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CHARACTERIZATION OF THE MECHANICAL BEHAVIOR OF HUMAN KNEE LIGAMENTS: A NUMERICAL-EXPERIMENTAL APPROACH

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Abstract—During knee-joint motions, the fiber bundles of the knee ligaments are nonuniformly loaded in a recruitment pattern, which depends on successive relative orientations of the insertion sites. These fiber bundles vary with respect to length, orientation and mechanical properties. As a result, the stiffness characteristics of the ligaments as a whole are variable during knee-joint motion. The purpose of the present study is to characterize this variable mechanical behavior. It is hypothesized that for this purpose it is essential to consider the ligaments mechanically as multi-bundle structures in which the variability in fiber bundle characteristics is accounted for, rather than as one-dimensional structures. To verify this hypothesis, bone-ligament-bone preparations of the ligaments were subjected to series of unidirectional subfailure tensile tests in which the relative insertion orientations were varied. For each individual test specimen, this series of tensile tests was simulated with a mathematical ligament model. Geometrically, this model consists of multiple line elements, of which the insertions and orientations are anatomically based. In a mathematical optimization process, the unknown stiffness and recruitment parameters of the line elements are identified by fitting the variable stiffness characteristics of the model to those of the test series. Thus, lumped parameters are obtained which describe the mechanical behavior of the ligament as a function of the relative insertion orientation. This method of identification was applied to all four knee ligaments. In all cases, a satisfactory fit between experimental results and computer simulation was obtained, although the residual errors were lower for the cruciate ligaments (1.0–2.4%) than for the collateral ligaments (3.7–8.1%). It was found that models with three or less line elements were very sensitive to geometrical parameters, whereas models with more than 7 line elements suffered from mathematical redundancy. Between 4 and 7 line elements little difference was found. It is concluded that the present ligament models can realistically simulate the variable tensile behavior of human knee ligaments. Hereby the hypothesis is verified that it is essential to consider the ligaments of the knee as multi-bundle structures in order to characterize fully their mechanical behavior.

Keywords: Knee; Ligaments; Mechanical properties.

INTRODUCTION

Much attention has already been paid to the mechanical properties of human knee ligaments. Both the structural properties, or mechanical properties of the ligament as a structure, and the material properties, or mechanical properties of the material of which the ligaments are made, have been evaluated. In experimental studies, the structural properties have been quantified by unidirectional tensile tests, in which the bone-to-bone angle is fixed relative to the axis of pull (Butler *et al.*, 1978; Kennedy *et al.*, 1976; Noyes and Grood, 1976). From the force-displacement registrations, a number of parameters has commonly been calculated, such as linear stiffness, maximal force and maximal displacement. While some mathematical models for evaluating the role that ligaments play in the mechanics of the knee do not utilize the load-deformation characteristics of the ligaments at all (Zavatsky and O'Connor, 1992), in those that do, the shape of the whole nonlinear force-displacement curve should be emphasized (Butler *et al.*, 1978; Huiskes,

1992; Woo and Adams, 1990). For that purpose, mathematical formulations have been derived (Carlstedt and Skagervall, 1986; Elden, 1968; Viidik, 1980). The unknown parameters of these constitutive model equations have been determined by fitting the predicted curves to the experimentally obtained ones by methods of minimal least squares. The combination of experiments and model simulations for quantitative determinations of unknown model parameters is indicated by the 'identification method' (Norton, 1986). This method has been applied earlier in the biomechanics field (Hendriks, 1991; Lin *et al.*, 1978; Oomens *et al.*, 1993; Yettram and Vinson, 1979).

In the traditional approach of ligament testing, uniform strain across the ligament is often implicitly assumed. However, for the posterior cruciate ligament as well as for the anterior cruciate ligament it has been shown that the material properties differ between fiber bundles (Race and Amis, 1994; Butler *et al.*, 1992) as well as their lengths and orientations (Butler *et al.*, 1989). During knee motions, these fiber bundles are non-uniformly loaded in a sequence which depends on the relative orientations of the insertion sites. Because of these variabilities between the fiber bundles in the ligaments, the mechanical behavior of the ligament as

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a whole is variable (Figgie *et al.*, 1986; Mommersteeg *et al.*, 1995; Rogers *et al.*, 1990; Woo *et al.*, 1991). It is obvious that tests in which the ligament is pulled in only one direction do not provide sufficient information to characterize mathematically these nonuniform loading patterns of fiber bundles during knee motion. To improve the description of the mechanical behavior of the ligaments in present knee-joint models (Blankevoort and Huijskes, 1991; Essinger *et al.*, 1989), a re-examination of the traditional way of tensile testing and related mathematics is required.

In this paper an alternative concept of a ligament structure is proposed. The ligaments are modeled as three-dimensional structures of more than one force transferring element of variable orientations, lengths and mechanical properties. To define the parameters of this so-called multi-line-element model, the ligaments were tensile tested in several different test configurations. The geometry of the model, i.e. the positions of the line elements, are based on the actual fiber bundles in the ligaments. The hypothesis is that with this anatomically more realistic model in which the fiber bundles are the tensile units, the predictability of the functional behavior of the ligaments in different relative insertion orientations, i.e. in knee-joint motion, is improved as compared to the traditional approach, in which the whole ligament is considered as the tensile unit. To test this hypothesis, ligament models consisting of multiple bundles were defined of the medial collateral ligament (MCL), the lateral collateral ligament (LCL), the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL). With these models subfailure ligament tests performed in different relative insertion orientations, were simulated.

METHODS AND MATERIAL

The identification method was applied to six ligaments obtained from three human knee joints of which two ACLs (ages 67 and 71), two PCLs (ages 63 and 71), one MCL (age 71) and one LCL (age 71). The protocol was (a) to record the ligament forces for different relative orientations and positions of the insertion sites in a series of tensile tests, (b) to record the anatomy of the insertion sites, (c) to construct a model of the ligament, using the experimentally determined kinematics and anatomy as input, (d) to obtain initial estimates of the model parameters from the experimental raw data, and finally (e) to minimize the ligament force differences between model and experiment in order to calculate the model parameters.

