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KINEMATICS OF THE HUMAN KNEE JOINT

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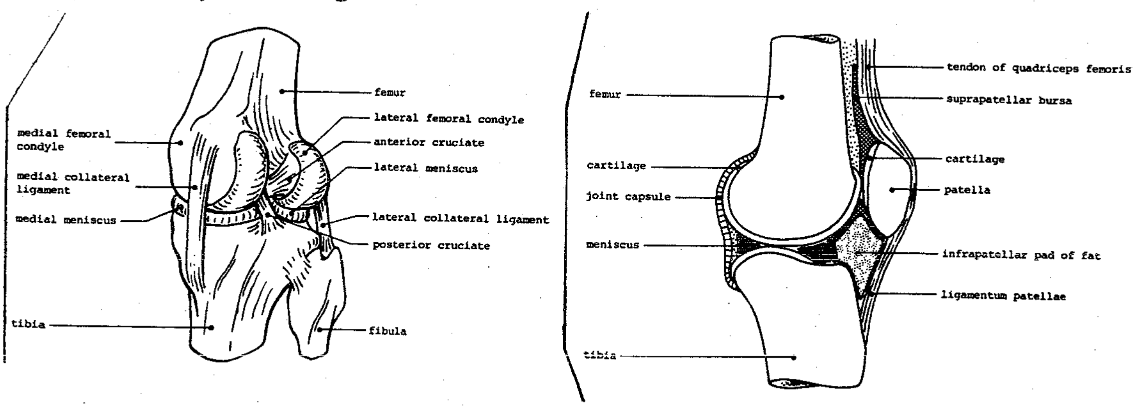
1. INTRODUCTION

The human knee joint is probably one of the most complicated joint structures from a kinematics point of view, and certainly more complex than any technical joint design known. Viewed as a mechanical system it consists of two relatively irregular bearing surfaces, the tibial and femoral condyles, covered with articular cartilage. Interposed between these relatively rigid structures are the compliant menisci. The bones are connected by collageneous fibers organised in a capsule and several ligaments, of which the two cruciate ligaments and the two collateral ligaments are the most important (Figs. 1 and 2). The principal motion of the joint is flexion, although a considerable amount of rotation around the



(a) (b)

Fig. 1 (a) The knee joint viewed from postero-medially, without muscles and joint capsule; (b) a sagittal section; the terminology for some joint structures is indicated (reproduced from Ref. 46).

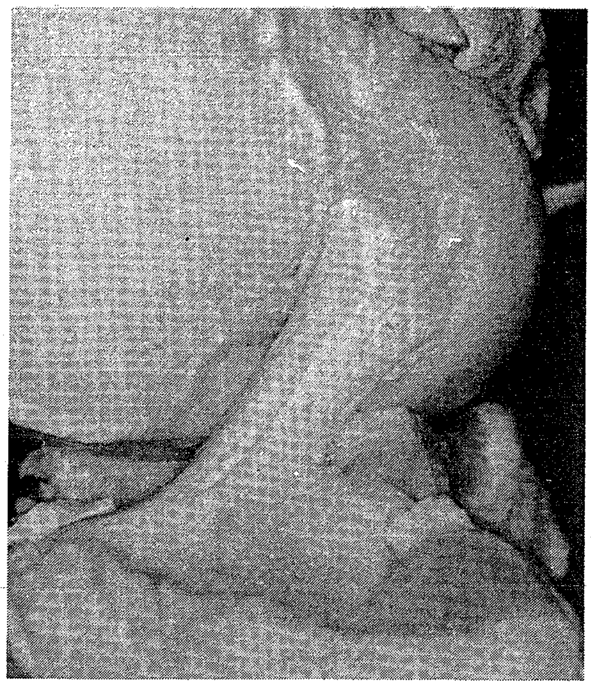


Fig. 2 The anterior cruciate ligament viewed medially; the medial femoral condyle is excised. The three-dimensional fiber structure of this ligament is easily recognized (reproduced from Ref. 3).

longitudinal tibial axis, called exo-endorotation, is also possible when the joint is not in full extension. The most important extensor muscle group of the knee, the quadriceps, is connected to the patella, which slides over a frontal femoral articulating surface. The patella, in turn, is connected to the tibia by the patellar tendon, thereby increasing the lever arm of the quadriceps with respect to the joint (1,2).

Kinematics is the study of motion without taking the cause of the motion into account. However, the actual kinematic behavior of the knee joint, i.e. the motion patterns of the bones relative to each other, varies depending on the muscle loads. Hence, a kinematic analysis of the knee can either be an analysis of joint motion in specific functions (e.g. walking, running, stair climbing, etc.), or an analysis of the motion feasibilities, and the freedom of motion characteristics of the joint. In this brief introduction we will confine ourselves to the latter aspect. Where knee motion analysis during specific functions is concerned, most work has been done in the area of gait of normal and pathological subjects (e.g. 4,5).

Analyzing the motion patterns of the knee joint, and determining how these are influenced by the characteristics of the articular surface geometry, the menisci, and the ligamentous structures is certainly of great interest. First of all, this knowledge is essential for fundamental concepts, i.e. kinematic joint models, on which functional analyses, such as gait analysis, must be based (6). Secondly, understanding the kinematics of the joint in terms of objective, quantitative concepts must be the basis for further dynamic analyses. Finally, but most important of course, is the need for more precise, quantitative data about normal and pathological knee joint behavior in fields such as orthopaedic surgery, rehabilitation, sports, and ergonomics.

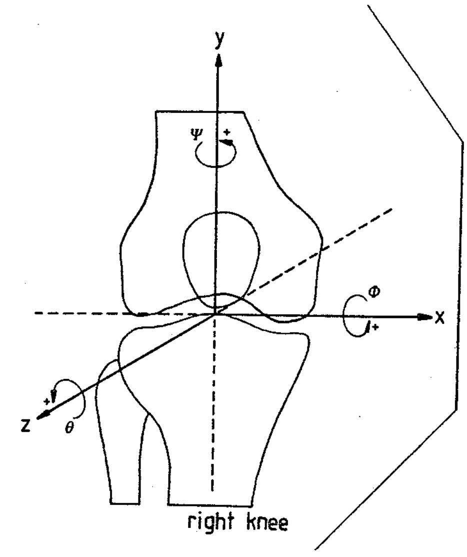
In clinical orthopaedics, for example, the knee is a frequent subject of treatment. Most often occurring knee disorders are arthrosis, and traumatic ligament and meniscus lesions. In methods of diagnosis and treatment of such cases, a sound understanding of knee joint kinematics is important. A treatment of severe arthrosis, for instance, is replacement of the joint by a prosthesis. The kinematic characteristics of the prosthetic device in relation to the remaining joint structures and the normal properties of the intact joint play a major role in a successful procedure (e.g. 7). In ligament trauma, the severity of the lesions must be assessed by objective diagnostic methods, which are partly based on evaluations of the kinematic behavior of the joint (e.g. 8,9). Severe lesions require surgery and ligament repair or replacement, procedures for which understanding of the ligament functions in joint kinematics is essential (e.g. 10). Not surprisingly, therefore, the knee joint motion patterns have been subject to studies for a long time (e.g. 11-15).

