The envelope of passive knee joint motion

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THE ENVELOPE OF PASSIVE KNEE JOINT MOTION*

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Abstract—The purpose of this study is to create an accurate experimental database for the passive (*in vitro*) freedom-of-motion characteristics of the human knee joint on a subject to subject basis, suitable for the verification and enhancement of mathematical knee-joint models. Knee-joint specimens in a six degree-of-freedom motion rig are moved through flexion under several combinations of external loads, including tibial torques, axial forces and AP-forces. Euler rotation angles and translation vectors, describing the relative, spatial motions of the joint are measured using an accurate Roentgen Stereo Photogrammetric system. Conceptually the joint is considered as a two degrees-of-freedom of motion mechanism (flexion-tibial rotation), whereby the limits of internal and external tibial rotation are defined at torques of ± 3 Nm. The motion pathways along these limits are defined as the envelopes of passive knee joint motion. It is found that these envelope pathways are consistent and hardly influenced by additional axial forces up to 300 N and AP-forces of 30 N. Within the envelope of motion, however, the motion patterns are highly susceptible to small changes in the external load configuration. It is shown that the external tibial rotation during extension ('screw-home mechanism') is not an obligatory effect of the passive joint characteristics, but a direct result of the external loads.

Anatomical differences notwithstanding, the inter-individual discrepancies in the motion patterns of the four specimens tested, showed to be relatively small in a qualitative sense. Quantitative differences can be explained by small differences in the alignment of the coordinate systems relative to the joint anatomy and by differences in rotatory laxity.

INTRODUCTION

The actual motion patterns of the human knee joint depend on a combination of its passive motion characteristics and the external loads. The former are determined by the bony and ligamentous constraints and the latter depend on muscle activity and external forces during specific knee joint functions. A number of authors have studied knee joint motions during specific tasks, such as walking, running and stair climbing (e.g. Chao *et al.*, 1983). It must be recognized that these motions are subject to the high variabilities inherent to the system, which are provided by the combination of freedom-of-motion and redundancy in the muscle activation characteristics. Inter- and intra-variability in motion patterns can be expected for comparable tasks. In other words, subjects can change their walking patterns at will or walk differently by habit, and evidently different tasks produce different motions (Lafortune, 1984; Murphy *et al.*, 1984). The basis for the understanding of the kinematics of the knee joint lies in the description of its passive motion characteristics.

The purpose of the present investigation is to measure and describe the passive motion characteristics of the human knee joint and to quantify the role individual joint structures play, on a subject-to-

subject basis. The first task requires an accurate measurement system for three-dimensional joint kinematics and a six degrees-of-freedom motion and loading rig to apply the moments and forces necessary to move the joint along and within its constraints in a natural way. These requirements are realized by the combined application of a Roentgen Stereo Photogrammetric (RSP) system (Selvik, 1974; de Lange *et al.*, 1985) and a sophisticated motion and loading apparatus (Hamer, 1982). The purpose of this paper is to report on the technique, the precision and the results of the motion measurements. The passive freedom-of-motion characteristics obtained are considered to serve as an accurate and complete database for verifying, developing and manipulating analytical knee joint models. In view of the complexity of the joint structure and its mechanical inter-relations, these models are indispensable for the assessment of the role of the joint structures in knee motion characteristics, in normal and pathological conditions (Huiskes *et al.*, 1985a).

A number of authors have studied passive knee joint motion, laxity and stability. Investigative tools for control of motions and external loads have included direct manipulation (Hallen and Lindahl, 1966; van Dijk *et al.*, 1979), instrumented handlebars (Markolf *et al.*, 1976; Nielsen *et al.*, 1984a, b), and specially instrumented fixtures in MTS or Instron testing machines (Wang and Walker, 1974; Hsieh and Walker, 1976; Piziali *et al.*, 1980; Markolf *et al.*, 1981; Fukubayashi *et al.*, 1982). Triaxial goniometers (Chao *et al.*, 1983; Nielsen *et al.*, 1984a, b), six degrees of freedom instrumented spatial linkages (Kinzel *et al.*, 1972; Suntay *et al.*, 1983), photogrammetric methods (Walker *et al.*, 1972; Blacharski *et al.*, 1975) and

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Roentgen photogrammetric systems (van Dijk *et al.*, 1979; Kurosawa *et al.*, 1985) have been used for motion measurements. These studies have contributed greatly to the understanding of knee mechanics in general. However, several controversies still exist with respect to specific aspects of the kinematic behavior of the knee. These controversies were sometimes promoted by confusion over anatomical reference systems in which motions are evaluated and presented, differences in experimental set-up in which the susceptibility of joint behavior to external forces and motion restraints were not always recognized, neglectance of anatomical variations among specimens and finally, confusions over differences in motion feasibilities and actual *in vivo* motions in specific tasks. More importantly however, most experimental efforts have been directed at specific aspects of knee joint behavior over ranges of experimental specimens, whereas very little, if anything, has been done to extensively document the properties and behavior of individual joints. Such a database is a prerequisite for the enhancement of knee joint motion concepts and mathematical models.

MATERIALS AND METHODS

To describe joint motion, Cartesian coordinate systems, E_f and E_t , respectively are applied to the femur and the tibia. The tibia is considered to move relative to the femur. Hence, E_f and E_t are referred to as space-fixed and body-fixed reference systems respectively. The position of a tibial point $\mathbf{p}$ in the body-fixed system E_t is expressed in the space-fixed system E_f as

$$\mathbf{P} = \mathbf{D} + \mathbf{R}\mathbf{p}, \quad (1)$$

where $\mathbf{D}$ denotes the position of the origin of E_t with respect to E_f . The rotation matrix $\mathbf{R}$ depends on three independent variables, usually referred to as Euler angles (Goldstein, 1959). In this case subsequent rotations around the three axes of the reference system E_f were chosen. The reference systems are chosen such that the rotations correspond by approximation with the accepted anatomical terminology. In the initial or reference position of the knee, i.e. extension, the body-fixed and the space-fixed systems are defined such that they coincide. In that reference position, the tibial and the femoral origins are located in the posterior insertion of the anterior cruciate ligament on the tibia. The x-axis points to medial, the y-axis aligns with the tibial shaft and points proximally and the z-axis points anteriorly. The representation of the Euler angles in the rotation matrix $\mathbf{R}$ is sequence dependent

$$\mathbf{R} = \mathbf{R}_\phi \mathbf{R}_\psi \mathbf{R}_\theta, \quad (2)$$

where ϕ , ψ and θ represent subsequent rotations about the tibial, body-fixed x-axis (flexion), y-axis (tibial rotation) and z-axis (varus-valgus rotation) respectively (Fig. 1a). The translations that are calculated in this system, i.e. the translations of the tibial origin with respect to the femoral coordinate system E_f , cannot be