Tensile testing (a)

The ligaments were prepared as bone–ligament–bone preparations from the knee joints. The bones accompanying the ligaments were marked with minimally five tantalum pellets (0.5 mm diameter) each. Then, the bones were embedded in polymethylmethacrylate (PMMA) such that, by visual examination, as many ligament fibers were tensed as possible. In this initial orientation of the

bones (init.), the bone–ligament–bone preparations were fixed in a specially designed loading rig in a material testing machine (MTS, Berlin, Germany). A complete description of the loading rig is given by Mommersteeg *et al.* (1995). A device was attached between the pot containing the femoral bone block and the actuator of the testing machine. This device enabled tilting of the femoral bone-block around two perpendicular axes, a medial–lateral and an anterior–posterior axis, which intersected approximately in the center of the femoral insertion site of the ligament. Different tilting positions were achieved by successively lifting the anterior, posterior, medial and lateral sides of the ligament in steps of minimally 5°, up to a maximum of 25°. These successive orientations of the bones were called the anterior (A₁ through A₅), posterior (P₁ through P₅), medial (M₁ through M₅) and lateral tilt direction (L₁ through L₅), respectively. In each orientation of the bones, ten subultimate loading cycles were performed to obtain reproducible force–displacement characteristics (preconditioning). The actuator displacement applied was about 7 and 10% of the initial length, as estimated with the aid of calipers, for the collateral and the cruciate ligaments, respectively. Based on this estimated initial length, the strain rate was 66.6% s⁻¹. At the end of all series, the first test was repeated to detect possible tissue damage. The ligament was eliminated from the study, if remarkable differences between the two tests exist (Mommersteeg *et al.*, 1995). The forces were measured with a force transducer, while the relative orientations and displacements of the bones were measured with Röntgen Stereophotogrammetric Analysis (RSA, Selvik, 1974). All these force–displacement curves together represent the relationship between the ligament force and the kinematics of the bones. Moistening of the tissue occurred through spraying the ligaments with saline before and after each test. Between the tensile tests and during the preparation process the ligaments were wrapped in saline-moistened gauze, which was remoistened frequently.

Anatomy of the insertion sites (b)

The anatomy of the origins and insertions of different ligament bundles was determined in two different ways. The first method, in which RSA was used, was applied for the two ACLs. The origins and insertions of eight superficial fibers were marked with additional tantalum pellets (0.8 mm diameter) and RSA was performed to obtain the three-dimensional coordinates of these markers. Two of these superficial fibers represented the margins between the anteromedial and posterolateral bundle, one of them the margins between the anterior and posterior part of the anteromedial bundle (AMBa and AMBp, respectively) and one of them the margins between the anterior and posterior part of the posterolateral bundle (PLBa and PLBp, respectively; Buck, 1985). The other four pellets marked fiber insertions in between these margins to obtain a more appropriate outline of the ligament insertion sites. On the basis of these peripheral markers, the insertion sites were geometrically subdivided into the inser-

tion sites of the anterior and posterior part of both the anteromedial bundle and the posterolateral bundle [Fig. 1(a)].

In a second way to determine the three-dimensional coordinates of the insertion sites the 3Space Isotrak system (Polhemus Navigation Sciences, Colchester, VT, U.S.A.) was used according to Sidles *et al.* (1988). The cemented bone–ligament–bone preparations were mounted in a perspex support. The outlines of the insertion sites of the ligament were measured three times. Subsequently, the bundles were separated bundle-by-bundle from the rest of the ligament, being very careful to follow the fiber orientations. Then, also the outlines of both the insertion sites of these bundles were measured in triplicate. This method was applied for one of the ACLs (ACL2), the two PCLs, the LCL and the MCL. The ACL was divided into 9 bundles [Fig. 1(b)], the PCL into 6 (PCL 1) and 7 bundles (PCL 2), the MCL into 4 bundles and the LCL into 3 bundles. To transpose the three-dimensional coordinates of the insertion sites measured relative to the Isotrak coordinate system, to the RSA coordinate system, the perimeter of the insertion sites was measured with the aid of the Isotrak system as described above, as well as with the aid of RSA. Therefore, additional tantalum pellets (0.8 mm diameter) were

placed around the insertion sites of the ligament and the RSA and the Isotrak coordinate systems were chosen to be parallel. Subsequently, to transpose the measured insertion sites to the initial orientation of the bones in the material testing machine, a coordinate transformation was performed, using the 0.5 mm pellets in the bones.

Ligament model (c)

Based on the anatomy of the insertion sites, measured as described in the preceding section, a mathematical model of each individual ligament was constructed. The models consist of nonlinear elastic line elements, representing the bundles defined in the ligament with the aid of RSA (Fig. 2) and Isotrak (Fig. 3). The centroids of the measured bundle insertions are the insertions of the line elements. The model describes the elastic behavior of the ligament as follows.

Because the line elements are straight lines, their length patterns were completely determined from the coordinates of their insertion sites in the initial orientation of the bones and the experimentally determined three-dimensional kinematics expressed by the translation vector \mathbf{D} and the rotation matrix R . The length L_j of a line element j was calculated from the insertion points \mathbf{X}_{ij} on