2. DEFINITION OF THE PROBLEM

To describe joint motion, the femur and the tibia are fitted with Cartesian coordinate systems, $E_{\underline{X}}$ and $E_{\underline{x}}$, respectively. We assume that the tibia moves with respect to the femur, and we refer to $E_{\underline{X}}$ and $E_{\underline{x}}$ as the "space-fixed" and the "body-fixed" systems, respectively. In the fully extended position of the joint, both systems coincide (Fig. 3). The reference systems are chosen such that translations and rotations more-or-less correspond with the accepted anatomical terminology. The X-axis points from lateral to medial, the Y-axis runs axially, pointing from distal to proximal, the Z-axis from posterior to anterior. The XY-plane is the frontal plane, the XZ-plane the horizontal plane, and the YZ-plane the sagittal plane.

Following the principles of rigid body kinematics (e.g. 16,17) the position of a point in the body-fixed system, $\underline{x} = (x, y, z)^T$, $\underline{X} = (X, Y, Z)^T$ by using

$$\underline{X} = [R]\underline{x} + \underline{d} \tag{1}$$



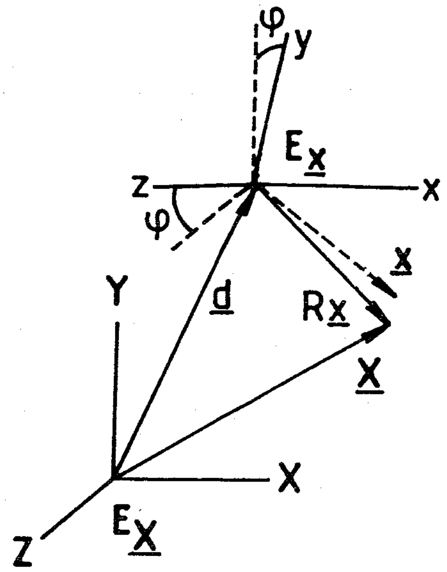


Fig. 3 Cartesian (body-fixed) coordinate system $E_{\underline{x}}$ attached to the tibia, right-handed for a right knee, left-handed for a left knee. In full extension of the knee this coordinate system coincides with the femoral (space-fixed) system $E_{\underline{X}}$. Rotations about the x-axis (ϕ, flexion), y-axis (ψ, exo-endorotation) and z-axis (θ, ab-adduction) are indicated; the x-axis points medially, the z-axis anteriorly.

Fig. 4 An arbitrary finite motion step of $E_{\underline{x}}$ with respect to $E_{\underline{X}}$ is characterized by a transaltion $\underline{d}$ and subsequent rotations of $E_{\underline{x}}$ in the translated position about the x-, y-, and z-axes, respectively. Here, only one rotation (ϕ) is assumed.

where $\underline{d} = (d_X, d_Y, d_Z)^T$ denotes the position vector of the origin of $E_{\underline{x}}$ with respect to $E_{\underline{X}}$, and [R] denotes the orientation matrix of $E_{\underline{x}}$ with respect to $E_{\underline{X}}$. The orientation matrix [R] depends on three independent variables, commonly referred to as Euler angles. If $E_{\underline{X}}$ and $E_{\underline{x}}$ coincide in one position of the joint, then a subsequent position can be described as a translation $\underline{d}$ of the origin of $E_{\underline{x}}$ with respect to $E_{\underline{X}}$, and subsequent rotations of $E_{\underline{x}}$ around the coordinate axes, expressed by [R], which is also referred to as the rotation matrix.

The rotation angles chosen is rather arbitrary, although the choice determines their anatomical significance. Here we use rotations about the x-axis (ϕ), y-axis (ψ) and z-axis (θ) of the body-fixed, tibial reference system (Figs. 3 and 4). The representation of these Euler angles in the rotation matrix is sequence dependent (16,17). Generally, as will be discussed later, these

