Mathematical simulations of passive knee joint motions

MATHEMATICAL SIMULATIONS OF PASSIVE KNEE JOINT MOTIONS

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1. INTRODUCTION

In view of the complex biomechanical behavior of the knee joint, mathematical simulations of its mechanical behaviour are found to be useful in the scientific process of untwining the complex relationships between the functional biomechanical characteristics and the structural properties. As far as the passive properties are concerned, several three-dimensional models are known. Of the more sophisticated models (1,2) only few results are known. The purpose of the present investigation is to measure and describe the passive freedom-of-motion characteristics of the human knee joint, on a subject-to-subject basis. The approach is to obtain experimentally the passive motion characteristics of the joint for various combinations of external loads and to simulate this behavior in a mathematical model. The geometric properties of the individual joint specimens are measured and are used as input for the model.

This paper describes the general characteristics of the model and some results of parameter studies. In addition, the effect of malpositioning an anterior cruciate (ACL) substitute is determined, in order to illustrate the advantage of using a mathematical model for the evaluation of surgical procedures. This question of the possible effects of malpositioning an ACL substitute is an actual one, since Hefzy and Grood (6) found a change of the insertion location of an ACL substitute to have a large influence on the length patterns of the ACL fiber bundles. For one specific flexion motion they choose different tibial and femoral attachment locations and calculated what the length patterns would be if a fiber bundle was located between variable attachment locations. In this study this experiment was repeated with the model, but in this case the effects of alternative insertions on the motion parameters were taken into account as well.

2. METHODS

The mathematical model used is based on the work of Wismans et al. (1) and is adapted to accomodate for extensive parameter variations (3). The model calculates the equilibrium position of two rigid bodies connected by nonlinear spring elements and contact points, while kinematic constraints and external loads are imposed.

The attachment locations of the ligaments on the tibia and femur were measured on the experimental specimen (4) and determine the attachments of the line elements in the model. The strain-force relation of the spring-type line elements is described by (1):

$$f(\varepsilon) = K\varepsilon^2/\varepsilon_1 \qquad 0 \leq \varepsilon \leq 2$$

$$f(\varepsilon) = K(\varepsilon - \varepsilon_1) \qquad \varepsilon > 2\,\varepsilon_1$$

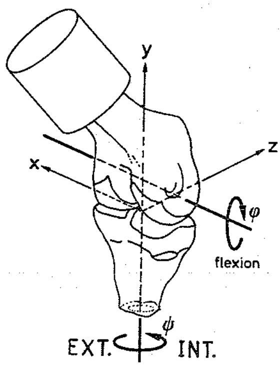
in which f $=$ ligament force
K, ε_l $=$ material parameters
ε $=$ ligament strain

The material parameters are derived from literature data. Since the initial strain at the reference extension position ε_r is unknown, this parameter is approximated by comparing the model findings with the experimental results.

The articular surfaces are considered to be rigid. The geometry of the surfaces was measured from the experimental specimen (5). The coordinates of points on the surfaces are used to fit 3-D polynomials. Since the femoral posterior condyles can be approximated by spheres, a polynomial fit in spherical coordinates is applied. The simplest geometrical approximation is by fitting planes on the tibial points and spheres on the femoral points. Increasing the polynomial degree will increase the quality of the fits, but will introduce possible problems in maintaining the conditions for point contact. Therefore, a method is introduced to fit low degree polynomials on those surface points that are closest to the contact points. This means that after every calculation of the equilibrium position of the joint, for a given set of loading conditions and kinematic constraints, a new polynomial surface is obtained and a new equilibrium position is calculated.

In accordance with the experimental protocol (7), subsequent flexion positions, from extension to 90 degrees flexion, are imposed for a given loading configuration. The resulting kinematic parameters and the ligament lengths can be compared to the experimental data. Results which cannot be retrieved from the experiments are obtained in the model, i.e. the ligament strains, the ligament forces, the contact point locations and the contact forces.

The parameter study involves a variation of the ligament properties and the geometrical descriptions. Since it is not possible in the scope of this paper to extensively document the model characteristics, this paper focusses on the envelope of passive joint motion, as it was also obtained from the experiments: the internal and external tibial rotation as functions of flexion determined with tibial torques of respectively +3 and -3 Nm (Fig.1) (7).