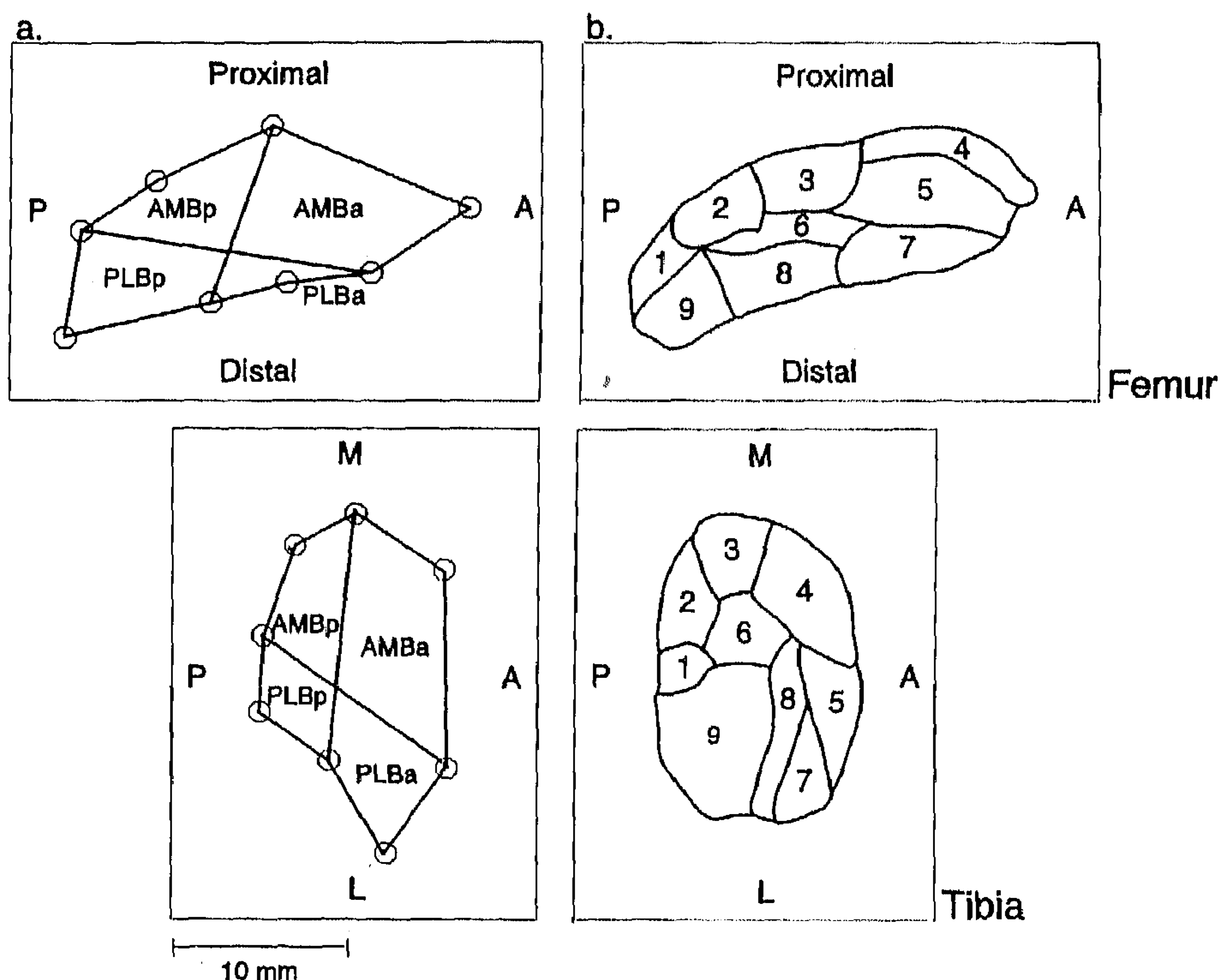


Fig. 1. The femoral and tibial insertion sites of bundles defined in the ACL 2 with two different methods. (a) The femoral and tibial insertion sites of four bundles of ACL 2 based on RSA of the insertion sites of eight superficial fibers (o). On the basis of these measurements, the origin and insertion of the ligament were geometrically subdivided into four areas, representing the insertion sites of the anterior and posterior part of the anteromedial bundle (AMBa and AMBp) and of the posterolateral bundle (PLBa and PLBp). (b) The femoral and tibial insertion sites of nine bundles of ACL 2 defined with the aid of the Isotrak system. Bundles 4 and 5 correspond approximately with the AMBa, bundles 2, 3 and 6 with AMBp, bundles 1 and 9 with PLBp and bundles 7 and 8 with PLBa. A: anterior; P: posterior; M: medial; L: lateral.

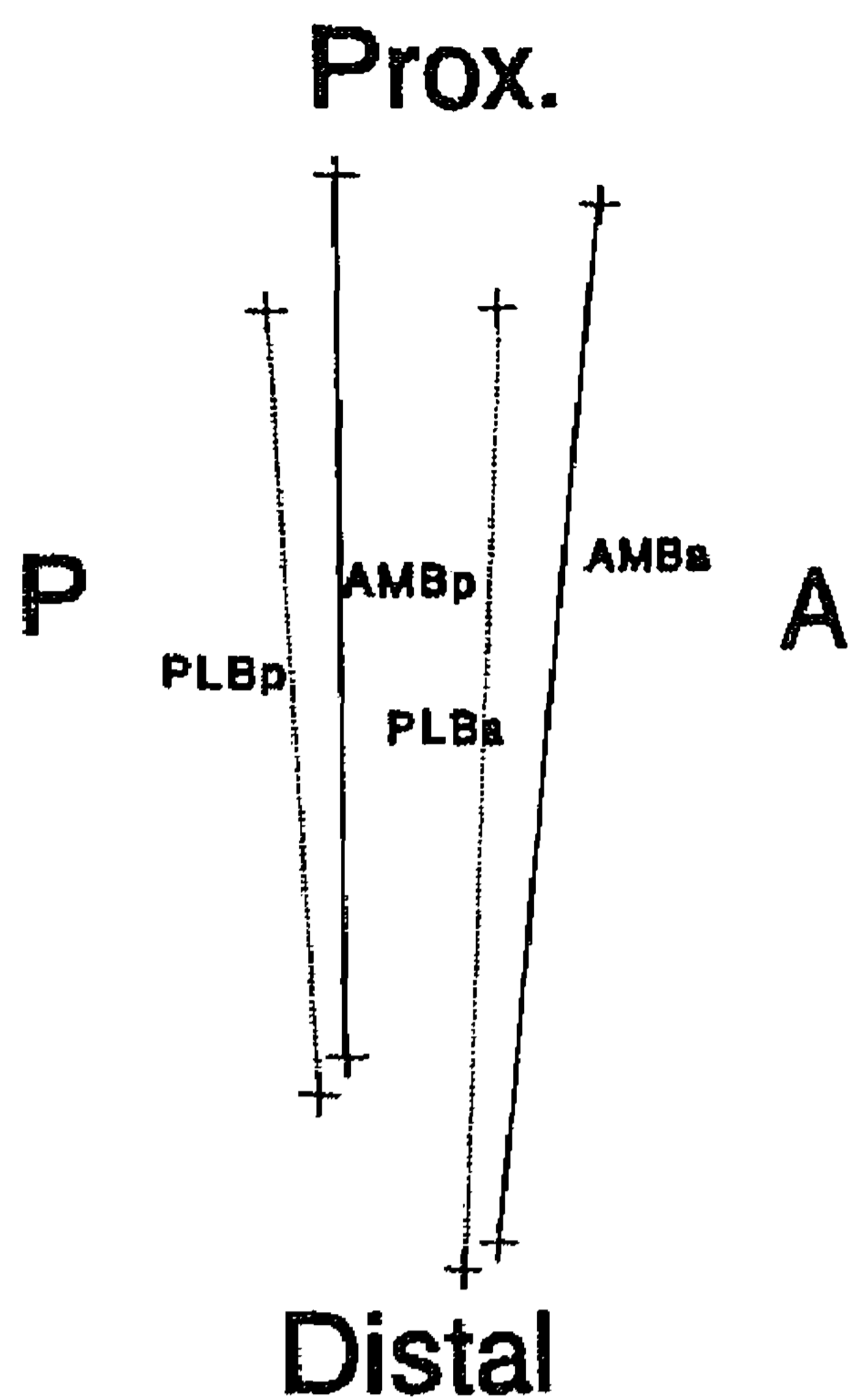


Fig. 2. A medial view of the four-line-element model of ACL 2 based on RSA measurements (Fig. 1). Legends: AMBa and AMBp: anterior and posterior part of the anteromedial bundle; PLBa and PLBp: anterior and posterior part of the posterolateral bundle. Prox: proximal; A: anterior; P: posterior.