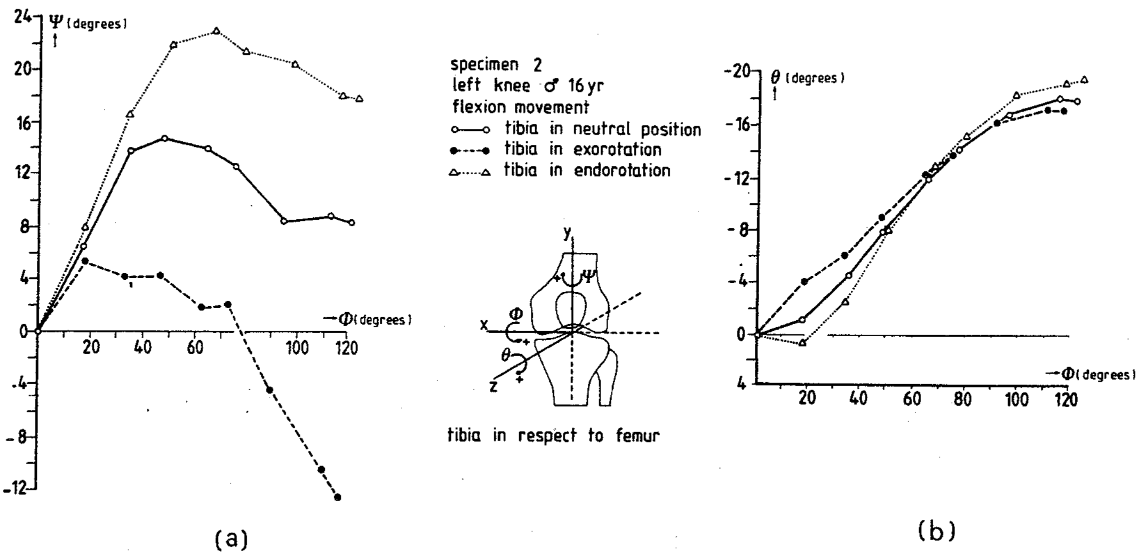


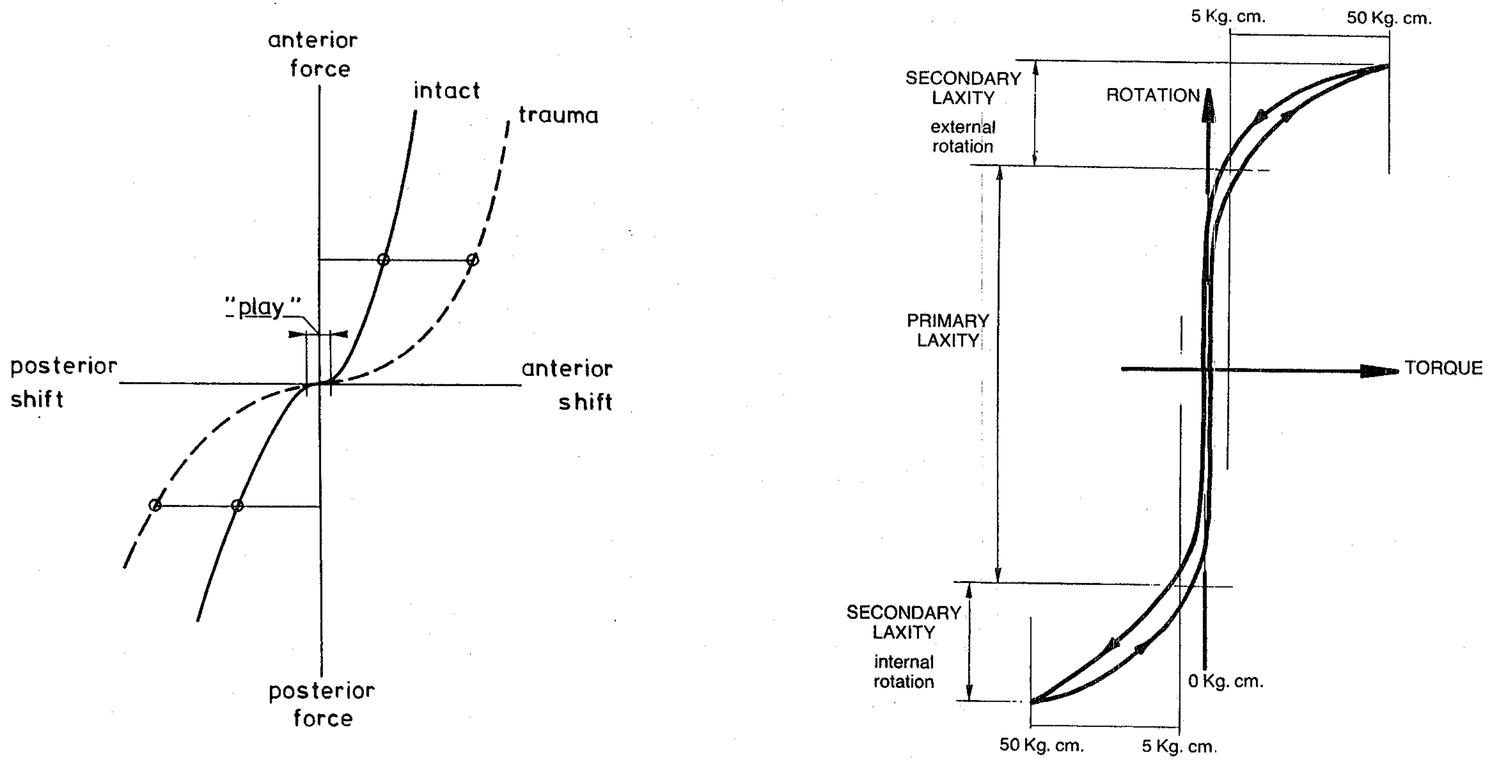
Fig. 5 Three step-by-step motion pathways in flexion of a knee joint specimen, measured with the roentgenstereophotogrammetric system: (a) exo-endorotation as function of flexion, and (b) ab-adduction as function of flexion (reproduced from Ref. 3).

rotations represent flexion-extension (ϕ, flexion positive), exo-endorotation (ψ, endorotation positive), and ab-adduction (θ, adduction positive).

The measurement, modeling and interpretation of joint motion in terms of the above kinematic parameters are complicated by number of problems and practical difficulties, which are inherent to the irregular biological nature of the structure or to the character of the descriptive parameters.

Firstly, although the knee joint has two major degrees of freedom (flexion-extension and exo-endorotation), kinematic coupling with other rotations and translations occurs to a significant degree. Figure 5.a shows the exo-endorotation (ψ) of a knee joint moved from full extension to 120 degrees of flexion (ϕ) along three different "pathways"; a graph which illustrates the two primary degrees of freedom. Figure 5.b shows, for the same motions, the ab-adduction rotation (θ) as a function of flexion. Apparently, although there is no significant freedom of rotation about the ab-adduction axis, there is rotation, coupled with the flexion and exo-endorotation motions. This coupling effect occurs in other rotations and translations as well.

A second problem is associated with the compliance of the joint restraints. The ligaments which, together with the articulating surfaces are the primary restraints of joint motion, have highly non-linear load-displacement characteristics. This has two



(a) (b)

Fig. 6 Due to the non-linear compliance of the ligaments, force-displacement and moment-rotation relations have non-linear characteristics. (a) progressively increasing resistance is found against anterior and posterior forces, with only small resistance (play) around the neutral position. The amount of play may be very small or even non-existent in some knees, but increases considerably after ligament lesions; (b) the exo-endorotation displays freedom of motion in a relatively large range, limited by progressively increasing resistance against rotational moments (torques); due to the non-linear character of the resisting elements, the limits of motion freedom are not well-defined.

important consequences. Firstly, very small forces and moments will generate a certain amount of "play", small rotations about the ab-adduction axis and translations in the antero-posterior (d_z), the medio-laterial (d_x), and the longitudinal (d_y) directions (e.g. 18). This "play" is most distinguishable in the A-P-direction, a shift which is often referred to as the "A-P-drawer" (Fig. 6.a). This lack of rigid restraints implies that the joint has actually six degrees of freedom, and that its designation as a two degree-of-freedom mechanism by considering the primary motions only, is rather an arbitrary one. A second consequence of the ligament laxity under relatively small loading conditions concerns the limits of exo-endorotation (Fig. 6.b). As the resistance to increased rotation builds up progressively, these limits are rather arbitrary, and depend on the applied torque. Hence, not only the actual motion of the joint, but also its degrees of freedom depend, to a certain extent, on the external loads. Quite often the terms "primary" and "secondary" laxity are used, as depicted in Fig. 6.b, with arbitrary limits, to describe joint motion.

A third, but major difficulty is that virtually no method exists to relate the coordinate systems to the local anatomy of the

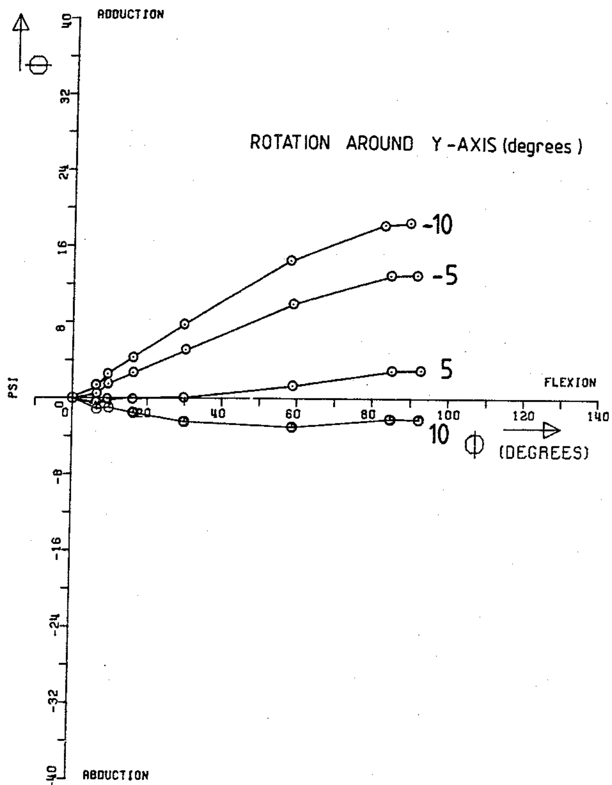


Fig. 7 Ab-adduction as a function of flexion for a specific motion pathway of one joint specimen. Different curves are obtained when the rotation angles are expressed in different body-fixed reference systems where the axes system is rotated -5, -10, +5, and +10 degrees about the y-axis (compare with Fig. 3).

bones in a precise and consistent manner. This particularly complicates a detailed comparison between motion characteristics of different joints. An illustration of the extent of this problem is shown in Fig. 7. As in Fig. 5.b, this graph presents the ab-adduction rotation (θ) as a function of flexion (ϕ) for a knee joint specimen. All four curves represent the same motion pathway, but for different body-fixed reference systems, where the axes system is rotated +5, -5, +10, and -10 degrees about the y-axis (compare with Fig. 3). Evidently, the orientation of the coordinate system with respect to the joint anatomy can have a significant influence on the motion curves obtained.

