624 KSME International Journal, Vol. 12, No. 4, pp. 624~641, 1998

In-vitro Measurement of the Human Knee Joint Motion During Quadriceps Leg Raising

Seonpil Kim*

(Received September 25, 1997)

This paper represents a three-dimensional motion analysis of the human knee joint under given conditions of loading and constraint. As the knee is extended by a known force applied to the quadriceps tendon, relative displacements of the femur, tibia, and patella are measured using a video motion analysis system. The most prominent motion of the tibia is external rotation and anterior displacement relative to the femur during knee extension. The patellar flexion angle decreases from 70° to 0°. The moment arm of the knee extensor mechanism exhibits a characteristic bell shape which peaks somewhere in the 40°-60° region of flexion. In general, the quadriceps force increases as the knee extends from 90° to full extension. The initial increase of quadriceps force results primarily from an increase in the torque exerted by the weight of the lower leg. In the range of 20°-60°, the quardricep force needed to extend the leg remains relatively constant. As the knee approaches full extension, the moment arm decreases due to the fact that the posterior capsule and the ACL begin to tighten in this region. Consequently, the quadriceps force increases rapidly.

Key Words: Human Knee, Kinematics, Screw Analysis, Moment Arm, Knee-Extensor Mechanism

1. Introduction

A number of studies have reported on the relative displacements of the bones at the knee obtained from in vitro experiments (van Dijk et al., 1979 ; Veress et al., 1979 ; Grood et al., 1984 ; van Kampen et al., 1987 ; Reuben et al., 1989 ; Rovick et al., 1991 ; Hirokawa et al., 1991 ; Hirokawa et al., 1992 ; Hefzy et al., 1992 ; Heegard et al., 1994 ; Nagamine et al., 1995). These studies have focused either on the mechanics of the patellofemoral joint (van Kampen et al., 1990 ; Koh et al., 1995; Nagamine et al., 1995) or on that of the tibiofemoral joint (van Dijk et al., 1979 ; Blankevoort et al., 1988). None of these studies, however, specify the three-dimensional movements of the bones for a specific activity in which the loading conditions and constraints are known.

* Technical Center Samsung Motors Inc. Yong-in, Kyungki-do, Korea

LaFortune et al. (1992) measured the relative displacements of the femur, tibia, and patella during normal gait by implanting bone pins in live subjects. However, the forces exerted by the muscle tendons on the bones during the activity were not known. Koh et al. (1995) also used bone pins to measure the relative movements of the femur, tibia, and patellar during maximum, voluntary contractions of the quadriceps muscles *in vivo*. Once again, however, the loading conditions imposed upon the bones could not be quantified. In contrast, van Kampen et al. (1990) and Nagamine et al. (1995) measured the full three-dimensional displacements of the femur, tibia, and patella under known conditions of loading and constraint in vitro. Unfortunately, however, these data do not define the movements of the bones at the knee during an actual task. Instead, these researchers quantified the relative displacements of the bones for passive flexion-extension movements of the knee in which the muscles remain inactive. Many of the experiments reported in the literature also have unknowingly over-constrained the knee, so much so that the resulting motion of the bones no longer represents that of the natural joint.

The experiments described in this paper are aimed at measuring anatomical, three-dimensional knee-joint motion under given conditions of loading and constraint. Specifically, relative displacements of the femur, tibia, and patella are measured in a fresh, cadaveric leg as the knee is extended by a known force applied to the quadriceps tendon. Based on these data, the mathematical basis for the moment arm of the knee extensor mechanism is presented in detail. It is difficult to qualitatively compare the experimental data obtained in this study with those data reported in the literature. In fact, it is difficult to compare the experimental data obtained from the various studies reported in the literature alone. This is because the loading conditions and constraints used in the experiments are inevitably different. In many respects, this was the motivation for undertaking the cadaveric experiments described in this paper.

2. Cadaveric Leg Experiments

2.1 Experimental test rig

Several mechanical knee-joint simulators have been constructed for the purpose of simulating both passive and active movements of the knee (Lewis, et al., 1988 ; Ahmed et al., 1987 ; Kurosawa et al., 1985 ; Rovick et al., 1991). In this study, a relatively simple loading rig was designed and built to test normal, fresh, cadaveric legs during flexion-extension movements of the knee (Fig. 1). This apparatus is comprised of a programmable stepper motor (0.16 Hp, model SX83-62, Parker Inc.), a gear box with a 20 : 1 reduction ratio (model NE34-20, Bayside Precision Inc.), a strain-gauged force transducer (capacity 896N ; model MLP-200, Transducer Techniques Inc.), a preamplifier (model TM-2, Transducer Techniques Inc.), and software for pre- and post-processing of the experimental data (LabView, National Instruments Inc.).