the tibia relative to the tibial coordinate system and x_{fj} on the femur relative to the femoral coordinate system by

$$L_j = |X_{tj} - D - Rx_{fj}|.$$

The tensile force F_j in a line element j was assumed to vary according to the square of the strain ε_j , or relative elongation, of this element (Crowninshield *et al.*, 1976; Grood and Hefzy, 1982; Wismans *et al.*, 1980; Elden, 1968) as

$$F_j = k_j \varepsilon_j^2, \quad \varepsilon_j > 0,$$

$$F_j = 0, \quad \varepsilon_j \leq 0,$$

in which k_j is a stiffness parameter and ε_j is the strain in line element j calculated from its actual length L_j and zero force length L_{0j} according to

$$\varepsilon_j = (L_j - L_{0j})/L_{0j}.$$

'Strain' should not be considered as real tissue strain in this context, but rather as an averaged strain over a line element length. As the bundles did not interact with each

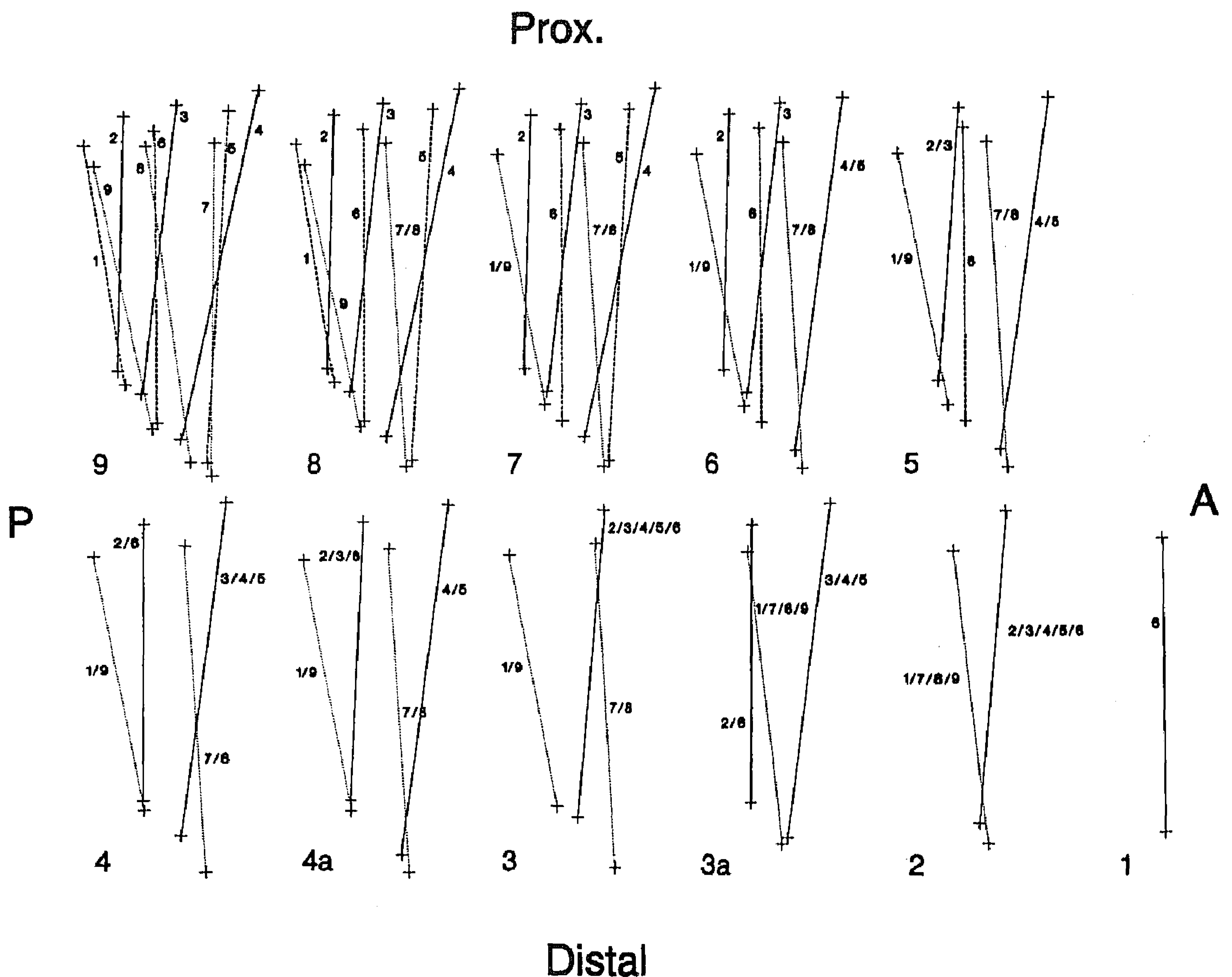


Fig. 3. Medial views of the nine-line-element model of ACL 2 based on Isotrak measurements as well as of the derivations of this model. The continuous line elements are positioned medially, the dotted ones laterally, and the dashed ones centrally. Note that two different three-line-element models [3 and 3(a)] as well as two different four-line-element models [4 and 4(a)] are defined. Prox: proximal; A: anterior; P: posterior.

other in the model, the total ligament force F_1 was found by summing the individual bundle forces according to

$$F_1 = \sum F_j v_j,$$

where v_j is the unit vector pointing along the line of action of the force in the line element j .