A fourth difficulty lies in the interpretation of kinematic motion parameters in anatomical terms. While exo-endorotation of the tibia is rotation about the body-fixed y-axis, flexion-extension is rotation about the space-fixed X-axis. The latter coincides with the x-axis in full extension, but not in flexion. Hence, the flexion angle ϕ represents flexion in the anatomical sense only if the flexion position of the tibia with respect to the femur refers to the fully extended position. The ab-adduction rotation is, in an anatomical sense, not well-defined; the use of the z-axis or the Z-axis would be defendable. The arbitrary character of the "anatomical" ab-adduction axis has resulted in many controversies in the literature, in particular where knee motion patterns in higher flexion angles were concerned (3).

A linkage system to be used as a reference which more closely corresponds with the anatomical terminology, a so-called gyroscopic system, has been suggested (19,20). This system is defined on the X-axis, the y-axis, and a third floating axis perpendicular to the first two. This linkage system, which is neither body-fixed nor space-fixed, is not orthogonal. However, in addition to its anatomical nature it also has the advantage that the rotation matrix describing the motion is sequence independent.

The kinematic parameters defined in any one system can be easily transformed to another one. It has been suggested (21) to relate reference systems to well-defined bony landmarks, so that all coordinate systems can then be related to an anatomical one. However, apart from the fact that bony landmarks are not easily identified in a precise reproducible fashion, there remains the problem of defining "anatomical" coordinate systems for joints in an unambiguous way.

3. EXPERIMENTAL METHODS AND MODELING TOOLS

A large number of methods have been proposed in the literature to measure knee joint motion both in vivo and in vitro (e.g. 22,23). Many of these techniques rely on the assumption of planar (flexion-extension) motion (e.g. 9). In general, experimental techniques can be divided into two, analog and digital. In the first case an instrumented linkage mechanism may be fixed to the bones, which moves in parallel with the joint; the motion of the linkage system is then monitored and transformed to the joint coordinate system. Well-known examples of such mechanisms are simple planar goniometers measuring flexion-extension only, the triaxial goniometer (24) monitoring the three joint rotations, and six degrees-of-freedom linkage mechanisms measuring the complete joint motion (25,26).

Use of the digital method implies measuring relative joint positions after finite motion steps. If the positions of three non-collinear landmarks i (i = 1, 2, 3) are measured before (x) and after ($\underline{X}_i$) a finite motion, then

$$\underline{X}_i = [R]\underline{x}_i + \underline{d} \qquad (i = 1,\ 2,\ 3) \tag{2}$$

gives a set of equations from which the three Euler angles and the translation components describing the change of position can be evaluated. The finite motion methods are based on this principle. Lacking well defined bony landmarks, object points are usually attached as landmarks to the limbs or the bones, although anatomical points have also been used. To register the coordinates of object points, different methods such as cinematography, sonic digitizing, photogrammetry, and optoelectronics can be used (22,23).

An experimental system for studies of the skeleton based on roentgenstereophotogrammetry was developed by Selvik (17), and this technique is used to measure and describe joint motion in vitro at the biomechanics laboratory in Nijmegen (3,27,28,29,30). Object points are small tantalum pellets, 0.5 to 1.0 mm in diameter, inserted in the bones. The object is imaged in successive positions on two roentgenograms, which are measured on a 2-D coordinate digitizer. The roentgenograms include the images of a calibration cage with markers of known 3-D positions. Using principles of analytical stereophotogrammetry (17), the 3-D locations of the object points are reconstructed by a computer program, based on the 2-D evaluations of the roentgenograms.

An essential feature of the subsequent calculations of the kinematic parameters describing the relative change in position is the redundancy of the landmark system. If a bone contains n tantalum pellets (n $\geqslant$ 3), the kinematic parameters [R] and $\underline{d}$ are evaluated by minimizing

$$\sum_{i=1}^{n} (\underline{X}_i - [R]\underline{x}_i - \underline{d})^2 \tag{3}$$