The femur is held stationary within an aluminum holder, which itself is bolted to a flat table.

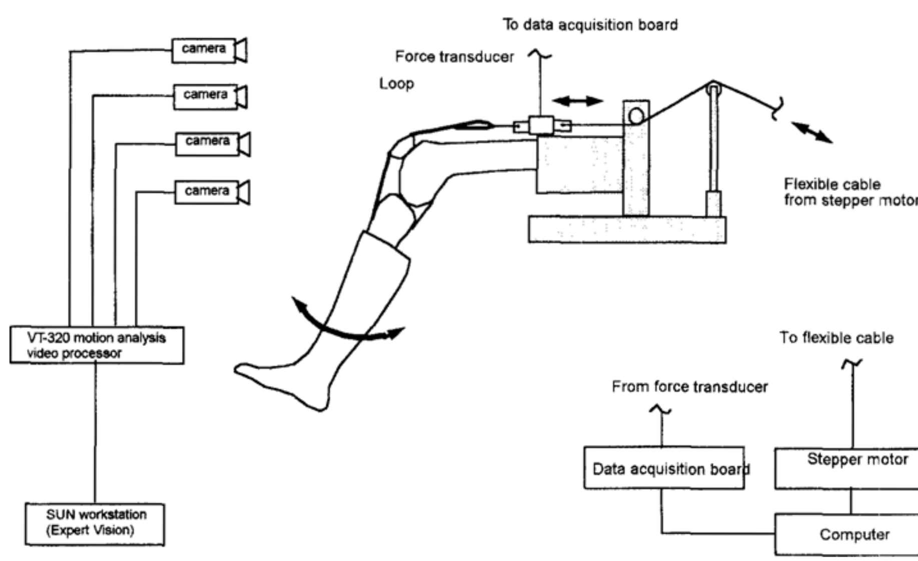


Fig. 1 Schematic drawing of the experimental test rig for the quadriceps leg raise task. The femur was mounted horizontally and fixed to the frame of the table, while the tibia hung vertically by its weight. The flexible wire cable was attached to the quadriceps tendon. The stepper motor controlled the length of the flexible wire causing the knee to extend at a prescribed rate. The force in the wire cable (quadriceps force) was measured during flexion-extension movements of the knee. The motion analysis system measured the relative displacements of the femur, tibia, and patella during the leg raise task.

The femur is potted tightly in the holder using bone cement. The specimen knee is loaded by attaching a flexible wire cable to the quadriceps tendon. The flexible cable is then taken to the programmable stepper motor which, when driven in one direction, winds the cable around its shaft, thereby exerting force on the quadriceps tendon, which extends the knee. To flex the knee, the motor is driven in the opposite direction, which unwinds the cable from the motor shaft so that the leg is lowered under its own weight.

2.2 Materials and preparation

Three normal whole legs were prepared for testing (see Table 1). Prior to the experiments, each specimen was visually inspected to ensure that the knee-joint appeared at least outwardly normal. This preliminary assessment was verified for each specimen when the joint was subsequently opened to perform a bilateral meniscectomy. During the meniscectomy, the menisci and ligaments were examined by a surgeon to ensure that they were originally intact. Furthermore, no visual evidence of degenerative changes was found in the articular surfaces of the patella, in the patellar surfaces of the femoral condyles, in the articular surfaces of the tibia, and in the articular surfaces of the posterior femoral condyles.

Prior to preparation, each specimen was stored in a freezer at-20 degrees Celsius. The night before testing, the specimen was removed from storage and allowed to thaw at room temperature. On the day of testing, the femur was resected 35 cm above a medial-lateral line located approximately at the center of the knee. All passive soft tissues including the muscles were removed from the femur. Only some 10 cm of the quadriceps tendon was left intact. All passive soft tissues including the muscles were then removed from a portion of the tibia, beginning at the medial-lateral line located at the center of the knee joint, up to a point 10 cm below the center of the knee (see Kim, 1996 for details). A 2.5 cm loop was then sutured in the most proximal portion of the quadriceps tendon. This loop was needed to attach the flexible cable from the loading rig to the quadriceps tendon.