Fig. 1. Schematic representation of the body-fixed coordinate axes of the tibia. (a) Subsequent rotations around the axes represent flexion (x-axis), tibial rotation (y-axis) and varus-valgus rotation (z-axis). (b) Loading configuration F_z : A-P forces; F_y : axial forces; M_y : tibial torques.

interpreted directly as anatomical motions. Therefore the translation of a point in the femur relative to the tibial coordinate system is calculated. This point is defined in the reference extension position of the joint to be 15 mm more proximally (positive y-direction) with respect to the coinciding origins of E_f and E_t . This choice is based on the observations of Blankevoort *et al.* (1986) that the helical axes of knee motion cross inbetween the attachments of the cruciate ligaments on the femur.

Motion measurement is performed by measuring relative joint positions after subsequent finite motion steps. If the positions of three non-collinear landmarks, with known positions $\mathbf{x}_i$ relative to E_t , are measured after a finite motion step, which will give $\mathbf{X}_i$ relative to E_f , the three Euler angles and the three translations for

this motion step can be calculated from

$$\mathbf{X}_i = \mathbf{D} + \mathbf{R}\mathbf{x}_i \quad i = 1, 2, 3. \quad (3)$$

In the case where more landmarks are measured ($n > 3$), the kinematic parameters are evaluated by minimizing

$$\sum_{i=1}^n (\mathbf{X}_i - \mathbf{D} - \mathbf{R}\mathbf{x}_i)^2. \quad (4)$$

This ensures a higher accuracy in the kinematic parameters, depending on the number of markers (Selvik, 1974). In the present study both tibia and femur move relative to the absolute laboratory coordinate system. The motions of the tibia relative to the femur are obtained by incorporating identical calculations in the numerical procedure (Selvik, 1974). Lacking well-defined bony landmarks, tantalum pellets inserted in the bones are used for this purpose in the Roentgen Stereo Photogrammetric (RSP) system.

Four fresh knee joints (Table 1) are obtained from autopsy or amputation and kept frozen until the time of use. Prior to preparation, each knee is roentgenographically and manually examined for abnormalities which could lead to exclusion from the experiment. During preparation and experimentation the specimens are at room temperature and kept moist with a Ringers solution. The bones are transected approximately 15 cm above and 15 cm below the joint line. All soft tissues, including the skin, are left intact. The fibula is fixed to the tibia using a Kirchner wire. Six tantalum pellets (0.8 mm diameter) are fixed into the cortex of each bone using a two-component glue.

Each specimen is then fixed in a specially developed motion and loading apparatus (Fig. 2; Hamer, 1982), allowing the tibia six degrees of freedom-of-motion relative to the femur, of which flexion is prescribed in steps of 5° minimally. Except for some small friction forces in the ball-bearings, there are no constraints with respect to the remaining five degrees of freedom. The three rotation axes of the apparatus are flexion (1), tibial rotation (3) and varus-valgus rotation (4), of which flexion and varus-valgus rotation occur in mutually perpendicular planes. Axial translation (6) is possible through a four-bar-linkage mechanism on the femoral side. The anterior-posterior (2) and medial-lateral (5) translations are possible via rotations around two axes located distally on the tibial side, relatively far from the joint. Although the mechanism of the apparatus is explained by using anatomical terminology, it is merely designed to ac-

comodate six degrees of freedom-of-motion, independent of the actual definition used to describe the motions.

In the mounting procedure, the tibial shaft is aligned with rotation axis 3 of the apparatus. The femur is positioned with the intercondylar notch in the intersection of axes 1 and 4 (Fig. 2). The loading apparatus with the joint is then positioned in the laboratory coordinate reference system (defined by a calibration cage). This positioning procedure is executed as well as possible. However, small deviations, discussed later, can occur because of its relatively arbitrary nature.

Loading of the knee is realized by applying weights to those parts of the apparatus that correspond with the respective motions (Fig. 2b). Axial forces are applied through a pair of sheaves and wires on the four-bar-linkage mechanism (a) at the femoral side where the flexion mechanism is located. Consequently, for each specific flexion angle prescribed, the axial force is applied on the femur, but remains directed approximately along the tibial axis. Due to motions of the tibia, small deviations of the axial force with respect to the tibial shaft axis might occur. Rotation torques are applied through a pair of sheaves on the tibial side (b,c). AP-forces are applied on the tibial side (e,d), independent of tibial rotation and torques. It must be stressed at this point that the apparatus was designed to accommodate freedom of motion to the joint at moderate loading, rather than to test knee joint behavior at high loads.

Each knee is tested in a number of flexion series with a specific loading configuration. A flexion series consists of a number of flexion positions (5–8) from extension to approximately 95° of flexion. The loading configurations of the flexion series and the numbers of specimen that were tested in a specific flexion series are listed in Table 2. The loads applied (Fig. 1b) are axial forces (F_y) of 150 and 300 N, AP-forces (F_z) of 30 and 45 N and tibial torques (M_y) of 3 and 6 Nm. These moderate loads are used only to investigate knee motion within and along the envelope of passive freedom-of-motion, where deformations of the soft structures, i.e. the ligaments, menisci and cartilage, are relatively small. At the start of a particular flexion series, the loads of that series are applied and remain unaltered for the subsequent flexion steps. Time dependent effects are neglected, since the position measurements are made after several seconds.

At each flexion position, two Roentgen exposures of the bones with the markers are taken. The exposures are evaluated on a two-dimensional digitizer*, measuring the marker images with an accuracy of 20 µm. A computer program calculates the kinematic parameters which describe the finite motions of the tibia relative to the femur after each motion step.

Before the experiment begins, the relative positions of the Roentgen foci, the photographic plate and the

Table 1. Characteristics of the tested knee joint specimens

Specimen	Left/Right	Age	Sex
1	R	43	M
2	L	53	F
3*	R	74	F
4*	L	74	F

* Bilateral.