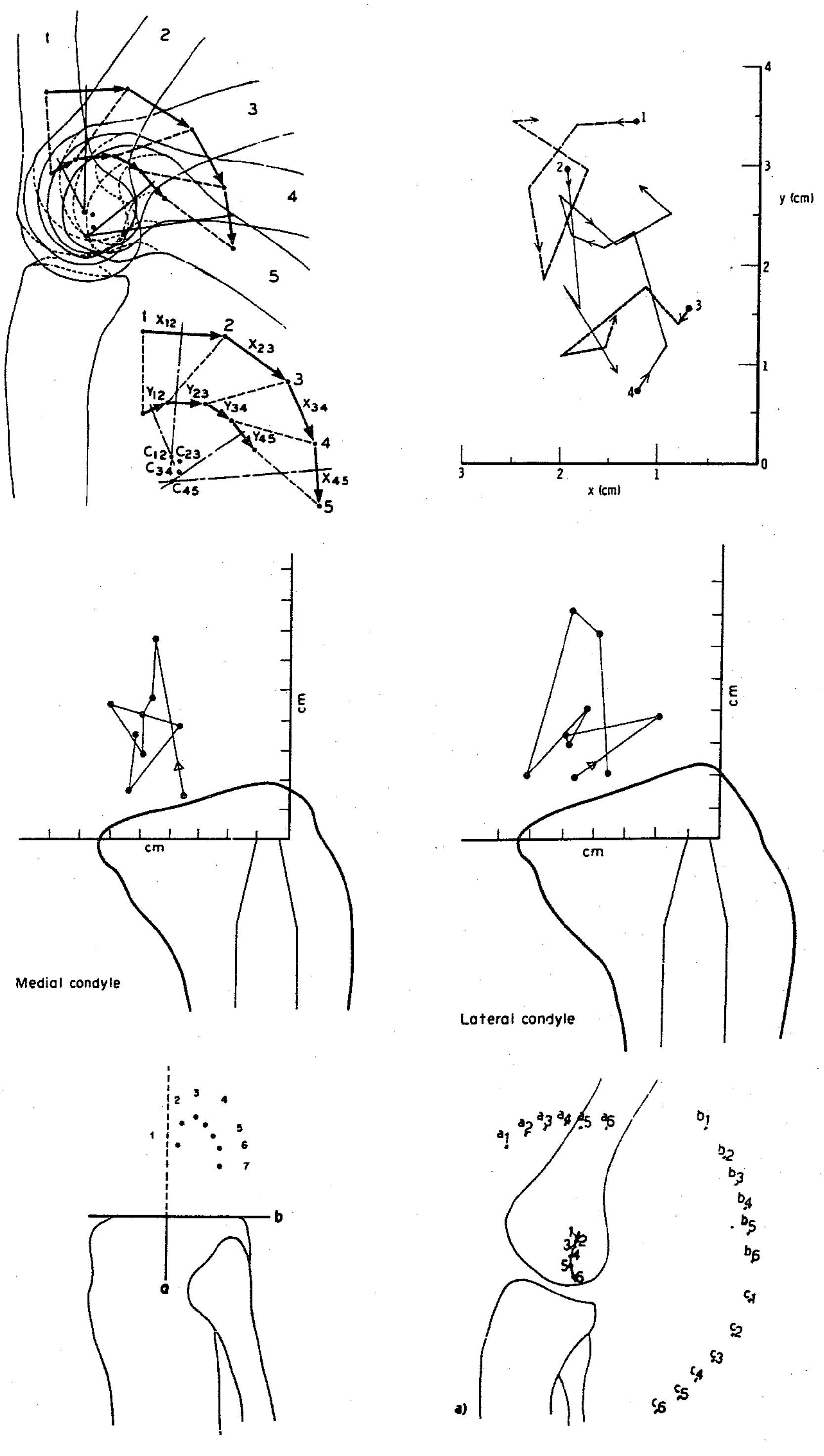


Fig. 8 "Instant center of rotation" pathways as evaluated in planar and quasi-planar knee joint kinematic studies, usually calculated from sequential lateral roentgeongrams (adapted from Refs. 9 (top left), 36 (top right), 38 (middle), 37 (bottom left), and 34 (bottom right); reproduced from Ref. 3).

which is solved using a non-linear least square method (17). The redundancy of the markers ensures a higher accuracy in the kinematic parameters, depending on the number of markers implanted.

A well-known method to describe changes in relative positions between rigid bodies, as an alternative to Euler angles and translation vectors, is the use of the so-called helical axis, or screw axis (16,17). From Eqn. 2 it follows that when a point q undergoes a translation

$$\underline{d}_q = \underline{X}_q - \underline{x}_q = [R]\underline{x}_q - \underline{x}_q + \underline{d} \qquad (4)$$

This implies that all points on the line $\underline{x}_q = \underline{x}_s + \gamma\underline{n}$ $(-\infty<\gamma<\infty)$ undergo the same translation if $[R]\underline{n} = \underline{n}$, which is satisfied for an eigenvector of [R]. These points lie on the helical axis, identified by the unit direction vector $\underline{n}$ and the position vector $\underline{x}_s$. The motion of the two bones relative to each other can now be characterized by a rotation about, and a translation t along this axis. The helical axis parameters $\underline{x}_s$, $\underline{n}$, α and t can be evaluated from the rotation matrix [R] and translation vector $\underline{d}$ by using several different methods (e.g. 17,31).

For planar motions the helical axis is perpendicular to the plane of motion and hence can be characterized by a point, which is called the "instant center of rotation", and represents a point about which a finite rotation takes place (Fig. 8).

An example of subsequent helical axes in knee flexion is shown in Fig. 9.a. Each axis describes a flexion step of approximately 15 degrees. An attractive aspect of the helical axis representation is its illustrative quality, giving a more direct impression of joint motion as compared to the abstract Euler angles and translation vector representation. Another advantage is the invariance of the helical axis position for the chosen coordinate systems, although the axes must eventually be related to the joint anatomy for visual interpretation.

The helical axis method is a versatile tool to represent a specific joint motion, but not so much to describe the freedom of motion in multi degrees-of-freedom joints. Fig. 9.b, for example, shows the successive helical axes for the same knee of Fig. 9.a, but this time flexed along another pathway (exorotation; compare with Fig. 5). The apparent discrepancy between the two bundles of axes cannot readily be interpreted physically.

The helical axis position is also strongly influenced by other slight differences in motion characteristics, as for instance "play" and, for the same reason, by stochastic measurement errors in the motion assessment (i.e. the calculated 3-D positions of object landmarks).

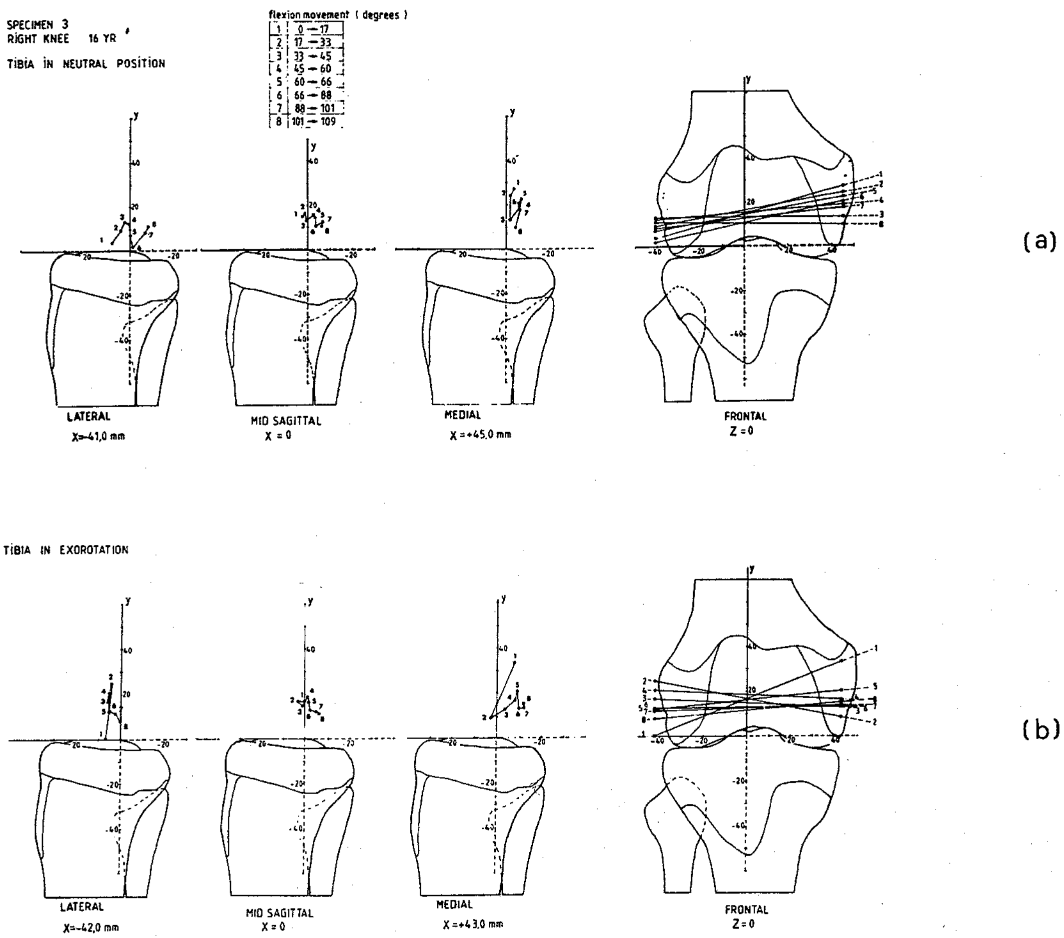


Fig. 9 Helical axes for sequential flexion steps of a knee joint specimen shown in a frontal projection (right), and the piercing points of these areas with the laterial, mid-sagittal and medial planes (left); (a) for a neutral pathway (compare with Fig. 5), and (b) for a pathway more towards exorotation (reproduced from Ref. 3).