Three sets of retroreflective markers were used to measure the relative displacements of the femur, tibia, and patella during flexion-extension movements of the knee. Each set contained three retroreflective markers mounted on a triad. Triads fixed on the patella and tibia were each constructed in the form of a cruciform affixed to a 10 mm

Table 1 Major dimensions of each specimen leg. Femur width is the distance measured between width of the femoral epicondyles. Leg length is the distance from the femoral epicondyle to the tibial malleolus. Leg weight is the weight of the shank and foot combined.

Specimen	Gender	Femur Width (mm)	Leg Length (mm)	Leg Weight (N)
1	F	81	350	20.0
2	M	95	470	32.0
3	F	71	360	20.4

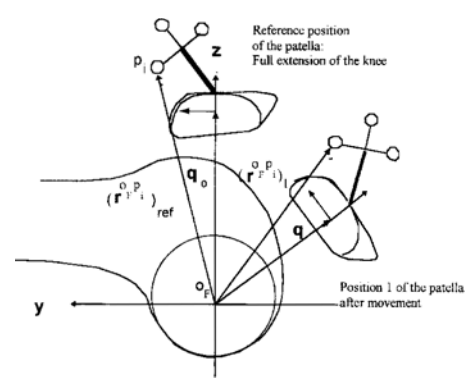


Fig. 2 The triad implanted in the patella: The patella glides downward along the femoral groove from full extension position during knee flexion. All axes of the patellar frame are parallel with those of the coordinate frame of the femur. $(r^{O_F P_i})_i$ is the position vector of markers of the patellar triad during movement. $(r^{O_F P_i})_{ref}$ is the position vector of the markers at the full extension of the knee. q is the position vector of the origin of patellar frame. q_0 is the position vector of the origin at the full extension of the knee.

diameter hub. Two markers were located horizontally 38 mm from the center line of the hub. The third marker was located vertically 38 mm from the top of the hub (see Fig. 2).

Each triad was attached to a 10 cm long steinmann pin, which was then inserted directly into the bone. Two retroreflective pin-markers, each shaped as a 5 mm sphere, were also embedded in the bones. One pin was embedded in the patella at the approximate location of the patellar ligament origin; the other pin was embedded in the tibial tubercle, thereby marking the approximate insertion of the patellar ligament on the tibia. Measurements associated with these markers were used to calculate the direction of the patellar ligament force, and therefore, the moment arm of the patellar ligament at the knee.

Prior to data collection, the specimen knee was placed in full extension and video data were collected to measure the positions of the markers with the leg in this configuration. These data were used to define the reference configuration of the leg. All kinematic measurements were made using a four-camera, 60 Hz, video system (Motion Analysis Inc., Santa Rosa).

2.3 Experimental protocol

For a known force applied to the quadriceps tendon, we measured the relative motion of the bones as the knee was extended from 90° of flexion to full extension. This leg raising task has been studied previously by Grood et al. (1984).

Prior to data collection, the specimen leg hung under its own weight with the knee flexed to 90°. The knee was then extended quasi-statically (i. e., almost zero velocity and zero acceleration of the bones) from 90° flexion to full extension. During this movement, the shortening length of the flexible wire cable was measured. These measurements were then used to plan the trapezoidal velocity profiles of the cable so that the knee could be extended at three different angular speeds : 16 deg/sec, 4.5 deg/sec, and 1.5 deg/sec. This meant that the total time taken to move the knee from 90° flexion to full extension was 5 seconds, 20 seconds, and 60 seconds, respectively.

Subsequent to specifying the length trajectory of the cable during knee extension, the leg was returned to its initial configuration with the knee flexed to 90°. From this position, the knee was extended from 90° flexion to full extension at one of the aforementioned speeds. For each speed, the time history of the applied quadriceps tendon force was measured, together with the time histories of the absolute three-dimensional positions of the markers mounted on the patella and tibia. Marker positions for the femur were not measured during the experiment as the femur remained stationary. For each testing speed, data were collected for three separate trials. These data were subsequently averaged, and the resulting data used to represent the motion of the bones for the speed in question.