* Aristo 104-S®, equipped with an Aaton® video camera.

reference markers (which are imaged in every double exposure, see Fig. 3) are determined with an exposure of a calibration cage with 18 markers divided over two parallel plates, of which the positions are *a priori* known (Selvik, 1974). The reference plate (Fig. 3) then defines the laboratory coordinate system relative to which the joint is aligned in extension.

After the experiments, the joint is dissected and the posterior apex of the tibial anterior cruciate insertion region is marked with an additional tantalum pellet to define the origin of the coordinate systems in extension, by evaluating an additional double exposure.

Further tantalum markers are placed to define the insertion regions of the four major ligaments on both bones (van Dijk *et al.*, 1979), and the articular surfaces are stereo photographed (Huiskes *et al.*, 1985b), in order to obtain geometric information for later data analysis with computer models (Wismans *et al.*, 1980; Andriacchi *et al.*, 1983).

ACCURACY

The accuracy of the kinematic data depends on a number of factors, for example the accuracy of the three-dimensional reconstruction of the positions of the tantalum pellets, the number of pellets defining the rigid bodies and the relative positions of the pellets in the bones (Woltring *et al.*, 1985). An estimate for the precision of the spatial coordinates of the markers is calculated by remeasuring a series of roentgenograms. The precision of the translations and the Euler rotations is estimated in two ways: by remeasuring a series of roentgenograms and by repeating specific flexion series.

First a series of roentgenograms is remeasured and the spatial coordinates of the markers and the kinematic parameters are recalculated. Consequently, two sets of parameters are obtained, the original and the remeasured set. These two sets of duplicate values are used to calculate the standard deviation according to (Mikhail, 1976)

$$\text{S.D.} = \sqrt{\frac{\sum_{i=1}^n (P_{i1} - P_{i2})^2}{2n}}$$

in which S.D. = standard deviation, P_{i1} = result of first measurement, P_{i2} = result of second measurement, and n = number of duplicate values.

Secondly, specific flexion series are repeated as a whole, thus in fact repeating a part of the experiments. The time lapse between the original and the duplicate series is 6 h. The two sets of translations and Euler rotations, from the first and the second experiment, are again used to calculate the standard deviations in the same manner.

The reconstruction of the spatial coordinates of the markers was performed with an estimated error of less than 50 μm (S.D.). The x-coordinate, which direction is towards the Roentgen tubes, is less precise (50 μm S.D.)

as compared to the y- and z-coordinates (30 μm and 20 μm , respectively). The standard deviations for the translations and Euler angles are given in Table 3. Those for the repeated reconstruction give an estimate for the precision of the RSP system and those for the repeated experiment an estimate for the reproducibility of the experiments. It must be noted, however, that the specimens remained fixed in the motion rig between the first and the repeated experiment. Hence, deviations in the positioning procedure are not accounted for in these accuracy tests.

RESULTS

The rotation envelope of passive knee motion

The limits of tibial rotation are not well specified due to the non-linear torque-rotation characteristics of the knee (Wang and Walker, 1974). An example of a torque-rotation curve is given in Fig. 4. Evidently, every choice of torques defining the rotation limits is an arbitrary one. In this case the limits of internal and external rotation are defined as the amount of rotation occurring at torques of ± 3 Nm. This amount is easily realized *in vivo*, by manual handling of the lower leg in clinical practice.

The limits of tibial rotation, as defined by the ± 3 Nm torques, are shown in Fig. 5 as function of flexion for the four specimens. These limits determine the freedom-of-motion range of the knee, from full extension up to approximately 95° of flexion, and are together considered to comprise the rotation envelope of passive knee motion. The increase of the freedom-of-motion ranges with torques of ± 6 Nm, as compared to ± 3 Nm, is relatively small (Fig. 6), due to the relatively high stiffness in that region (compare with Fig. 4).

The total rotatory laxity, i.e. the total amount of tibial rotation between the limits, is shown in Fig. 7 as function of flexion for all specimens. There is a close quantitative similarity of the bilateral knees (3 and 4). Evidently, the total amounts of laxity as function of flexion among the three individuals are different, but the shapes of the curves are remarkably similar.

The simultaneously occurring (coupled) varus-valgus rotations for these pathways along the envelope of motion are shown in Fig. 8. There is a qualitative similarity among the four knees in the sense that they tend towards the valgus direction when moving along the internal pathway, and towards the varus direction when moving along the external pathway. Quantitatively, however, the differences, although small in absolute sense when compared to the flexion and tibial rotations, are large in a relative sense, which is especially remarkable for the bilateral knees (3 and 4). The causes for these differences are discussed later.

The translations of the tibial origin, i.e. the posterior insertion of the anterior cruciate ligament, for the motion pathways along the envelope of motion are highly reproducible (Fig. 9). Because the x-translation

Fig 2. Knee joint motion and loading apparatus. F = femur, T = tibia. (a) **Movabilities:** 1: flexion-extension, 2: anterior-posterior translation, 3: internal-external rotation, 4: varus-valgus rotation, 5: medial-lateral translation, 6: proximal-distal translation. (b) **Loading configurations:** A: axial force, B and C: internal and external torques, D and E: anterior and posterior forces.

Table 2. Loading program showing the number of specimens tested under given combinations of forces and torques

		Rotation torque (Nm)			
		± 6	± 3	0	x
Zero forces		2	4	4	1
Axial force	150	2	4	4	0
(N)	300	2	4	4	0
A-P force	± 30	0	2	4	2
(N)	± 45	0	0	2	2

x = Tibial rotation suppressed.

(lateral-medial) is relatively small, the pathway of this point can be considered as occurring approximately in the sagittal plane. It must be noted at this point, that these translations depend on the location of the particular base point chosen. Hence, they cannot be classified as the relative translations of the tibia as a whole. Relative translations of a point in the femur with respect to the tibia can be expected to be more physically meaningful. This is illustrated in Fig. 9b for one specimen. This base point is located inbetween the femoral attachments of the cruciate ligaments, which is the region where the helical axes of knee motion cross (Blankevoort *et al.*, 1986). Hence these translations are small. Femoral roll-back, which is described as the posterior shift of the tibial-femoral contact areas, cannot be noticed from the z-translations. But again, the translations depend on the specific choice of a point.