Initial estimates of model parameters (d)

To obtain a reasonable starting point for the optimization process, rough estimates of the model parameters L_{0j} and k_j were obtained. The initial value for the parameter L_{0j} was estimated from the maximal L_j occurring during the experimental test series, while a zero force was measured [$L_{0j}(\text{est})$]. The parameter k_j was initially estimated [$k_j(\text{est})$] using the ligament stiffness k' (N mm^{-1}) determined in the experiment at a strain level ε of 6.5% (Mommersteeg *et al.*, 1995), equally divided over the number (N) of line elements defined, according to

$$k_j(\text{est}) = \frac{k' L_{0j}(\text{est})}{2N\varepsilon}.$$

Using the estimated model parameters and applying the appropriate kinematics, the ligament forces were calculated for the different relative orientations and positions of the bones and compared with the experimental measurements.

Optimization technique (e)

Subsequently, the differences between the experimentally determined ligament forces for each relative orientation and position of the bones and the corresponding predicted values with the model were minimized by adjusting the model parameters k_j and L_{0j} . The optimization function applied for nine equally distributed points of each experimentally determined force-displacement curve, was represented by the residual error vector

$$E(\mathbf{k}, \mathbf{L}_0) = \mathbf{P}_m(\mathbf{k}, \mathbf{L}_0) - \mathbf{P}_e,$$

in which the vector \mathbf{P}_e contains the experimentally determined ligament forces and the vector $\mathbf{P}_m(\mathbf{k}, \mathbf{L}_0)$ the results of the equivalent model calculations for a given set of the model parameters contained in the vectors \mathbf{k} and \mathbf{L}_0 . This function was minimized following a modification of the Levenberg-Marquardt algorithm by using the general least-squares solver LMDIF (from MINPACK (Moré *et al.*, 1980), Argonne National Laboratory, Argonne Il, U.S.A.). This minimization resulted in the identified values for k_j and L_{0j} of each line element. To evaluate the effects of the initial values of k_j and L_{0j} on the solutions of the optimization process, the process was started with different initial values.

Data analysis

The adequacy of the match between model and experiment was expressed as the magnitude of the residual error vector (= the root mean square error). The effects of the optimization process were evaluated by comparing the root mean square error before and after optimization of the model parameters. To study the effect of the

number of line elements on the adequacy of the fit between model and experiment, the tensile tests of ACL 2 were simulated with models consisting of a number of line elements varying from 1 to 9 (Fig. 3). These line element models were generated on the basis of the nine-line-element model. Two or more neighboring line elements of this model were concentrated into one line element. Though the bundle anatomy of the ACL according to Buck (1985) was taken into account, arbitrary choices had to be made in the process of combining line elements. Two different three- and four-line-element models were defined. To compare the two ways of measuring the anatomy of the insertion sites, the RSA-based four-line-element model (Fig. 2) and the Isotrak based four-line-element model of ACL 2 [Fig. (3a)] were compared. To be able to compare the stiffness parameters obtained with literature values, the stiffness parameter k_j (N) was converted into the tangent k'_j of the force-displacement characteristic at a strain level ε of 6.5% (N mm^{-1}) defined as

$$k'_j = 2k_j \frac{\varepsilon}{L_{0j}}.$$

RESULTS

The effects of the optimization procedure on the similarity between the model and the experimental data are obvious (Table 1). Before optimization, the root mean square errors between the calculated ligament forces and their corresponding measured values were high, indicating that a poor description of the experimental results was obtained with the estimated model parameters. Optimization of the model parameters k_j and L_{0j} for each line element resulted in a reduction of the residual error and thus in a considerable improvement of the fit between model and experiment. For the RSA based model of ACL 2, for example, an eight times reduction of the root mean square error was obtained (from 12.9 to 1.6% of the maximal force). For a root mean square error of 1.6% of the maximal force, the fit between model and experiment was adequate, i.e. the different force-displacement characteristics obtained in the experiment could be reproduced with the model (Fig. 4). Also for the other ligaments an adequate fit of models and experiments could be obtained after optimization of the model parameters (Table 1). However, for the medial and lateral collateral ligaments the fit was worse than for both cruciate ligaments.

It is apparent that the identified zero force lengths of the line elements do not deviate much from the estimated values, while the differences between the estimated and identified k -values are clear (Table 2). The reduction of the residual error after optimization of the model parameters was thus mainly achieved by a change in the k -value. Note that in the MCL model (Table 2) as well as in the eight- and nine-line-element models of ACL 2 (Fig. 5) line elements obtained a zero k -value, independent of the initially estimated value.

Table 1. The root mean square errors (RMSerror) between the measured and the calculated ligament forces before and after optimization in terms relative to the maximal measured force (%); $N(\text{exp})$ is the number of tensile tests of the ligament times the number of points at each registered curved used for the fit and N is the number of defined line elements

Ligament	RSA/ Isotrak	N	$N(\text{exp})$	RMSerror (% max. force) before optimization	RMSerror (% max. force) after optimization
ACL 1	RSA	4	117	13.1	2.3
ACL 2	RSA	4	144	12.9	1.6
	Isotrak	9	144	15.8	1.3
PCL 1	Isotrak	6	126	15.0	2.4
PCL 2	Isotrak	7	162	14.2	1.0
MCL	Isotrak	4	108	28.0	8.1
LCL	Isotrak	3	108	18.1	3.7