3. Analysis

3.1 Position and orientation of the patella and tibia relative to the femur

To describe the motion of the patella and tibia, three markers mounted on a single triad were attached to each bone. The position of each marker on the patella and tibia relative to an inertial reference frame was obtained from kinematic measurements made using the video system. On the basis of these data, it is possible to calculate the position and orientation of the patella and tibia relative to the femur during flexion-extension movements of the knee.

The origin of the femoral reference frame was defined to be at the center of the lateral condyle of the specimen knee. The axes of the femoral reference frame were oriented as follows (Kim, 1996) : the transverse x-axis was defined as the line passing through the center of the medial and lateral condyle of the specimen knee, with the positive direction pointing medially ; the y-axis was defined to be perpendicular to the x-axis and parallel with the long axis of the femur in the sagittal plane, with the positive dirction pointing toward the hip ; and the z-axis was taken as the vector cross-product of the x and y axes, with the positive pointing anteriorly.

In this study, only the *relative* displacements of the bones were measured. The reference configuration of the leg was defined with the knee at full extension. In this configuration, the position and orientation of the patella and tibia relative to the femur were taken to be zero. That is, the reference frames fixed on the patella and tibia were each assumed to be perfectly aligned with the reference frame fixed on the femur when the knee was placed in full extension. Thus, the displacements of the patella and tibia calculated during movement represent the displacement of each bone with respect to its position and orientation in the reference configuration.

Let the position and orientation of the patella and tibia relative to the femur in the reference configuration be given by position ref, and let any other position of the patella and tibia relative to the femur be given by position 1. The change in position and orientation of the patella or tibia from position ref to position 1 is given by

$$(r^{OfPi})_1 = {}_{ref}^{1}[R]\,(r^{OfPi})_{ref} + v \quad (i=1,\ 2,\ 3) \qquad (1)$$

where $(r^{OfPi})_{ref}$ represents the position of a marker mounted on either the tibia or the patella in position ref, $(r^{OfPi})_1$ represents the position of a marker mounted on either the tibia or the patella in position 1, ${}_{ref}^{1}[R]$ is the rotation matrix which contains the direction cosines defining the orientation of either the patella or the tibia in position 1 relative to its orientation in position ref, and v is a vector defining the translation of the origin of either the patellar reference frame or the tibial reference frame as the bone is displaced from position ref to position 1 (see Figs 2 and 3).

We used the algorithm given by Spoor and Veldpaus (1980) to calculate the rotation matrix ${}_{ref}^{1}[R]$ and the translation vector v for each new position of the patella and tibia relative to the position and orientation of each bone in the reference configuration. The translations of the tibia relative to the femur, however, were subsequently expressed in the tibial reference frame, since these quantities correspond more closely with the clinical translations of anterior-posterior displacement, medial-lateral displacement, and proximal-distal distraction of the tibia relative to the femur. The orientation of each bone, on the other hand, is defined by three rotations about

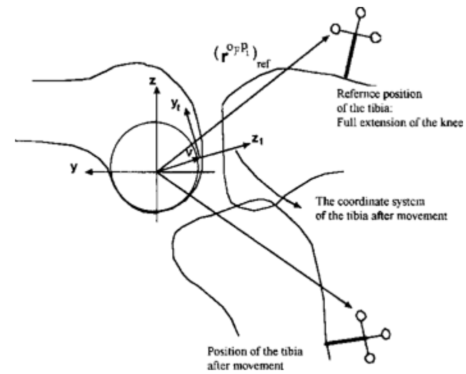


Fig. 3 The triad implanted in the tibia : The coordinate frame of the tibia is coincident with that of the femur at full extension of the knee. $(r^{OfPi})_1$ is the position vector of markers of the tibial triad during movement. $(r^{OfPi})_{ref}$ is the position vector of the markers at the full extension of the knee. v is the position vector of the origin of tibial frame.

mutually perpendicular axes. Values of these angles which define the change in orientation of each bone relative to its orientation in the reference configuration may be calculated using the rotation matrices found in Eq. (1).