The effects of axial loading

In the fully unloaded situation there is no single consistent motion pathway for all four specimens. Two

Fig. 3. Schematic representation of the experimental Roentgenstereophotogrammetric configuration. The knee joint specimen is placed in front of the reference plate and the filmcassette. At each flexion position the specimen is exposed by two Roentgen tubes.

examples of the motion pathways in one single specimen for the completely unloaded case are given in Fig. 10 (dotted curve). The differences between the two curves are the effect of small friction forces in the motion device, indicating that within the envelope of motion freedom, the actual motion pathways are very susceptible to minor changes in the external loads.

However, when small (300 N) axial forces are applied, each knee shows a consistent motion pattern,

Table 3. Standard deviations of the Euler rotations and the translations of specimens 3 and 4, for the repeated reconstruction of the Roentgenstereophotogrammetric measurements, and the repeated experiment

Standard Deviations (S.D.)	Repeated reconstruction	
	specimen 3 (n = 9)	specimen 4 (n = 13)
Rotations (°)		
Flexion (ϕ)	0.09	0.07
Rotation (ψ)	0.07	0.07
Varus-valgus (θ)	0.06	0.07
Translations (mm)		
x	0.10	0.10
y	0.10	0.04
z	0.11	0.07
	Repeated experiment	
	specimen 3 (n = 13)	specimen 4 (n = 14)
Rotations (°)		
Flexion (ϕ)	0.11	0.21
Rotation (ψ)	0.22	0.30
Varus-valgus (θ)	0.16	0.37
Translations (mm)		
x	0.10	0.18
y	0.10	0.08
z	0.11	0.19

Fig. 4. Torque-rotation curves of a knee joint specimen at 25° flexion. The limits of tibial rotation are chosen at torques of ± 3 Nm.

Fig. 5. The limits of internal and external rotation as function of flexion as measured in four specimens.

although no quantitative agreement is found in this case among the four specimens (Fig. 10). The tibia tends to move towards internal rotation, thus indicating that the axial forces produce internal torques. These resulting torques are relatively small, because all motion pathways remain well within the envelope of motion. The fact that one specimen (no. 3) reacts differently from the others, is probably a combined effect of a slightly different alignment, as discussed later and again, the susceptibility of the motion pathway to small differences in the external load, within the envelope of motion.

Fig. 6. The limits of internal (INT.) and external (EXT.) rotation at two flexion angles (20° and 90°) with torques of ± 3 and ± 6 Nm. Specimens 3 and 4.

Fig. 7. Tibial rotatory laxity as function of flexion of four specimens.

Contrarily to effects within the envelope of motion, the axial forces have relatively little influence on the motion patterns along the envelope of motion: tibial rotations (Fig. 11) and varus-valgus rotations change less than 2° and the translations less than 2.5 mm, maximally.

The effects of AP-loading

The application of AP-forces results in comparable effects (Fig. 12). An anteriorly directed force tends to produce an internal pathway and a posteriorly directed force an external pathway. This mechanism must again

Fig. 8. Varus-valgus rotations as function of flexion for the internal (INT.) and external (EXT.) motion pathways of four specimens.

be due to small reaction torques caused by loading of the ligaments and the capsule. For instance, a posterior force will tense the posterior cruciate ligament which is attached on the medial side of the femur and is therefore likely to produce an external tibial torque. These torques are small because the resulting motion pathways remain well within the limits of tibial rotation. The differences in the amount of tibial rotation upon AP-loading can again be caused by small differences in the alignment of the specimens in the motion rig. Evidently, a small deviation in the axial alignment of a specimen relative to the motion rig, and hence also relative to the AP-force direction, can result in notable differences in the torques produced by these forces. This is likely, the more so since increased internal rotation of a specimen usually coincides with increased external rotation (Fig. 12).

That the torques produced by the AP-forces are small is again evident from their effects on the rotation envelope of motion, which are hardly noticeable (Fig. 13). The varus-valgus rotations are only slightly affected, less than 1° . The AP-forces do, of course, affect the translations, because of the AP-laxity present in the joint. The maximum deviations that are found are 4 mm for the y- and z-components and 2 mm for the x-translation (in specimen 2).

THE ALIGNMENT PROBLEM

It is difficult to uniquely identify natural landmarks in the knee, enabling a precise and consistent definition of the anatomical reference system. The procedure

used to align the specimens in the experimental frame must by necessity rely on visual confirmation by the investigator. The arbitrary nature of this procedure may produce slightly different anatomical orientations in the motion rig and the laboratory coordinate system for each specimen. The consequences of this are two-fold. First, small deviations in the loading configuration between specimens relative to the joint anatomy may occur and second, the coordinate systems used to represent the Euler rotation angles may slightly differ between specimens, relative to the anatomy. These small deviations can account for some of the quantitative differences found in the results.

Evidently, in view of the 'firmness' of the envelope of motion pathways and the susceptibility of the joint mechanism to small changes in external loads within the freedom-of-motion region, these deviations in the loading configuration will have an effect in the freedom-of-motion region in particular, when tibial rotation torques are absent or small. This is clearly reflected in the effects of axial and AP-loading in the absence of tibial rotation torques, discussed in the previous section. In addition, the motion rig is constructed such that the axial force works, for each flexion position, in the sagittal plane of the femur (at least in the plane which was defined as such during the alignment procedure), under an angle equal to the prescribed flexion angle with the femoral y-axis in each specific flexion position. However, this direction does not necessarily precisely coincide with the tibial y-axis, since the tibia has five degrees of freedom left to find its own position. This is another reason why quantitative differences in the motion patterns within the susceptible freedom-of-motion region are not surprising.