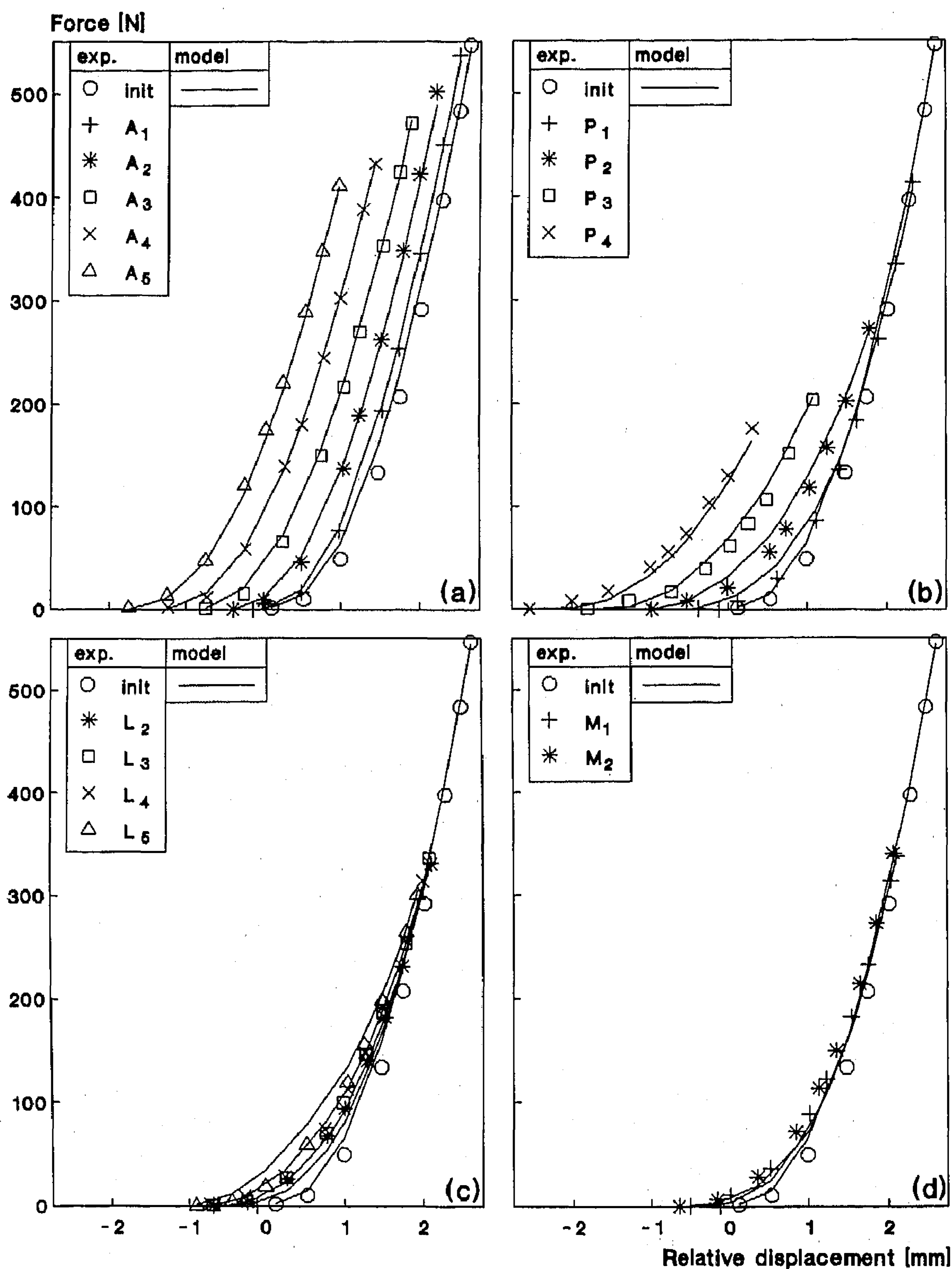


Fig. 4. The force-displacement curves of the ACL 2 in the initial orientation of the bones (init.) and of the (a) anterior (A₁-A₅), (b) posterior (P₁-P₄), (c) medial (M₁-M₂) and (d) lateral tilt direction (L₂-L₅) calculated with the model of Fig. 2 (continuous lines) and measured in the experiment (signs). The displacement is expressed relative to the position of the femoral insertion site in the initial orientation of the bones.

Table 2. Estimated (est) and identified model parameters zero force length L_0 (mm) and k (N) for the different line elements defined in each ligament as well as k' which is defined as $2ke/L_0$ with $\epsilon = 0.065$

Ligament	Line element	L_0 (est) (mm)	L_0 (mm)	k (est) ($\times 10^3$ N)	k ($\times 10^3$ N)	k' (N mm $^{-1}$)
ACL 1	AMBa	48.4	48.7	35.5	81.2	217
	AMBp	41.6	41.2	30.5	39.5	125
	PLBa	38.9	38.5	28.5	19.9	67
	PLBp	31.3	31.5	22.9	32.7	135
ACL 2	AMBa	37.0	36.9	25.5	64.5	227
	AMBp	30.8	30.6	21.2	12.6	54
	PLBa	34.1	33.5	23.4	15.7	61
	PLBp	27.4	27.1	18.8	13.0	62
PCL 1	1	45.8	46.9	18.9	21.4	59
	2	40.9	41.2	16.9	48.5	153
	3	47.0	47.4	19.4	24.4	67
	4	40.4	40.1	16.7	24.0	78
	5	47.7	47.9	19.7	3.2	9
	6	44.1	43.7	18.2	42.8	127
PCL 2	1	38.9	38.3	19.9	11.7	40
	2	37.6	37.6	19.2	63.3	219
	3	39.7	39.4	20.3	56.0	185
	4	41.2	41.5	21.1	30.0	94
	5	44.9	44.5	23.0	6.8	20
	6	43.4	44.5	22.2	15.9	46
	7	37.3	37.0	19.1	27.4	96
LCL	1	68.5	67.4	29.7	8.9	17
	2	70.4	69.4	30.5	32.3	60
	3	72.1	71.6	31.2	56.5	103
MCL	1	104.1	101.9	29.0	0.0	0
	2	111.1	111.8	31.0	87.4	102
	3	91.9	90.9	25.6	26.0	37
	4	106.8	104.6	29.8	63.3	79