Consider, for example, rotations of the patella about the following three mutually perpendicular axes. First, let γ represent the rotation of the patella about the x-axis of the femur (patellar flexion-extension). Next, let α represent the rotation of the patella about the rotated z-axis, which can be labeled as the z'-axis (patellar rotation). Finally, let β represent the rotation about the rotated y'-axis, which can be labeled as the y''-axis (patellar tilt) (see Fig. 3). For Euler rotations of the patella defined in this order (i. e., x-z'-y''), the rotation matrix is given by

$$ {}^{ref}_{1}[R] = \begin{bmatrix} c\alpha c\beta & -s\alpha & c\alpha s\beta \\ c\gamma s\alpha c\beta + s\gamma s\beta & c\gamma c\alpha & c\gamma s\alpha s\beta - s\gamma c\beta \\ s\gamma s\alpha c\beta - c\gamma s\beta & s\gamma c\alpha & s\gamma s\alpha s\beta + c\gamma c\beta \end{bmatrix} \qquad (2)$$

where $c\gamma = \cos\gamma$ and $s\gamma = \sin\gamma$. Note that the order of Euler rotations as defined above (i. e., x-z′-y″) corresponds with the convention recommended by the ISB (Wu and Cavanagh, 1995). The rotation matrices calculated in Eq. (2) can be used to find the corresponding values of the rotation angles γ, α, and β which specify the orientation of the patella relative to its orientation in the reference configuration (Craig, 1986). A similar procedure was carried out to calculate the rotation angles of the tibia which defined the change in orientation of the bone relative to its orientation in the reference configuration.

3.2 Calculation of the finite screw axis

The displacement between two successive positions of each bone can be described by a rotation matrix $^1_2[R]$ and a translation vector v using an equation similar to that of Eq. (1). Thus,

$$(r^{OfPi})_1 = {}^1_2[R](r^{OfPi})_2 + v \quad (i=1, 2, 3) \qquad (3)$$

The rotation matrix and the translation vector may be equivalently defined by a translation along and a rotation about a unique line in space, called the screw axis. Importantly, the position and orientation of the screw axis so calculated is *independent* of the coordinate system used to describe the motion of each bone. The location and direction of an instantaneous screw axis may also be found, but this requires that both the angular velocity of the bone and the linear velocity of at least one point on the bone be known. Since calculation of angular and linear velocities requires numerical differentiation of the measured displacement data, only the locations and directions of the *finite* screw axes were calculated in this study.

The rotational matrix may be solved uniquely for the direction cosines of the screw axis (k) and the equivalent rotation angle (ϕ) of the bone (refer to Craig, 1986). The translation vector v can be separated into two components: one parallel to the screw axis, k, and the other perpendicular to the screw axis. Thus,

$$v = v_n + sk \qquad (4)$$

where s represents the magnitude of the translation of the bone in the direction of the screw axis, k, and v_n is the component orthogonal to k (Fig. 4). The component v_n can be represented by

$$v_n = \phi(r \times k) \qquad (5)$$

where ϕ is the equivalent angle of rotation of the bone and r is a position vector leading from the

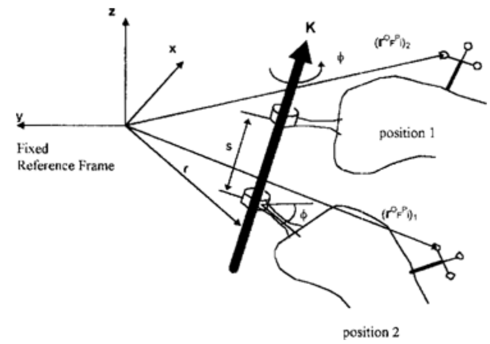


Fig. 4 Screw axis and screw motion. The movement of a rigid body from position 1 into another position 2 can be represented by rotation ϕ about the screw axis and the translation s along the screw axis. The position of the screw axis is indicated by the position vector r from the fixed reference frame.

origin of the femoral reference frame to an arbitrary point on the screw axis. The magnitude of v_n is the minimum (perpendicular) distance from the origin of the femoral reference frame to the screw axis. Thus, Eq. (4) becomes

$$\begin{aligned} v &= \phi(r \times k) + sk \\ &= \phi\left[(r \times k) + \frac{s}{\phi}k\right] \end{aligned} \qquad (6)$$

The position of a point on the screw axis given by the vector $r = (r_x, r_y, r_z)$ can be found in several ways (Kinzel, et al., 1972 ; Spoor and Veldpaus, 1980 ; Suh and Radcliffe, 1978). Spoor and Veldpaus (1980) determined r by defining it to be the vector associated with the *minimum* (perpendicular) distance from the origin of a global reference to the screw axis, k.