The firm envelope-of-motion pathways, however, with torque loading present, are less sensitive to these small differences in the loading configuration, but they are affected by differences in the definition of the anatomical reference system. These effects are investigated by simulating different reference systems for the joint and recalculating the motion parameters. A deviation of the orientation of the space-fixed coordinate system will change the coordinates of the landmark positions according to

$$\mathbf{X}_i' = \Delta \mathbf{R} \mathbf{X}_i, \quad (5)$$

in which $\mathbf{X}_i'$ denotes the position of landmark i with respect to a space-fixed coordinate system E_f' which is rotated $\Delta \mathbf{R}^{-1}$ with respect to the original coordinate system E_f . The finite motion equation (3) will then transform in

$$\mathbf{X}_i' = \Delta \mathbf{R} \mathbf{D} + \Delta \mathbf{R} \mathbf{R} \mathbf{x}_i. \quad (6)$$

The initial position $\mathbf{x}_i$ is also expressed with respect to the rotated coordinate system, $\mathbf{x}_i = \Delta \mathbf{R}^{-1} \mathbf{x}_i'$. Hence, finite motion can be described as

$$\mathbf{X}_i' = \Delta \mathbf{R} \mathbf{D} + \Delta \mathbf{R} \mathbf{R} \Delta \mathbf{R}^{-1} \mathbf{X}_i'. \quad (7)$$

This means that the same motion is expressed with different translation components $\mathbf{D}' = \Delta \mathbf{R} \mathbf{D}$ and a different rotation matrix $\mathbf{R}' = \Delta \mathbf{R} \mathbf{R} \Delta \mathbf{R}^{-1}$. Conse-

Fig. 9(a). Translations of the tibial origin for the motion pathways along the envelope of motion in each of the four specimens. INT. = internal pathway, EXT. = external pathway.

quently, the Euler angles on which the rotation matrix depends are influenced by differences in the orientation of the coordinate system.

The present test consists of subsequent variations in the reference positions of the coordinate system E_f and E_t by subsequent small rotations $\delta = \pm 5^\circ$ around the x-axis, y-axis and z-axis, respectively (Fig. 14).

In summary, the motions remain the same in this test, but they are expressed in slightly different reference systems, simulating small differences in the alignment. The results of these variations for the motion pathway along the internal envelope of motion (internal torque of 3 Nm), expressed as the internal and the varus-valgus rotations as functions of flexion (speci-

men 3) are shown in Fig. 15. Generally speaking, the effects on the values of the flexion angles are small, those on the tibial rotation angles moderate, and those on the varus-valgus rotation angle are large. Alignment around the flexion axis (Fig. 15a), which is also the easiest to perform in a reproducible way, has a minor effect only. The alignments around the tibial and varus-valgus axes (Figs 15b and c), which are more difficult to perform in view of the joint anatomy, have more serious consequences. The effects on the motion pathways along the external envelope of motion are similar.

Comparing these results with Figs 5 and 8 suggests that some of the quantitative intra-specimen dif-

Fig. 9(b). Translations of the femoral base point with respect to the tibia for the motion pathways along the envelope of motion (specimen 1).

Fig. 10. Tibial rotation as function of flexion. Dotted line: specimen 1, unloaded. Straight lines: specimens 1 through 4 with an axial force of 300 N.

ferences in the shapes of the curves can easily be explained by small deviations in the alignment. The more so, since the total tibial rotation laxity as function of the flexion, which is much less sensitive to alignment, shows very little differences in the shapes of the curves (Fig. 7).

DISCUSSION

Conceptually, the knee joint has six degrees of freedom-of-motion: three translations and three rotations. Several of these are either suppressed or coupled, due to the constraints provided for by the ligamentous and bony structures. In flexion-extension and tibial rotation, the passive joint will move within a

Fig. 12. Tibial rotation as function of flexion with anterior and posterior directed forces as measured in four specimens.

Fig. 11. Limits of tibial rotation at two flexion angles as influenced by moderate axial forces.

Fig. 13. Limits of tibial rotation at two flexion angles as influenced by moderate A-P forces (specimens 1 and 2)

Fig. 14. Schematic representation of the rotation of the reference coordinate system of the tibia around the y-axis.

Fig. 15. The influence of different orientations of the coordinate reference system on tibial rotation and varus-valgus rotation. The internal motion pathway is reconstructed with respect to six coordinate reference systems, which are obtained by subsequent rotation of the coordinate system with -5 and +5 degrees around each coordinate axis (specimen 3). int. = internal rotation. v/v = varus-valgus rotation. (a) Rotation around x-axis. (b) Rotation around y-axis. (c) Rotation around z-axis.

certain region without noticeable resistance to applied loads (e.g. Fig. 4). This region can be considered as a region of extremely low stiffness or as a region of freedom-of-motion. In this study the latter concept is adopted, whereby the joint is considered as a kinematic mechanism surrounded by elastic structures. This makes it easier to understand and explain the kinematic characteristics of the joint and the role of the supporting structures. To this effect, the motion region is divided into two parts, one within the envelope of passive motion, where stiffness does not play a role and one outside the envelope, where stiffness does play a role. The envelope itself is obtained by defining the limits of motion as the limits of tibial rotation from extension to 100 degrees of flexion. The motion envelope can be extended by using appropriate definitions to hyperextension and beyond 100 degrees of flexion. AP-laxity is not considered as a degree of freedom-of-motion. In a strict sense, it could be considered as such, depending on the definition of the stiffness level used to decide whether a noticeable resistance to AP-forces is present or not. In a general sense, other possible motions may show some relatively small laxity depending on the elastic nature of the constraints. Also in this case the motion limits could be defined at certain levels and be included in the envelope of motion concept.

In the unloaded configuration, there is no sign of an obligatory external rotation during extension, the so-called 'screw-home mechanism', at least not as a passive characteristic of the knee. This mechanism is described by several authors (Braune and Fischer, 1891; Hallen and Lindahl, 1966; Trent *et al.*, 1976; Markolf *et al.*, 1976) and is probably due to the function of the active stabilizers (muscles) in combination with geometric and ligamentous constraints. This is illustrated in the present experiments, where small axial forces tend to produce internal rotation of the tibia, thus creating, due to the ligament constraints, a motion pathway towards external rotation in the last phase of extension (Fig. 10). Conversely, when an external torque is applied on the tibia, the knee follows a motion pathway along the external rotation envelope and creates an internal rotation during the last phase of extension. A study from Draganich and Andriacchi (1985) shows that the quadriceps forces have a resultant internal torque on the tibia. Consequently, the motion pathway during extension is along the internal limits of rotation, resulting in the, previously described, 'screw-home mechanism' towards external rotation in the last phase of extension. This might explain the internal motion pathways found in studies from Lipke *et al.* (1981), Hungerford *et al.* (1984) and Kurosawa *et al.* (1985), where the quadriceps tendon was loaded.