It was shown in a theoretical analysis (32), assuming an isotropic object landmark distribution, that the error propagation in the helical axis position and direction (represented by $\underline{x}_s$ and $\underline{n}$) strongly depends on the distance between the helical axis and the center of gravity of the landmark distribution. Therefore, the object points should be chosen or placed such as to surround the anticipated axis. When both the distance of helical axis to landmark center of gravity, and the helical rotation are small ($|\alpha| \ll 1$ rad), it can be shown (32) that the propagation of a landmark position standard error per coordinate in the helical axis parameters follows from

$$\sigma_a = \frac{\sigma}{\rho}\sqrt{2/n} \quad , \quad \sigma_t = \sigma\sqrt{2/n} \quad , \quad \sigma_{\underline{n}} = 2\sigma/\rho a\sqrt{n} \quad ,$$
$$\sigma_{\underline{x}_s} = 2\sigma/a\sqrt{n} \tag{5}$$

where ρ is the effective landmark distribution radius, n the number of landmarks, σ_a and σ_t the standard errors in the helical rotation and the helical shift t, respectively, σ_n the direction standard error of $\underline{n}$, and σ_{x_s} the standard error in $\underline{x}_s$. These theoretical results have been confirmed in wrist joint kinematic measurements (30), and are in agreement with error analyses of the instant center of rotation for planar motions (33,34). Evidently, besides placing the landmarks around the anticipated helical axis, the width of the landmark distribution (2ρ) and the number of landmarks (n) must be as large as possible to ensure accurate results. Most importantly, however, the rotation step must not be too small.

The last aspect of error propagation leads to a controversy in the helical axis concept. The helical axis represents a model, describing the change in position of a rigid body as a pure rotation around, and a translation along the axis. The smaller the motion step, the better this model will describe the real motion pattern, but the less accurately the axis can be determined. In the knee joint motion evaluation of Fig. 9, six tantalum landmarks were applied in a distribution with an effective radius of about 10 mm. Given, for example, the standard error of the 3-D landmark position evaluation of approximately 50 μm, the standard direction error of $\underline{n}$ will vary according to the above formula, and will take values between 1.3 and 13 degrees corresponding to a helical rotation variation from 10 to 1 degrees. Hence, improvement of the helical axis model towards representation of continuous motions is achieved only at the cost of a considerable loss in accuracy.

Returning now to the results shown in Fig. 9, which are typical for several joint specimens (3), it is evident that the knee motion is not a planar one. Comparing Fig. 9.a with Fig. 9.b also indicates that the motion is not unique. However, for a rough, first order approximation of knee motion, the assumption of a fixed, "best fitting" axis, as has been proposed earlier (35), would not be too unrealistic. Although this "optimal" axis should be slightly inclined both in the frontal and lateral planes (Fig. 9.a), it would probably be possible to assess its position by rough approximation from lateral X-rays, assuming planar motion and using the instant center of rotation concept. However, in view of the rather wildly scattered intersection points of the successive axes with the sagittal planes (Fig. 9), it seems rather useless to designate any realistic value to the patterns of such a planar instant center of rotation, specifically for diagnostic purposes. The large differences in instant-center-of-rotation patterns reported in the literature support this conclusion (Fig. 8).

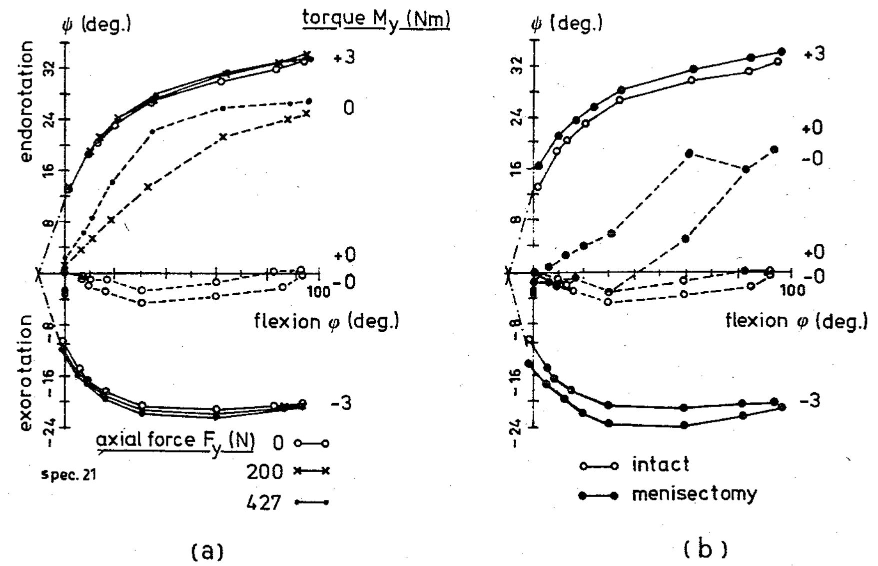


Fig. 10 Exo-endorotation as a function of flexion, as measured for one knee specimen; (a) unloaded and loaded with several axial forces and exo-endorotation moments; (b) unloaded and loaded with an exo- and an endorotation moment, before and after menisectomy.

4. THE EFFECTS OF KNEE STRUCTURAL ELEMENTS AND LOADING

The kinematic behavior of the knee in terms of freedom of motion is determined by the geometrical and material properties of the joint structures. To evaluate the influence of each structure on the knee kinematics is important, particularly from a clinical point of view. The actual motion of the knee joint in the performance of a certain task depends on the freedom of motion, as well as on the external loads. In the remainder of this chapter some of these effects will be discussed.

Figure 10 presents a number of curves, similar to those shown in Fig. 5.a. Each curve represents an exo-endorotation movement, ψ, as function of flexion, ϕ , determined for one knee joint specimen using the roentgenstereophotogrammetric method discussed previously. In Fig. 10.a the effects of two kinds of loads are shown, a torsional moment around the y-axis, M_y, and an axial compressive force, F_y. When the joint is unloaded ($F_y = 0$, $M_y = \pm 0$), hardly any exo- or endorotation occurs, but the precise motion path is rather uncertain ($M_y = \pm 0$ refers to slight rotational shifts applied at the first motion step). We will refer to these curves ($M_y = \pm 0$) as the "neutral" pathways. If endo- or exorotation moments are applied the joint follows corresponding endo- and exorotation pathways. The range of exo-endorotation motion arbitrarily measured here for applied torsional moments of -3 Nm and 3 Nm respectively, increases with flexion. It appears that this freedom of motion is not influenced by an axial force (Fig. 10.a). The neutral curve, however, displays a considerable shift towards endorotation for increasing axial loads, as also shown in the figure. Besides these shifts, there is a definite "firmness" of these pathways in axially loaded cases: the motion is less arbitrary and is not affected by small torsional moments ($M_y = \pm 0$).

These shifts in the neutral curve due to an axial load may be caused by cartilage deformation and/or as a result of articular surface geometry.

The influence of the menisci on the exo-endorotation freedom of motion is illustrated in Fig. 10.b. The freedom region increases approximately by 5 to 10 per cent, under torsional moments of $M_y = \pm 3$ Nm, after removal of the menisci. The neutral curves are less "firm", and more strongly affected by small rotational shifts. Under axial loads, however, the original curves for the intact joint are almost reproduced. Hence, it appears that the influence of menisci on the kinematic behavior of the knee joint is not very pronounced. It must be remarked, however, that the meniscus has an important function in static and dynamic load transmission.