The translation s of the bone along the screw axis can be determined by taking the dot product of the translation vector v and the unit vector in the direction of the screw axis, k. Thus, from Eq. (6),

$$s = v \bullet k \qquad (7)$$

Finally, the relative displacement of a bone between two successive positions of the bone can be written in compact form by combining the pure rotation of the bone about the screw axis with the pure translation of the bone along the screw axis. The resulting screw $\mathbf{\$}$, is given by the unit vector k, which defines the direction of the screw axis, the position vector r, which defines the location of the screw axis, the angle of rotation of the body about the screw axis ϕ, and the translation s of the bone along the screw axis. Thus,

$$\begin{pmatrix} \phi k \\ - \\ v \end{pmatrix} = \phi \begin{pmatrix} k \\ -------- \\ r \times k + \frac{s}{\phi} k \end{pmatrix} = \phi \$ \qquad (8)$$

where the screw $\$$ is given by

$$\$ = \begin{pmatrix} k \\ -------- \\ r \times k + \frac{s}{\phi} k \end{pmatrix} \qquad (9)$$

In this study, the locations of the finite screw axes for displacements of the tibia relative to the femur were defined with respect to the origin of the femoral reference frame. Note that s/ϕ the ratio of the translation of the bone along the screw axis to the angle of rotation of the bone about the screw axis, is called the *pitch* of the screw.

3.3 The moment arm of the knee-extensor mechanism

Various estimates for the moment arm at the knee appear in the literature. Smidt (1973) calculated the moment arm of the patellar ligament force by estimating the location of the instantaneous center of rotation of the tibia relative to the femur during flexion-extension movements of the knee. In a similar study, Nisell (1985) calculated the moment arm of the patellar ligament about the tibiofemoral contact point using X-rays to measure the required positions of the femur and tibia in the sagittal plane. Grood et al. (1984) calculated the effective moment arm of the knee-extensor mechanism by measuring the quadriceps force needed to extend a cadaveric leg, and by performing a static analysis for force and moment equilibrium of the patella and tibia. In another entirely different approach, Spoor et al. (1990) estimated the moment arm of the rectus femoris muscle at the knee by calculating the change in length of the muscle as a function of a change in the angle of knee flexion. None of these methods, however, estimate the moment arm of the knee-extensor mechanism in three dimensions, since this requires knowledge of the position and orientation of the finite or instantaneous screw axis of the tibia relative to the femur during flexion-extension movements of the knee.

The actual moment arm of the knee-extensor mechanism can be calculated only once the physical meaning of this quantity has been understood. Under all circumstances, the instantaneous torque exerted by a muscle force to cause pure rotation of a limb segment is given by the magnitude of the muscle force multiplied by some quantity which has the units of distance. This quantity may arbitrarily be given the name "moment arm" of a muscle force. In fact, it can be shown that the quantity referred to is equal to the perpendicular distance from the line of action of the muscle force to the instantaneous screw axis about which the body is purely rotating, multiplied by the sine of the angle between the screw axis and the line of action of the applied muscle force. The truth of this statement can be shown by considering the rate at which the applied muscle force does work to purely rotate the body about the instantaneous screw axis (Hunt, 1978).

In the case of the knee, there are actually two bodies which move relative to the femur during flexion-extension of the joint : the patella and the tibia. Therefore, two such instantaneous axes of rotation may be found : one which describes relative movements of the femur and patella, and the other which describes relative movements of the femur and tibia. Consequently, two corresponding moment arms may be calculated: one for the quadriceps tendon about the instantaneous axis of rotation of the patella relative to the femur, and the other for the patellar ligament about the instantaneous axis of rotation of the tibia relative to the femur. The first quantity, when multiplied by the magnitude of force in the quadriceps tendon, expresses the torque exerted by the quadriceps tendon which causes pure rotation of the patella relative to the femur. Furthermore, when this torque is multiplied by the angular velocity of the patella relative to the femur, the result gives the rate at which work is done by the quadriceps tendon to rotate the patella about its instantaneous screw axis relative to the femur. The second quantity, when multiplied by the magnitude of force in the patellar ligament, expresses the torque exerted by the patellar ligament which causes pure rotation of the tibia about the femur. When this torque is multiplied by the angular velocity of tibia relative to the femur, the result is the rate at which work is done by the patellar ligament to rotate the tibia about its instantaneous screw axis relative to the femur. Because the mass of the patella is so much smaller than that of the tibia, it may be argued that the rate of doing work to rotate the patella is negligible in comparison with the rate of doing work to rotate the tibia during flexion-extension movements of the knee. Thus, the moment arm of the knee-extensor mechanism may be represented by the perpendicular distance from the line of action of the patellar ligament force to the instantaneous axis of rotation of the tibia relative to the femur, multiplied by the sine of the angle between these two lines in space.