In fact, within the envelope of passive motion, the knee is an unstable system and as a result the actual motion pathway is very susceptible to small loads. Small changes in the directions of forces can have extensive effects on the resulting tibial rotations, whereas motion pathways along the envelope of

motion are consistent and reproducible. The total rotatory laxity pattern as function of flexion is highly reproducible, although the amount of laxity depends on the individual specimen (Fig. 7). This would suggest that 'knee laxity' per individual can be approximated in a single-value parameter, at least for rotatory laxity. The shape of the laxity vs flexion curves are similar among different specimens and only one laxity parameter can describe interindividual differences. Comparable results were found in this respect by van Kampen *et al.* (1986, 1987) and Nielsen *et al.* (1984a, b), as illustrated in Fig. 16, where rotatory laxity data from the literature are summarized. In the above studies the same definition was used for the limits of tibial rotation as in the present study. In a study by Markolf *et al.* (1976), where the forces and torques were manually applied through instrumented force handles, the de-

Fig. 16. Comparison of rotatory laxity data from literature. WW: Wang and Walker (1974), *in vitro*, eight individual specimens. Laxity at ± 0.5 Nm, no compressive force. HW: Hsieh and Walker (1976), *in vitro*, four individual specimens. Laxity at ± 5 Nm, no compressive force. M: open circles: Markolf *et al.* (1976), *in vitro*, average of 35 specimens. Laxity between breakpoints of stiffnesses at ± 8 Nm and neutral stiffness of torque-rotation curves; no compressive force. M: closed circles: Markolf *et al.* (1981), *in vitro*, averages of eight specimens. Laxity at ± 5 Nm, no compressive force. H: Hungerford *et al.* (1984), *in vitro*, one specimen. Laxity at ± 3 Nm, simulated standing position under a weight of 222 N and quadriceps loading; increased quadriceps forces with flexion (deduced from graph, a typical case). S1: Shoemaker *et al.* (1983), *in vitro*, averages of unknown number of specimens. Laxity at ± 5 Nm, no compressive force. S2, S3, S4: Shoemaker and Markolf (1982), *in vivo*, averages of 20 specimens. Laxity at ± 10 Nm, no compressive forces, relaxed musculature. S2: hips flexed; S3: hips extended, S4: hips flexed or extended have the same values. N: Nielsen *et al.* (1984a, b), *in vitro*, two individual specimens out of 20, one with high and one with low laxity. Laxity at ± 3 Nm, no compressive force (deduced from graphs). K: van Kampen *et al.* (1986, 1987), *in vitro*, three individual specimens. Laxity at ± 3 Nm, quadriceps loaded (100 N). 1, 3: Specimens 1 (highest laxity) and 3 (lowest laxity) of the present study.

finition of the limits of tibial rotation was different. From the torque-rotation curves, the intersections of the tangents of the terminal stiffness at ± 8 Nm and the neutral stiffness were chosen as the limits. Consequently, the corresponding torques are not equal for all flexion angles, with values in the range of approximately 0.5–2 Nm. As a consequence, the average laxity values were lower (Fig. 16). In a later study, Markolf *et al.* (1981) used an instrumented fixture in a MTS testing machine, allowing for unconstrained motion of the knee. The average rotatory laxity values at ± 5 Nm for two flexion angles, 0 and 25°, are comparable to those in the present study (Fig. 16).

Wang and Walker (1974) defined two types of rotatory laxity, an approach which is not widely adopted in the literature. According to these authors, primary laxity is the rotation obtained up to a torque of 0.5 Nm, whereas secondary laxity is the additional rotation occurring between 0.5 Nm and maximum torque levels at 5, 12.5 or 25 Nm. Although the laxity definition remains of an arbitrary nature, the primary laxity does represent the free range of tibial rotation better than in the case of a definition based on a higher torque level, as in the present case. But due to the low stiffness in the 0.5 Nm region, the primary laxity is more susceptible to small variations of the torque and the envelope of motion would be less consistent. Wang and Walker (1974) found from their *in vitro* experiments that the variations in the primary laxities were as much as four times from one specimen to another (Fig. 16), which would suggest that knee laxity depends strongly on the individual subject. Their internal and external secondary laxities for a torque level of ± 5 Nm however showed much less variation, i.e. within a factor of 1.5. The total rotatory laxities for a ± 5 Nm torque were on the average 14° higher than the primary laxity, which is more consistent with the present study and with the result of Hsieh and Walker (1976) (Fig. 16).

Hungerford *et al.* (1984) tested knees in a simulated standing position under a weight of 222 N. The quadriceps tendon was loaded and the degree of knee flexion was controlled by moving the quadriceps. An increase of flexion gave a higher load on the quadriceps tendon and higher tibial femoral contact forces. At lower flexion angles their rotatory laxity results for a normal knee are comparable to ours. However, at 60 and 90° of flexion there is a decreased rotatory laxity due to the increased contact forces (Fig. 16). As shown in the present study, axial forces up to 300 N and AP-forces of 30 N have relatively small effects on the limits of tibial rotation. The axial forces increase the laxity only very slightly (Fig. 11). This effect is probably caused by the lateral femoral condyle sliding off the downward slope of the lateral tibial plateau, on the posterior part when internally rotated and on the anterior part when externally rotated, thus creating small additional torques. Conversely, *in vitro* studies from Markolf *et al.* (1981) showed a decrease of

rotatory laxity with axial loading. However, these authors used axial forces of 925 N, three times higher than in the present case, resulting in higher deformations of the cartilage and the menisci, thus increasing the articular resistance against tibial rotation. It has to be recognized that joint forces during physical activity are much higher than those used in this study, and the stabilizing mechanisms due to increasing conformity of the joint surfaces and menisci will certainly come into effect, thus possibly protecting the ligaments. However, the concept of the envelope of passive knee joint motion remains important, since it stresses the fact that motions can only occur within the limits of joint laxity. Beyond these limits possible damage of the joint structures might occur. Secondly, this concept is useful for the development of diagnostic procedures for ligament lesions (Edixhoven *et al.*, 1987).