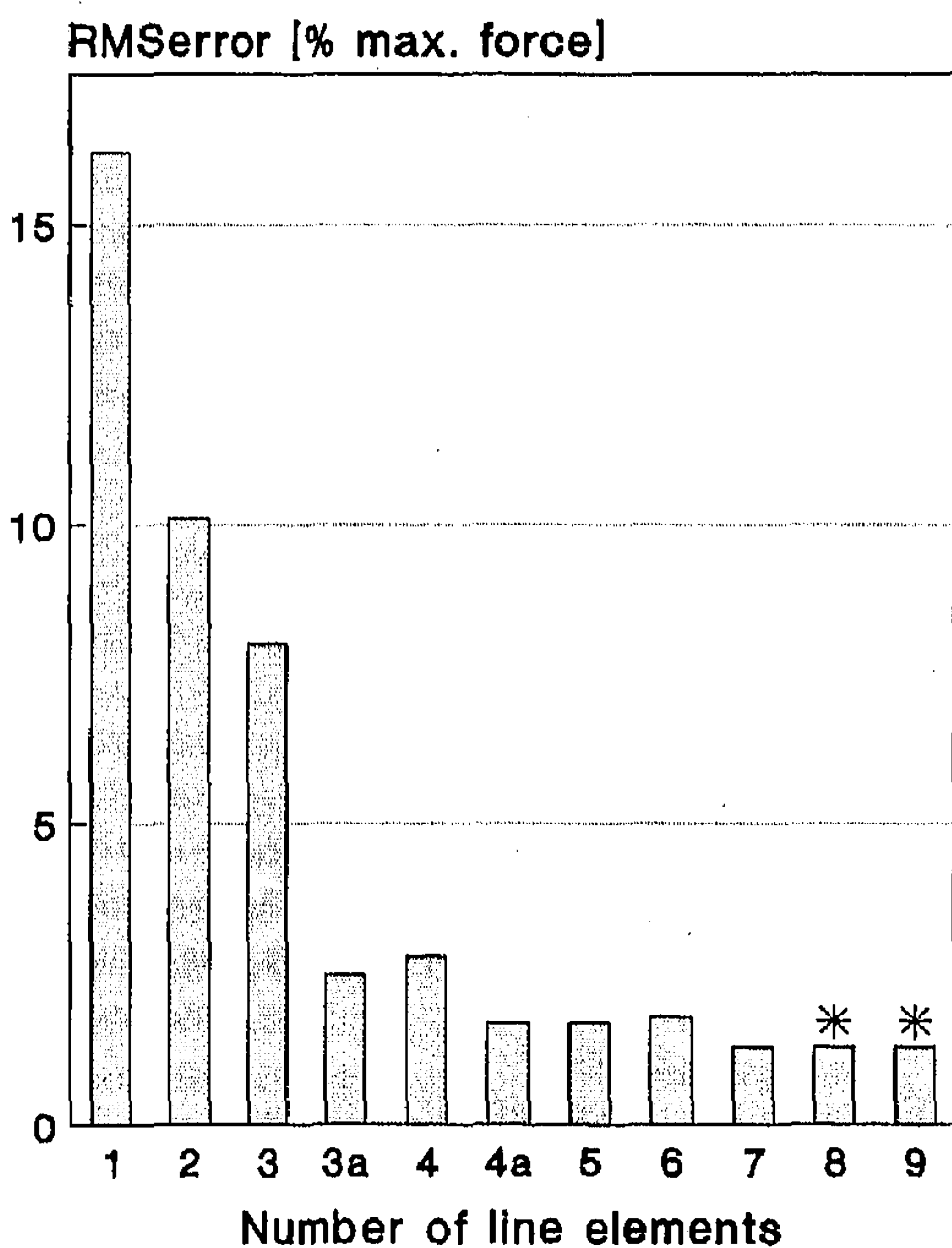


Fig. 5. The root mean square errors relative to the maximal measured force (% max. force) for the various line element models of ACL 2 (Fig. 3). *: one or two line elements were eliminated.

The models were stable in the sense that for wide-ranging realistic initial values of k and L_0 of the line elements, the models converged to the same optimized values. Only in those cases where the starting values were extremely high or extremely low, did the solutions differ. In these physiologically unrealistic cases, the similarity between model and experiment became worse.

The number of line elements affected the adequacy of the fit between model and experimental results. This is illustrated for ACL 2 (Fig. 5). It is obvious that models with 4, 5, 6 or 7 line elements adequately reproduced the experimental results, while models with 1 or 2 line elements did not. The three line element models were very sensitive to the way in which the line elements were positioned anatomically. The eight- and nine-line-element models could describe the experiment well, although one and two line elements, respectively, obtained a zero k -value during the optimization process, leaving seven-line-element models in both cases.

For the several line element models of ACL 2 for which a small root mean square error was obtained after optimization, equal distributions of the resulting model parameters were obtained across the ligament. For all models the line elements with the highest k and L_0 were positioned anteriorly. This is illustrated with force-strain curves for the four- and seven-line-element models based