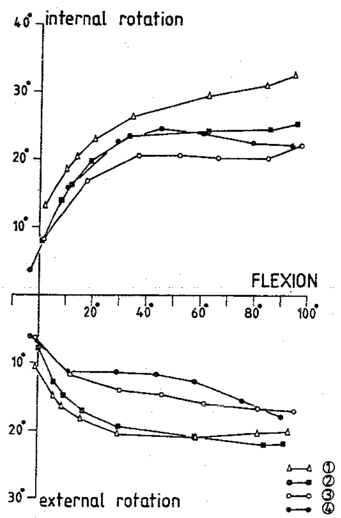


FIGURE 1:
The envelopes of passive knee joint motion of four specimens: internal rotation with a torque of 3 Nm and external rotation with a torque of -3 Nm as functions of flexion.

The influence of a variation of the femoral ACL attachment on the ligament length patterns is studied for a neutral pathway (no tibial torques). First non-functional phantom line elements are defined which are attached on the femur 5 mm more anterior and 5 mm more posterior, and the resulting length patterns are calculated, in fact repeating the experimental procedure of Hefzy and Grood (6). Secondly, the femoral attachments of the line elements representing the ACL are actually replaced anteriorly and posteriorly, and the motion simulation is repeated. Thus influencing the length patterns as well as the motion parameters.

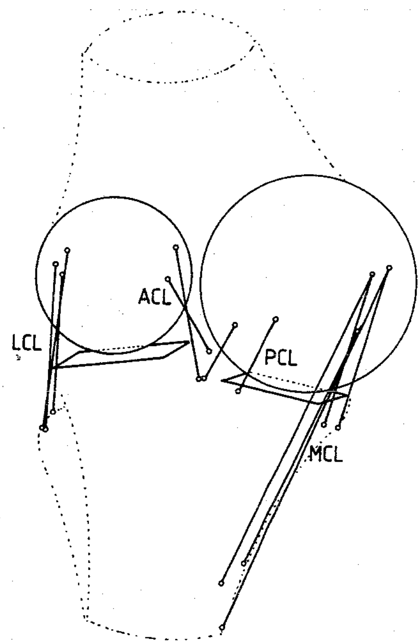


FIGURE 2:
The femoral surfaces approximated by spheres, the tibial surfaces by planes. The line elements represent the ligaments.

3. RESULTS AND DISCUSSION

The geometrical description of the joint surfaces in the model used for the following parameter analysis consists of two spheres for each of the femoral condyles and two planes for the tibial surfaces (Fig.2). The medial collateral ligament is represented by three line elements (ant: K= 4000, ε_r= -.03; inf: K= 4000, ε_r= 0.0; post: K= 4000, ε_r= .06), the posterior oblique by two (ant: K= 500, ε_r= .03; post: K= 500, ε_r= .03), the lateral collateral by three (ant: K= 1000, ε_r= .04; sup: K= 1000, ε_r= .03; post: K= 1000, ε_r= .04), the anterior cruciate by two (ant: K= 1500, ε_r= -.04; post: K= 1500, ε_r= 0.075) and the posterior cruciate ligament by two line elements (ant: K= 2250, ε_r= -.035; post: K= 2250, ε_r= .045). Decreasing the stiffness parameter K and decreasing the initial strain of all ligaments has the same effects on internal and external rotation, increasing rotatory laxity in the model (Fig.3). Increasing the stiffness or the initial strain has a reversed effect. Changing the overall stiffness and the overall initial strains of the ligament does not dramatically change the motion characteristics. Altering the coordinates of all the ligament attachments with respect to the geometry of the articular surfaces does have a large effect, particularly for changes in the x- and z-coordinates (Fig.4).