The effects of the axial and torsional loads, and those of meniscotomy on the kinematic characteristics discussed here are reproducible in other specimens as well. They are also in agreement with observations reported elsewhere (18, 40-43). However, due to the arbitrary nature of the imposed load-dependent exo-endorotation limits the terminology of the interpretations varies to some extent (e.g. 43).

A major role in knee motion characteristics has traditionally been attributed to the cruciate ligaments (e.g. 3,11,14,15,44,46). A well known concept for planar knee motion assumes the cruciates to act as rigid bars, kept taut by distraction forces generated due to articular contact (e.g. 15,44). The joint thus behaves as a four-bar-linkage mechanism, with an instant rotation center coinciding with the point of intersection of the ligaments. Essential in this concept is the close interaction between articular surface geometry and ligaments.

Evidently, however, the ligaments are relatively compliant 3-D structures rather than rigid line elements (Fig. 2), and the question is whether such a simple and therefore attractive model can be maintained in spatial motion concepts.

When the rotation matrix [R] and the translation vector $\underline{d}$ have been calculated for each relative knee joint attitude, the location of a femoral insertion point of a ligament, $\underline{X}_{fj}$, can be related to a tibial insertion point $\underline{x}_{tj}$, if these vectors are known in their

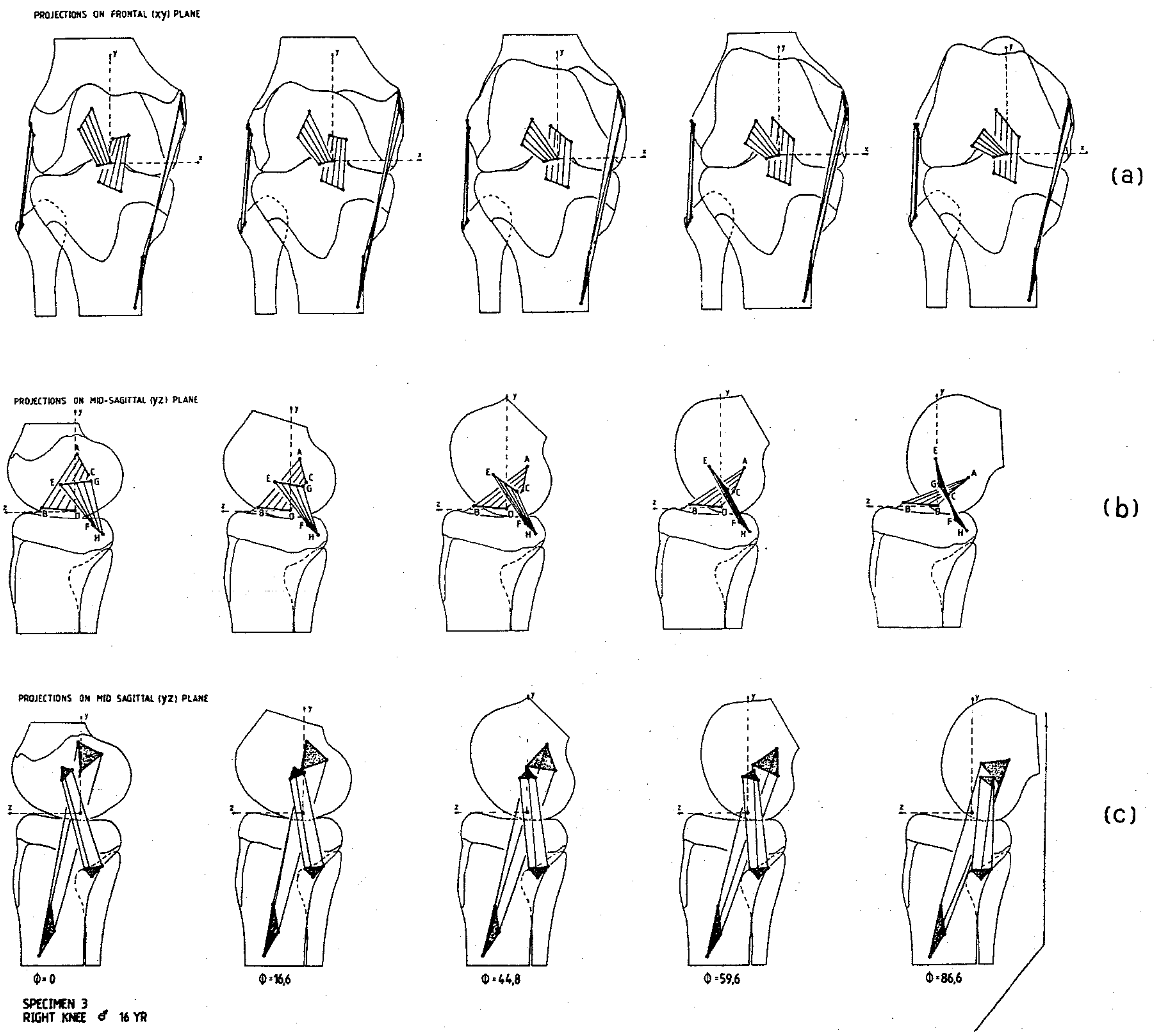


Fig. 11 Frontal (a) and sagittal (b,c) views of a knee joint specimen in successive flexion steps along a neutral pathway as shown in Fig. 5. The cruciate and collateral ligament insertion regions are marked with two and three pellets, respectively. This way the geometrical changes of these ligaments can be interpreted visually as represented by the line-elements (adapted from Ref. 3).

respective coordinate systems. In several motion evaluation experiments the insertion regions of the cruciate and collateral ligaments were marked with two and three tantalum pellets respectively, and the 3-D locations of these pellets were measured (3,28). As a result, a geometric configuration model of the ligaments and their deformation during joint motion are determined, as shown in Fig. 11. The flexion positions represented here correspond to a neutral, unloaded pathway, depicted in Fig. 5.

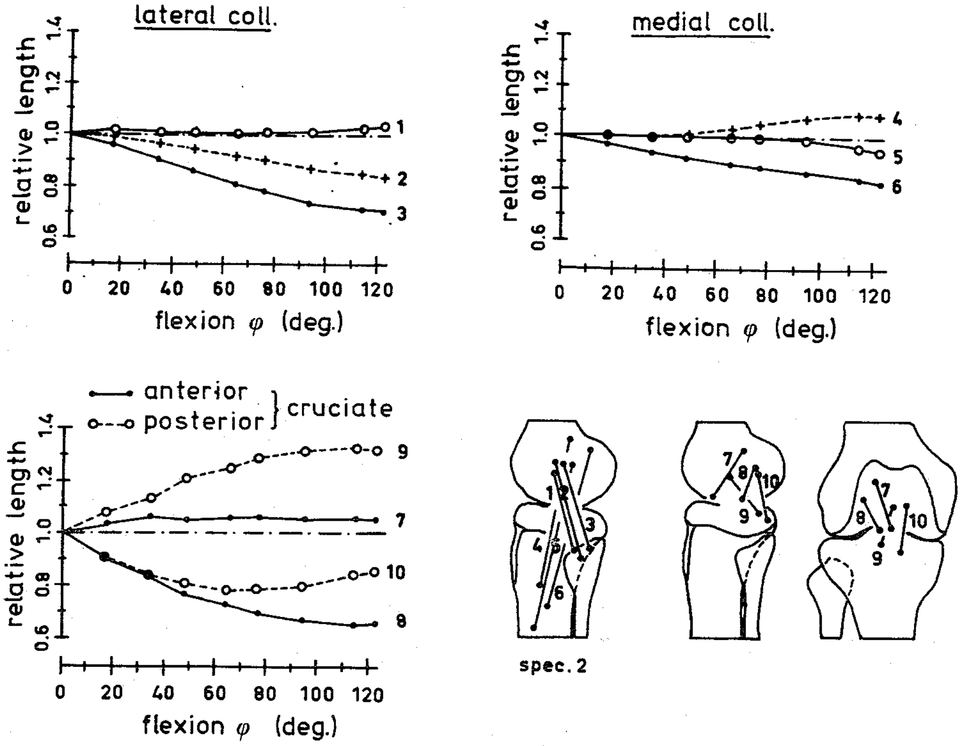


Fig. 12 Changes of the line-element lengths representing ligament bands of a knee joint specimen in flexion (neutral pathway). Lengths are expressed as a percentage of ligament lengths in full extension (adapted from Ref. 3).