Let the direction of the force applied by the patellar ligament to the tibia be represented by the unit vector n_{pl}. In cadaveric experiments, the unit vector n_{pl} was calculated from measured positions of the markers mounted at the origin and insertion sites of the patellar ligament. Thus, the force applied by the patellar ligament can be represented as

$$f\$_{pl}=f\begin{pmatrix} n_{pl} \\ \text{-------} \\ r_{pl}\times n_{pl}\end{pmatrix} \tag{10}$$

where r_{pl} is the position vector of an arbitrarily chosen point on the patellar ligament relative to the origin of the reference frame fixed on the femur, and $\$_{pl}$ designates the screw which contains the direction of the patellar ligament force as well as the moment of this force about the origin of the femoral reference frame, and f is the magnitude of the patellar ligament force. The instantaneous work done to move the tibia can be found by taking the *reciprocal product* of two screws, $\phi\$$, given by Eq. (8), and $f\$_{pl}$, given by Eq. (10). Thus,

$$\begin{aligned} f\$_{pl}\circ\phi\$ &= f\phi\times\begin{pmatrix} k \\ \text{-------} \\ r\times k+\frac{s}{\phi}k\end{pmatrix}\circ\begin{pmatrix} n_{pl} \\ \text{-------} \\ r_{pl}\times n_{pl}\end{pmatrix} \\ &= f\phi[n_{pl}\times(r\times k)+k\times(r_{pl}\times n_{pl})+n_{pl}\times\frac{s}{\phi}k] \\ &= f\phi[k\bullet((r_{pl}-r)\times n_{pl})+n_{pl}\times\frac{s}{\phi}k] \end{aligned} \tag{11}$$

The first term within the brackets represents the perpendicular distance from the line of action of the patellar ligament force, n_{pl}, to the finite axis of the tibia relative to the femur, k, multiplied by the sine of the angle between these two lines. This term, when multiplied by $f\phi$, gives the work done by the patellar ligament force to *rotate* the tibia from one position to the next about its finite screw axis. The second term within the brackets represents the distance translated by the tibia along its screw axis. This term, when multiplied by $f\phi$, gives the work done by the patellar ligament force to *translate* the tibia along its screw axis. The *moment arm* of the knee-extensor mechanism is therefore given by :

$$\begin{aligned} ma &= k\bullet((r_{pl}-r)\times n_{pl}) \\ &= d\sin(\alpha) \end{aligned} \tag{12}$$

where d is the perpendicular distance from the screw axis to the line of action of the patellar ligament force, and a is the "twist angle", or the angle between the line of action of the patellar ligament force and line representing the direction of the finite screw axis of the tibia relative to the femur (see Fig. 5). Note that since we calculated the location and direction of the finite screw axis for two successive positions of the tibia relative to the femur, the moment arm of the knee-extensor mechanism may be found using either of the two successive positions of the tibia. This is because the "moment arm" of a muscle force is an instantaneous property of the system. We found that the difference in the value of the knee-extensor moment arm calculated using either of the two positions of the tibia was negligible.

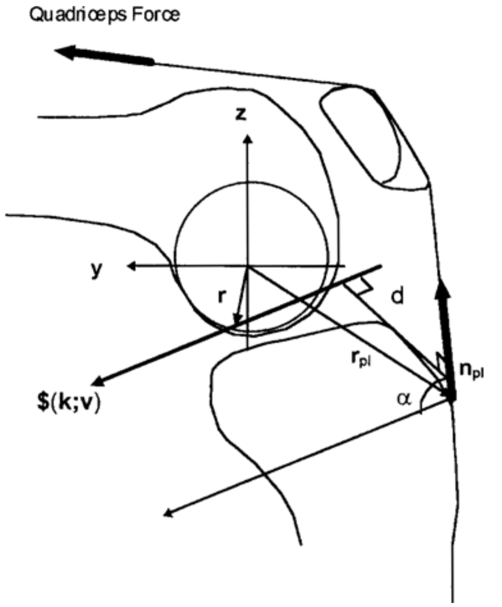


Fig. 5 The moment arm of the knee extensor is defined as the perpendicular distance between the line vector of the patellar ligament n_{pl} and screw axis k. Here, r_{pl} is the position vector of an arbitrarily chosen point on the patellar ligament with respect to the origin of the reference frame fixed on the femur, and r is the position vector which represents the perpendicular distance from the origin of the femoral reference frame to the finite screw axis.