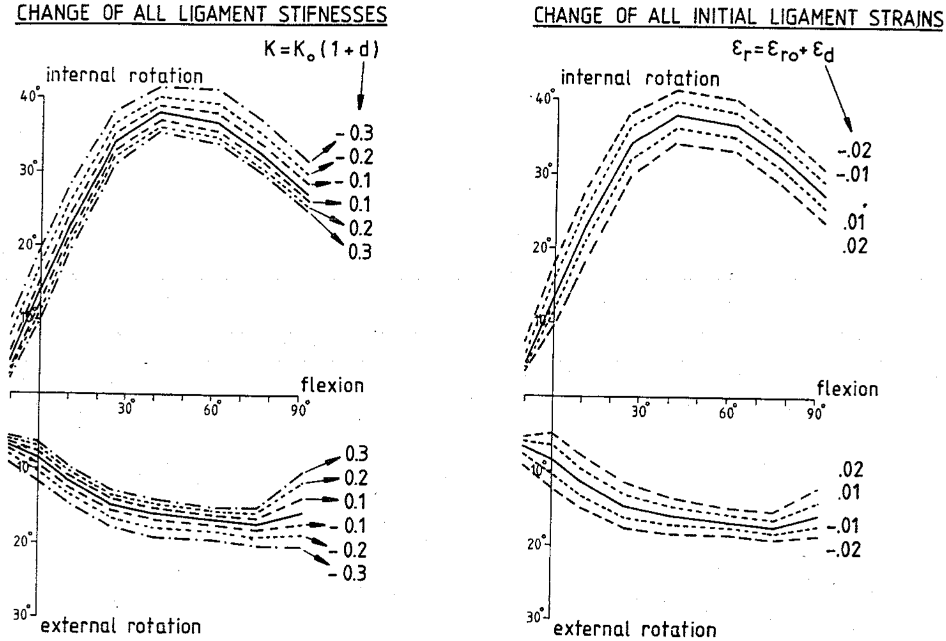


FIGURE 3:
The envelope of motion calculated with the spheres-and-plane-model. Variations of the ligament parameters. Stiffness K and reference strain ε_r being changed for all ligaments.

An internal motion pathway, i.e. flexion with an internal torque of -3 Nm, in case the surfaces (Fig.5) are approximated by more realistic local fits close to the contact points is shown in Fig. 6, compared with the experimentally obtained data. The decrease in internal rotation after 40 degrees flexion, which is evident in the spheres-and-plane model, is not present. A general improvement of the agreement between the experimental results and the model calculations is obtained after introducing this improved approximation of the articular surfaces.

The effects of a femoral attachment variation of the ACL on the length patterns of this ligament are shown in Fig.7. The experiments of Hefzy and Grood were actually reproduced in case the length patterns are calculated of ACL line elements that are attached 5 mm anterior or 5 mm posterior on the femur as compared to the original insertion. However, when the attachments are actually altered the length patterns are less affected (Fig.7.b), because the variation is partially absorbed by alternative motion characteristics.

This example shows that a mathematical model is a versatile tool for analyzing experimental results and surgical procedures, i.e. cruciate ligament reconstruction. However, the assumption of the articular surfaces being rigid is a critical one, since changes in surface representation do have a considerable influence on the model characteristics. Therefore, future developments of this mathematical knee joint model have to take into account the non-rigidity of the articular surfaces.

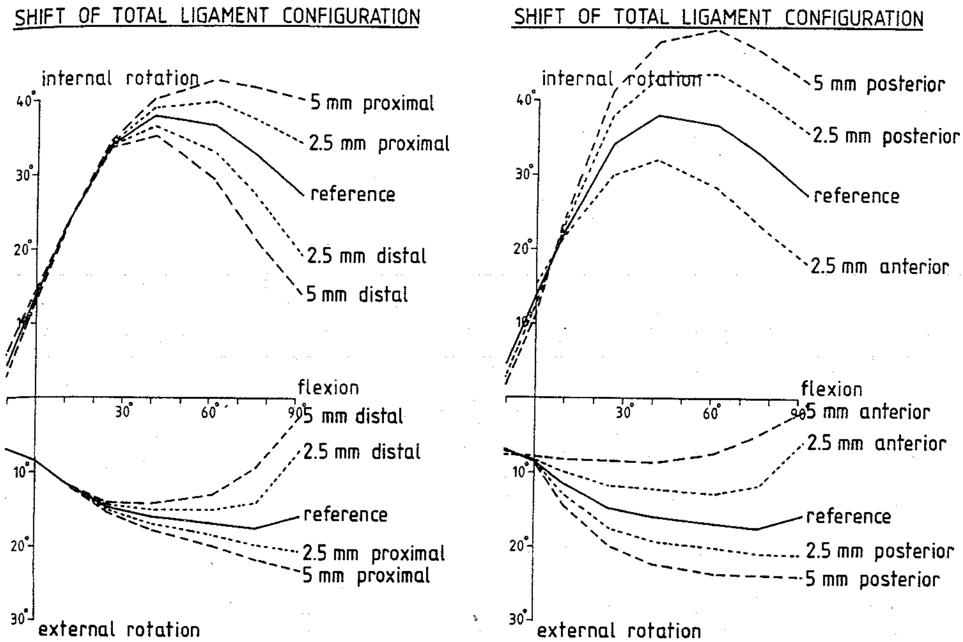


FIGURE 4:
Spheres-and-plane-model. Variations of the coordinates of the entire ligament configuration with respect to the articular surface geometry in the reference position of the joint (extension)
a. proximal and distal shift
b. anterior and posterior shift