Evidently, the relative imprecision of the anatomical reference system and the resulting arbitrary nature of the alignment plays an important role in the precise evaluation of knee joint motions. In the first place, within the envelope of motion, the motion pathway is very sensitive to small load changes and alignment differences of the applied forces. More importantly, the alignment of the coordinate systems with respect to the joint anatomy influences the kinematic parameters, i.e. the values of the translations and the Euler rotations. Recently Grood and Suntay (1983) and Kurosawa *et al.* (1985) have also stressed the importance of using reference systems which must be related to the joint anatomy in a reproducible way. The anatomic structures suitable for this purpose are only those that are functional in the mechanical behavior of the joint, such as the femoral condyles (Kurosawa *et al.*, 1985). The functional structures, i.e. the articular surfaces, the ligaments and their attachments are relatively inaccessible if the joint is to be left intact during the experiments. In this study the shaft axes and the femoral epicondyles were used in the positioning procedure. Once the joint specimen is mounted in the loading device, the reference system for the loading configuration cannot be altered. In this study the choice was made not to alter the reference systems afterwards but to demonstrate the influence of slightly differing systems on the kinematic parameters. The application of the Joint Coordinate System proposed by Grood and Suntay (1983) has the same practical problem of precise definition relative to the joint anatomy. Their definitions of the rotations are not equivalent to the corresponding rotations in the present reference system. A completely equivalent system would imply a rotation sequence ϕ , θ , ψ (flexion, varus-valgus rotation, tibial rotation) instead of ϕ , ψ , θ . Because the tibial rotation and the varus-valgus rotation are relatively small, the results would hardly differ in this case, as is demonstrated in the Appendix.

In this experimental study an accurate Roentgen stereophotogrammetric system and a six degrees of

freedom motion and loading apparatus are used to assess the passive freedom-of-motion characteristics of the human knee. Kinematic data for a number of loading conditions are obtained of four specimens. Together with geometric data of the articular surfaces (Huiskes *et al.*, 1985b) and ligament insertions (van Dijk *et al.*, 1979), these findings can be used as a database for verifying and enhancing existing knee joint models (e.g. Wismans *et al.*, 1980; Andriacchi *et al.*, 1983). The RSP method is tedious and time-consuming, in particular where it concerns the evaluation of the roentgenograms. Hence, this method is not suitable to experiment on large numbers of specimens. However, due to its high precision, the method is extremely valuable for the individual approach used in this study, because small individual differences in behavior among knee joint specimens and small deviations due to loading variations can easily be detected.

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REFERENCES

- Andriacchi, T. P., Mikosz, R. P., Hampton, S. J. and Galante, J. O. (1983) Model studies of the stiffness characteristics of the human knee-joint. *J. Biomechanics* 16, 23–29.
- Blacharski, P. A., Somerset, J. H. and Murray, D. G. (1975) A three-dimensional study of the kinematics of the human knee. *J. Biomechanics* 8, 375–384.
- Blankevoort, L., Huiskes, R. and de Lange, A. (1986) Helical axes along the envelope of passive knee joint motion. *Proceedings of the 32nd Annual Meeting of the Orthopedic Research Society*, p. 410.
- Braune, W. and Fischer, O. (1891) Die bewegungen des kniegelenks. *Abh. math.-phys. kl. Königl. sächs. Ges. Wiss.* 17, 1–76.
- Chao, E. Y. S., Laughman, R. K., Schneider, E. and Stauffer, R. H. (1983) Normative data of knee joint motion and ground reaction forces in adult level walking. *J. Biomechanics* 16, 219–233.
- Dijk, R. van, Huiskes, R. and Selvik, G. (1979) Roentgenstereophotogrammetric methods for the evaluation of the three-dimensional kinematic behaviour and cruciate ligament length patterns of the human knee joint. *J. Biomechanics* 12, 727–731.
- Draganich, L. F. and Andriacchi, T. P. (1985) Three dimensional moments produced by the knee joint musculature. *Proc. of the 31st Annual meeting of the Orthopedic Research Society*, p. 137.
- Edixhoven, Ph., Huiskes, R., de Graaff, R., van Rens, Th. J. G. and Slooff, T. J. (1987) Accuracy and reproducibility of instrumented knee-drawer tests. *J. Orthop. Res.* 5, 378–387.
- Fukubayashi, T., Torzilli, P. A., Sherman, M. F. and Warren, R. F. (1982) An *in vitro* biomechanical evaluation of anterior-posterior motion of the knee. *J. Bone Jt Surg.* 64A, 258–264.
- Goldstein, H. (1959) *Classical Mechanics*. Reading, Addison-Wesley.
- Grood, E. S. and Suntay, W. J. (1983) A joint coordinate system for the clinical description of three-dimensional motions: applications to the knee. *J. biomech. Engng* 105, 136–144.
- Hallen, L. G. and Lindahl, O. (1966) The screw-home movement in the knee-joint. *Acta orthop. scand.* 37, 97–106.
- Hamer, A. (1982) A static knee-loading apparatus. *3rd Meeting of the European Society of Biomechanics*, Nijmegen, The Netherlands.
- Hsieh, H.-H. and Walker, P. S. (1976) Stabilizing mechanism of the loaded and unloaded knee joint. *J. Bone Jt Surg.* 58A, 87–93.
- Huiskes, R., Dijk, R. van, Lange, A. de, Woltring, H. J. and Rens, Th. J. G. van (1985a) Kinematics of the human knee joint. *Biomechanics of normal and pathological human articulating joints*. (Edited by Berme, N., Engin, A. E. and Correia da Silva, K. M.), pp. 165–187. Martinus Nijhoff, Dordrecht.
- Huiskes, R., Kremers, J., Lange, A. de, Selvik, G., Woltring, H. J. and Rens, Th. J. G. van (1985b) An analytical stereophotogrammetric method to determine the three-dimensional geometry of articular surfaces. *J. Biomechanics* 18, 559–570.
- Hungerford, D. S., Kenna, R. V. and Haynes, D. W. (1984) Relevant biomechanics of the knee for knee replacements. *Total Knee Arthroplasty: a Comprehensive Approach* (Edited by Hungerford, D. S., Krackow, K. A. and Kenna, R. V.), pp. 21–34. Williams Wilkins, Baltimore London.
- Kampen, A. van, Huiskes, R., Blankevoort, L. and Rens, Th. J. G. van (1986) The three dimensional tracking pattern of the patella in the human knee joint. *Proceedings of the 32nd Annual meeting of the Orthopedic Research Society*, p. 386.
- Kampen, A. van (1987) The three-dimensional tracking pattern of the patella. Dissertation, University of Nijmegen, The Netherlands.
- Kinzel, G. L., Hall, A. S. and Hillberry, B. M. (1972) Measurement of the total motion between two body segments—I. Analytic development. *J. Biomechanics* 5, 93–105.
- Kurosawa, H., Walker, P. S., Abe, S., Garg, A. and Hunter, T. (1985) Geometry and motion of the knee for implant and orthotic design. *J. Biomechanics* 18, 487–499.
- Lafortune, M. A. (1984) The use of intra-cortical pins to measure the motion of the knee joint during walking. Ph.D. dissertation, Pennsylvania State University, University Park, PA, U.S.A.
- Lange, A. de, Kauer, J. M. G. and Huiskes, R. (1985) Kinematic behaviour of the human wrist joint: a roentgen-stereophotogrammetric analysis. *J. Orthop. Res.* 3, 56–64.
- Lipke, J. M., Janecki, C. J., Nelson, C. L., McLeod, P., Thompson, C., Thompson, J. and Haynes, D. W. (1981) The role of incompetence of the anterior cruciate and lateral ligaments in anterolateral and anteromedial instability. *J. Bone Jt Surg.* 63A, 954–960.
- Markolf, K. L., Mensch, J. S. and Amstutz, H. C. (1976) Stiffness and laxity of the knee—The contributions of the supporting structures. *J. Bone Jt Surg.* 58A, 583–593.
- Markolf, K. L., Barger, W. L., Shoemaker, S. C. and Amstutz, H. C. (1981) The role of joint load in knee stability. *J. Bone Jt Surg.* 63A, 570–585.
- Mikhail, E. M. (1976) *Observations and Least Squares*. Dun-Donneley, New York.
- Murphy, M. C., Zarins, B., Jasty, M. and Mann, R. W. (1984) *In vivo* measurement of the three dimensional skeletal motion at the normal knee. 1984 *Advances in Bioengineering* (Edited by Spilker, R. L.) pp. 64–85. American Society of Mechanical Engineers, New York.
- Nielsen, S., Ovesen, J. and Rasmussen, O. (1984a) The anterior cruciate ligament of the knee: An experimental study of its importance in rotatory knee instability. *Arch. Orthop. Traum. Surg.* 103, 170–174.
- Nielsen, S., Rasmussen, O., Ovesen, J. and Andersen, K. (1984b) Rotatory instability of cadaver knees after transection of collateral ligaments and capsule. *Arch. Orthop. Traum. Surg.* 103, 165–169.

