of the ligament interrelationship and, thus, a change in the motion characteristics. (Reproduced with permission from Blankevoort L:* Passive Motion Characteristics of the Human Knee Joint, *thesis. The Netherlands, University of Nijmegen, 1991.)*

for clinicians to navigate solely on simulation results. Before definite conclusions with respect to optimal graft location can be drawn, the theoretical results must be confirmed experimentally, or at least corroborated by other experiences. But the model results do provide an excellent starting point for further experimentation.

Discussion

Many authors used knee-joint models in the past. These included simple mechanical analogies, 2-D mathematical models, geometric models, computer-graphics models, and inverse dynamics and direct dynamics mathematical models. This review was focused mainly on 3-D direct dynamics (quasi-static) models, which are, without doubt, the most complex and sophisticated ones, and which are rightfully called computer simulation models. That does not mean that these models are considered better than the others. In mathematical modeling, good is whatever does the job. There are many examples in the literature in which simpler models were successfully applied to clarify anatomic relationships, to generate useful information for the solution of clinical problems, or to determine mechanical variables based on experimental information. The limitation to computer simulation models is merely a matter of scope, not of quality.

In the early, optimistic years, when the development of these models started, much effort was put into the derivation of the mathematical equations and into the development of the numeric solution techniques. Owing to advances in mechanics and informatics in general, to the pioneering work of Wismans and associates[12] and Andriacchi and associates,[14] and to the exponential growth of computer capacity, these problems have now become much less important. In fact, if the mechanical characteristics of an anatomic knee-joint structure can be described mathematically, its inclusion in the computer-simulation models no longer presents major problems. Description of those mechanical characteristics, however, has turned out to be a far greater problem than originally anticipated. For example, the menisci are not excluded from these models because they are not believed to have a function, but simply because not enough is known about their mechanical behavior to model them mathematically. The same is true for the relatively primitive way in which the ligaments are represented; not enough experimental information exists about their time-dependent mechanical characteristics as 3-D structures for a more sophisticated representation. The lack of experimental information is why the first five years of development were focused on the mathematics, and the next ten years were devoted primarily to the collection of experimental data for the triggering of model parameters and for the validation procedures.

Presently, two 3-D direct dynamics models are actively used: one in Nijmegen, The Netherlands,[16-18] and one in Lausanne, Switzerland.[15] Both are of similar scope and subject to similar limitations. These limitations include predominantly a restriction to moderate loading cases, to relatively slow loading rates, to situations near to the normal knee configuration (but including total knee replacement), and to the level of detail for which realistic results can be expected. Both models have been subjected to extensive validation procedures, but, as discussed above, this process has not yet been completed. Therefore, important conclusions of practical or scientific significance, based on the results of the models, should be corroborated by experimental verification. Or, in other words, these models are excellent tools to generate hypotheses that can be used to direct and target experimental protocols in the laboratory, in animals, and in the clinic, and they also are great assets for explaining, extrapolating, and generalizing experimental and clinical findings.

In addition to further validation, application and improvement of the models should be emphasized in the coming years. Application could be in the areas of ligament reconstruction, analyses of manual and instrumented knee-laxity tests, analyses of intra-individual variability in knee-joint mechanical behavior, and development of artificial knee joints. Other applications may be in the area of teaching or instruction for orthopaedic residents, in which interactive and animated computer programs could be applied. Improvements are possible in accuracy and in scope. For more accuracy, ligament representation in the models should be improved as a first priority. Scope should be enlarged by including the time-dependent behavior of ligaments, particularly the effects of preconditioning, and by the inclusion of the menisci. However, for validation studies, applications, and improvement of the models, the objectives must be defined precisely in order to obtain useful results within a reasonable time period. This implies that these efforts would be enhanced by the active interest and participation of clinicians. It is interesting to note that most development of computer-simulation models has taken place outside the United States and in the periphery of orthopaedic and sports medicine research. There is no question in my mind, however, that these tools deserve a wider interest.

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