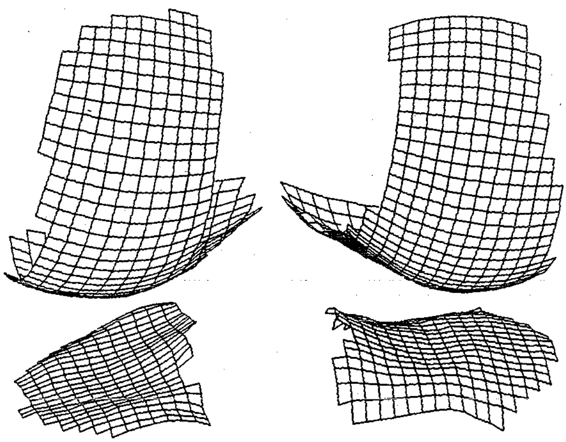


FIGURE 5: Representation of the articular geometry in case a local fit is applied.

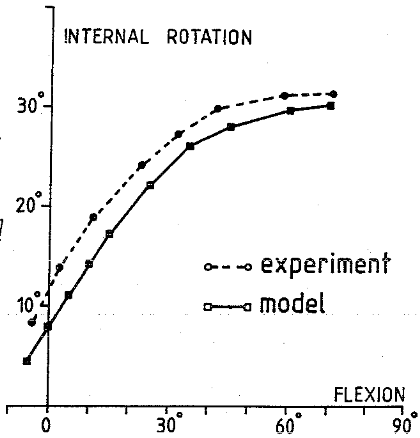
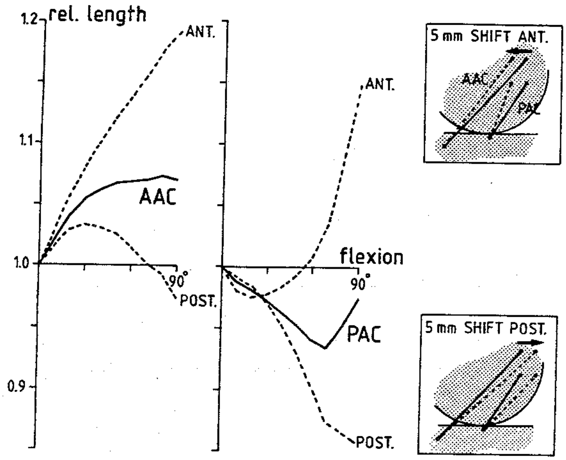
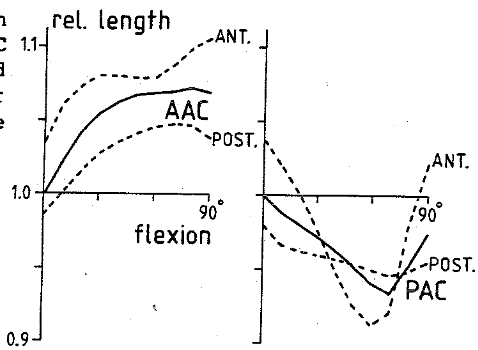


FIGURE 6: Internal motion pathway obtained with the local-fit-model compared with the experimental data.

FIGURE 7:
Relative length patterns of the anterior part of the anterior cruciate (AAC, N1) and the posterior part of the anterior cruciate (PAC, N2).
a) relative length of line elements with a 5 mm more anterior or posterior attachment on the femur.



b) relative length in case the AAC and PAC are actually attached 5 mmmore anterior or posterior on the femur.



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ACKNOWLEDGEMENT

This investigation is performed in co-operation with the dept. of Mechanical Engineering, Eindhoven Technical University and is partly sponsored by the Netherlands Organization for the Advancement of Pure Research (ZWO) grant 90-90.