The action of the cruciate ligaments as four-bar-linkage elements can be appreciated from Fig. 11.b. They appear to "pull" the femur posteriorly, controlling the sliding/rolling motion between the two articular surfaces. When combining these projections in the sagittal plane with those of the piercing points representing the helical axes of Fig. 9, it follows that the axes intersect the crossing regions of the ligaments in all positions (3).

The cruciates however, are relatively large in size, and their fibers follow different spatial courses in both the sagittal and frontal planes. It is therefore obvious that the various fibers play different roles at different knee joint positions, as also was demonstrated by anatomical observations (e.g. 3,14,46).

It is interesting to note that the collateral ligaments are also crossed in the sagittal plane (Fig. 11.c). However, unlike the cruciates, they uncross during flexion.

Although these pictures give a good impression of the geometrical complexity of the ligaments, their true contribution to knee kinematics depends on the tension developed within their fibers, which in turn relates to the length changes between the insertions. This latter variable can be assessed for any chosen line element defined by the insertion markers, from

$$L_j = |\underline{X}_{fj} - \underline{X}_{tj}| = |\underline{X}_{fj} - [R]\underline{x}_{tj} - \underline{d}| \tag{6}$$

where L_j denotes the length of a line element j between insertion points $\underline{X}_{fj}$ and $\underline{x}_{tj}$ in a certain joint position determined by [R] and $\underline{d}$. Defining L_{jo} as the length of j in full extension, then $e_j = L_j/L_{jo}$ gives the relative length in the element j.

The relative length changes of ten ligament elements as a function of the flexion angle ϕ are shown in Fig. 12. The figure shows that there is a significant influence of flexion motion on the length patterns. Increases and decreases of lengths up to 40% occur. With the exception of the anterior parts, the collateral ligaments seem to become untaut in flexion. The anterior parts of the cruciates apparently increase in length (elements 7 and 9), whereas the posterior parts decrease (elements 8 and 10).

As suggested by the curves of Fig. 12, the length changes of the elements are extremely sensitive to the location of the chosen insertion points. Nevertheless, the patterns shown for one joint specimen are qualitatively reproducible in other specimens, particularly for the collateral ligaments (3,28). In addition, they show only small changes when other pathways of flexion, discussed earlier, are followed; again these changes are mildest in the collateral ligaments (28).

Although the length patterns shown in Fig. 12 illustrate the complexity of the ligament influences on the kinematic behavior of the joint, it is difficult to interpret them in terms of restraints against freedom of motion. Firstly, as illustrated in Fig. 2 not all the collagen fibers in the ligaments follow a parallel course from femoral to tibial insertion (e.g. 3,46). Therefore, when a line element increases or decreases in length, the true fibers may not follow the same pattern. Secondly, the relative length changes are related to the initial length, and since it is unknown whether a ligament is taut or relaxed in that state, the relative length changes do not represent element strain. It is probably, for example, that a ligament fiber represented by line element 9 (anterior part of the posterior cruciate) is relaxed in full extension, since a strain of 40% would generate a force close to its tensile strength. Finally, effectiveness of a ligament band in restraining a certain freedom of motion depends on its 3-dimensional configuration. It is interesting to note, for example, that element 9, which is probably subjected to the highest force, has in flexion a course almost perpendicular to the joint surfaces (Figs. 11.a and b, line element E-F). Therefore, if indeed a high force is generated in this band of the posterior cruciate, then it would act only to compress the joint surfaces, without giving much resistance to motion in other directions.

5. KNEE JOINT MODELS

It should be evident from the above discussion that the human knee joint is a rather complex mechanical system. Kinematic analyses of its motion characteristics have increased the knowledge about its capabilities and performance, as have anatomical observations, functional gait analyses and experimental evaluations of properties to isolated knee joint structures. However, a complete understanding of the joint function, and in particular, the quantitative effects of the knee structural elements can only evolve if the available knowledge is combined in a mathematical concept, a model. We have already discussed a simple kinematic model, the four-bar-linkage mechanism (15,44), which describes the flexion-extension motion as influenced by the cruciate ligaments.

Although a model could be a simplified representation of reality, in order to be representative of the complex reality, it should include and account for the essential features of the real system. In that respect it seems obvious that a realistic knee joint model should be three dimensional and represent the compliance of the knee structures. That is it should include the external loads as well as the stiffness characteristics of the knee tissues. Such a model combines kinetic and kinematic effects. Two models of this kind have been proposed in the literature (47,48).

The model of Wismans (47) describes the articulating bones as three dimensional rigid surfaces, in contact at two points (medial and lateral condyles), and the ligamentous connections as line elements with non-linear elastic properties. The number of line elements is arbitrary in principle. In the model, the flexion angle is prescribed, as are the external forces and moments, with the exception of the flexion-extension moment. This leaves five degrees of freedom and one loading variable to be determined. Further unknowns are the locations of the contact points on the medial and lateral condyles (2 coordinates per condyle, 8 in total), the magnitudes of the two contact forces, and the forces in the ligament line elements. The latter are described by a priori known data about force-displacement characteristics of the ligaments, which leaves 16 unknown variables. These are evaluated by a system of 16 equations resulting from requirements of equilibrium (6 equations) and contact conditions (10 equations).

The menisci are neglected in this model, as is the compliance of the articular cartilage-bone complex. In addition, the validity of a multiple line element approximation of the ligaments is as yet unknown. The question is not, of course, if these assumptions are very accurate, but whether such an assembly of simplified descriptions can produce valid predictions of knee behavior. It is obvious that such a model must be validated extensively and only then it can be applied to circumstances for which it was developed. The menisci, for example, have shown to exert only minor influences on the normal knee joint kinematics in the experiment discussed in the previous section. However, if one or more knee structures, as ligaments, were deficient or ruptured, it is quite possible that the relative importance of the menisci could increase. A situation which could not be accounted for in this model.

Another problem of sophisticated and complex mathematical models of this kind is that they depend heavily on a priori known data on geometrical and material properties, e.g. quantitative mathematical descriptions of joint surfaces and ligament geometrical and elastic properties. The sensitivity of the model results on these properties must be in balance with the accuracy by which this data can be experimentally determined; the higher the sensitivity, the higher the requirements for accuracy. This again, emphasizes the need for continuing experimental efforts of increasing sophistication.

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