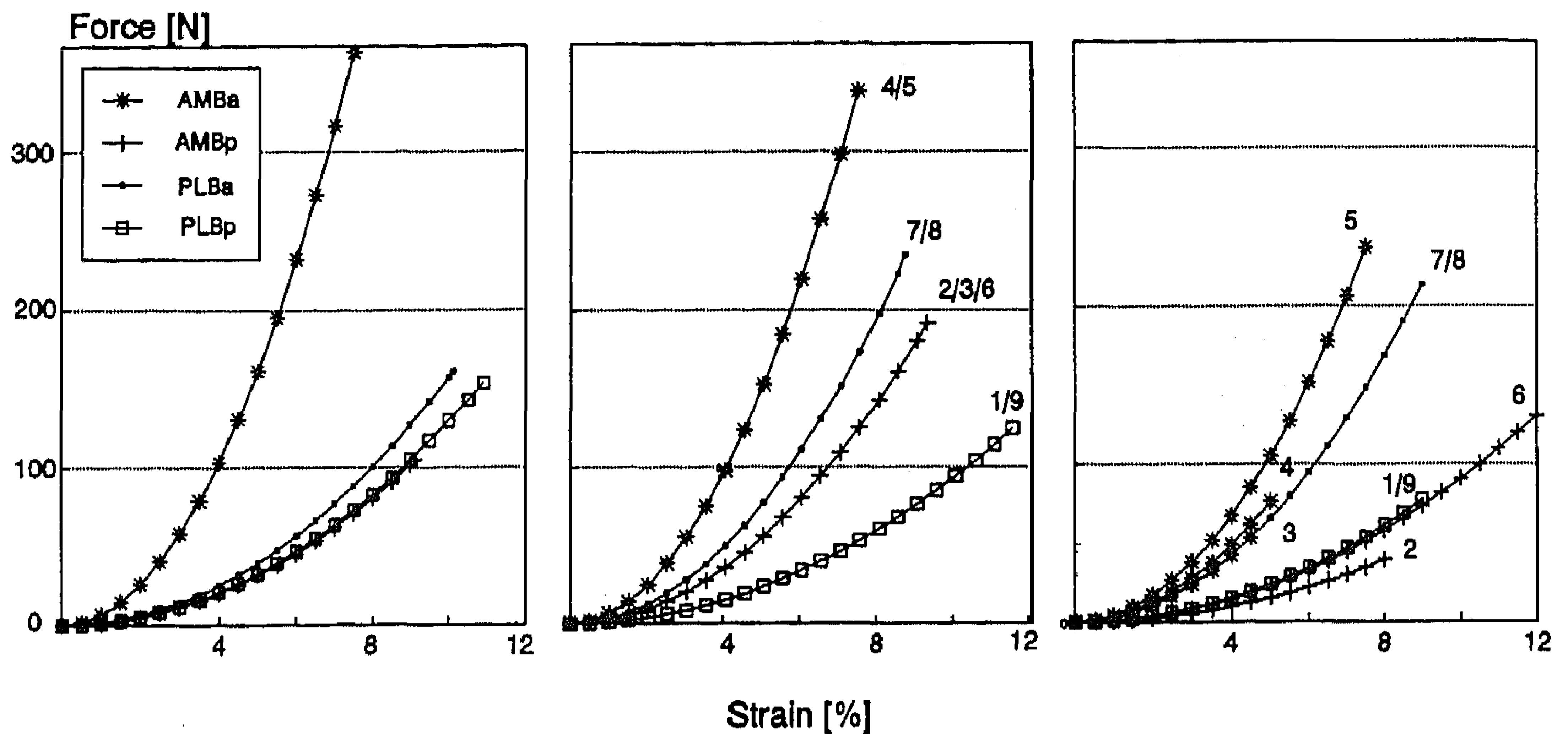


Fig. 6. The force-strain relationships of the three different line-element models of ACL 2. (a) The four-line-element model of ACL 2 depicted in Fig. 2. (b) The four-line-element model depicted in Fig. 3 (4a). (c) The seven-line-element model depicted in Fig. 3 (7). The signs marking the anterior and posterior parts of the anteromedial bundle (AMBa and AMBp) and of the posterolateral bundle (PLBa and PLBp) in (a) correspond to signs marking the line elements approximating these bundles in (b) and (c). Only the ranges of forces experienced by a line element during the experiment is presented.

on Isotrak measurements as well as for the four-line-element model based on RSA measurements (Fig. 6).

The way in which the anatomy of the line elements was measured did not affect the validity of the models. The four-line-element models of ACL 2 defined with RSA and with Isotrak agree with each other both with respect to the model errors (Table 1 versus Fig. 5) and with respect to the model results (Fig. 6).

DISCUSSION

In the present paper, the hypothesis that a multi-bundle concept is essential to characterize fully the mechanical behavior of human knee ligaments, is verified. The characterization is based on the variable recruitment of the fiber bundles. It is aimed at the determination of lumped parameters, which together describe the mechanical behavior of the ligaments as a function of relative insertion orientation. These parameters, which are recruitment parameters, or zero force lengths, and stiffness parameters of line elements, are directly applicable in knee-joint models, in which the ligaments are represented by nonlinear elastic line elements (Blankevoort and Huijkes, 1991; Crowninshield *et al.*, 1976; Essinger *et al.*, 1989). In present knee models, the stiffness parameters are estimated from literature data and are assumed to be equal for the different line elements of each ligament. The recruitment parameters are estimated by minimizing the differences between the experimental and model results with respect to the laxity of the knee in axial rotation (Blankevoort and Huijkes, in press). Although actually complex three-dimensional structures, the ligaments are represented by only one to four line elements, which are not based on detailed anatomical analyses of

the ligament structures. In the present study more detailed specimen-based line element representations of the ligaments were obtained of which the nonuniform stiffness and recruitment parameters were also determined. Supplying the present knee models with these representations, these models are expected to provide an improved understanding of the role the ligaments, and in particular the fiber bundles, play in knee mechanics. Furthermore, it should be possible to study the effects of inhomogeneous mechanical properties, initial tensioning, geometry and number of ligament bundles on the mechanical behavior.

Besides the root mean square error, the small differences between the estimated and calculated zero force lengths of the line elements, notwithstanding the initial values, validate the new approach. The estimated k -values were, however, much smaller than the calculated ones. This can be explained by the fact that the estimated k -values were based on a k -value obtained by whole ligament testing, in which not all fiber bundles were recruited simultaneously.

As in all studies in which a mathematical model is applied to experimental observations, the model parameters can be estimated only adequately from the experiment if the number of observations is attuned to the number of parameters. In cases where the number of observations is too low in relation to the number of model parameters, for example, in case of the MCL and the eight- and nine-line-element model of the ACL 2, line elements were eliminated by the optimization process itself, where they adopt zero k -values. This implies that these line elements were not required to describe the experimental results. On the other hand, the number of model parameters should be high enough to describe the effects of the relative orientations of the insertion sites

sufficiently. This was, for example, not the case for the one- and two-line-element model of ACL 2 for which after optimization still a substantial error between model and experimental results remained.

The fit between model and experiment was worse for the collateral ligaments than for the cruciate ligaments. One possible explanation is that the effects of the relative orientations of the insertion sites on the force-displacement characteristics are much smaller for the collateral ligaments than for the cruciate ligaments (Mommersteeg *et al.*, 1995). A possible solution is to extend the series of tensile tests to higher tilt angles.

An alternative way to deal with the impossibility of uniform loading of the fiber bundles of knee ligaments and the inhomogeneous distribution of material properties throughout the ligaments, is to subject bone-bundle-bone preparations individually to experimental testing (Butler *et al.*, 1992; Race and Amis, 1994). Though the data obtained in these studies and in the present study serve the same purpose, they are different. The bundle parameters Butler and Race obtained, describe the mechanical behavior of each bundle separately, while the model parameters found in the present study describe the mechanical behavior of the ligament as a whole. Butler *et al.* (1992) tested a few fibers of each bundle of three bundles defined in the ACL to determine their relative material properties. Race and Amis (1994) obtained a stiffness value of $75 \pm 31 \text{ N mm}^{-1}$ for the posterior bundle, or reinforcing bundle, of the PCL which is in agreement with the bundle stiffness of fiber bundle 1 in the present study. The obtained stiffness value of the anterior bundle ($306 \pm 130 \text{ N mm}^{-1}$) is, however, lower than the sum of the stiffnesses of the anterior PCL bundles in the present study. An explanation for this is that in the smaller entities the fiber bundles are loaded more uniformly than in the anterior bundle as a whole, resulting in higher stiffness values. Additional explanations are differences in health, *ante mortem* activity level and gender of the donors. The ages of the donors used in our study are within the range of the donor ages of the study of Race and Amis (1994).

In conclusion, it can be stated that in all cases the line element models are capable of simulating the experimental results, although much better for the cruciate ligaments as compared to the collateral ligaments. ACL models with one or two line elements were not capable of describing the experimental results properly, which supports the idea that a multi-bundle concept of the ligaments is warranted to characterize fully the mechanical behavior of human knee ligaments. Models with three line elements were very sensitive to the geometry of the line element configuration and models with more than 7 line elements suffered from mathematical redundancy. Between 4 and 7 line elements little difference was found. The method described in this paper could also be applied for other ligamentous structures.

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