Piziali, R. L., Rastegar, J., Nagel, D. A. and Schurman, D. J. (1980) The contribution of the cruciate ligaments to the load-displacement characteristics of the human knee. *J. biomech. Engng* **102**, 277-283.

Selvik, G. (1974) A roentgenstereophotogrammetric method for the study of the kinematics of the skeletal system. Dissertation, University of Lund, Sweden.

Shoemaker, S. C. and Markolf, K. (1983) Contributions of the ACL and MCL to A-P and torsional stability of the loaded knee *in vitro*. *Proceedings of the 29th Annual Meeting of the Orthopedic Research Society*, p. 98.

Shoemaker, S. C. and Markolf, K. L. (1982) *In vivo* rotatory knee instability. *J. Bone Jt Surg.* **64A**, 208-216.

Suntay, W. J., Grood, E. S., Hefzy, M. S., Butler, D. L. and Noyes, F. R. (1983) Error analysis of a system for measuring three-dimensional joint motion. *J. biomech. Engng* **105**, 127-135.

Trent, P. S., Walker, P. S., Wolf, B. (1976) Ligament length patterns, strength and rotational axes of the knee joint. *Clin. Orthop. Rel. Res.* **117**, 263-270.

Walker, P. S., Shoji, H. and Erkman, M. J. (1972) The rotational axis of the knee and its significance to prosthesis design. *Clin. Orthop. Rel. Res.* **89**, 160-170.

Wang, C-J. and Walker, P. S. (1974) Rotatory laxity of the human knee joint. *J. Bone Jt Surg.* **56A**, 161-170.

Wismans, J., Veldpaus, F., Janssen, J., Huson, A. and Struben, P. (1980) A three-dimensional mathematical model of the knee-joint. *J. Biomechanics* **13**, 677-685.

Woltring, H. J., Huiskes, R., Lange, A. de and Veldpaus, F. E. (1985) Finite centroid and helical axis estimation from noisy landmark measurements in the study of human joint kinematics. *J. Biomechanics* **18**, 379-389.

Hence

$$R = R_\phi R_\psi R_\theta, \quad (A1)$$

whereby

$$R_\theta = \begin{bmatrix} \cos \theta & -\sin \theta & 0 \\ \sin \theta & \cos \theta & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad (A2)$$

$$R_\psi = \begin{bmatrix} \cos \psi & 0 & \sin \psi \\ 0 & 1 & 0 \\ -\sin \psi & 0 & \cos \psi \end{bmatrix} \quad (A3)$$

$$R_\phi = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos \phi & -\sin \phi \\ 0 & \sin \phi & \cos \phi \end{bmatrix} \quad (A4)$$

The sequence convention of Grood and Suntay (1983) is ϕ , θ , ψ , hence

$$R^* = R_\phi R_\theta R_\psi. \quad (A5)$$

The angles are calculated from the components of the rotation matrix R . Using the corresponding elements of the compound matrix products (A1) and (A5) the following relation between the two rotation conventions can be derived:

$$\phi = \phi^* - \arctan(\tan \psi^* \sin \theta^*) \quad (A6)$$

$$\sin \psi = \sin \psi^* \cos \theta^* \quad (A7)$$

$$\tan \theta = \frac{\tan \theta^*}{\cos \psi^*}. \quad (A8)$$

APPENDIX: ROTATION SEQUENCE OF EULER ANGLES

The representation of the Euler angles in the rotation matrix is sequence dependent. In this study the sequence ϕ (flexion), Ψ (rotation), θ (varus-valgus) is used.

If θ^* and ψ^* are sufficiently small the differences between the two conventions are not dramatic, as is illustrated by the next example: $(\phi^*, \psi^*, \theta^*) = (80, 20, 7.5)$ according to the convention of Grood and Suntay corresponds with $(\phi, \psi, \theta) = 77.28, 19.82, 7.98)$ according to the convention in this study.