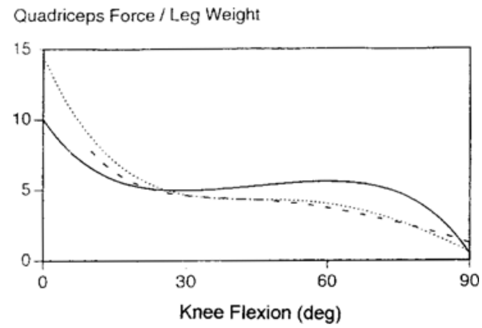


Fig. 6 Quadriceps force needed to extend an intact cadaveric knee during the leg raise task. The lower leg of the cadaver (shank and foot) was left intact. Experimental data are shown for three specimens: specimen 1 (solid line), specimen 2 (dashed line), and specimen 3 (dotted line).

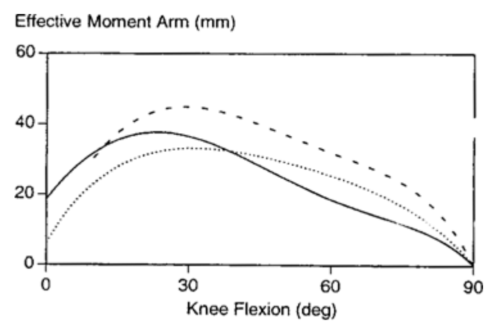


Fig. 7 Effective moment arm of the knee extensor mechanism calculated for three specimens during the leg raise task. The experimental result was obtained using the values of applied quadriceps force shown in Fig. 6.

4. Results

4.1 Applied quadriceps force

Figure 6 shows the variation of the applied quadriceps force with knee flexion angle for three intact cadaveric knees, with the average rate of knee extension being 4.5 deg/sec for each specimen. In general, quadriceps force increases as the knee extends from 90° flexion to full extension. This variation can be explained by considering a simple model of the knee in the sagittal plane (Fig. 8). The following analysis is similar to that given by Grood et al. (1984). For static equilibrium of the tibia, summing torques about the tibiofemoral joint gives

$$F_q = \frac{W L_m + W_a L}{d\eta} \cos\theta \qquad (13)$$

where W is the weight of the leg, W_a is the ankle weight added to the leg, L_m is the distance from the center of mass of the leg to the tibiofemoral contact point, d is the moment arm of the patellar ligament force about the tibiofemoral joint, η is the mechanical advantage of the patella, F_{pl} is the force in the patellar ligament, and θ is the angle of knee flexion. Since W, W_a, L_m, and L are all constants, Eq. (13) shows that the variation in quadriceps force depends on the variation in the moment arm of the patellar ligament force about the tibiofemoral joint (d), on the angle of knee flexion (θ), and on the mechanical advantage of the patella (η). Grood et al. (1984) called the quantity given by ($d\eta$) the *effective moment arm*

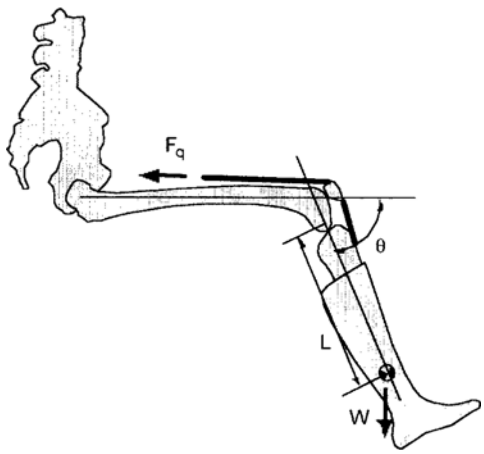


Fig. 8 Schematic diagram illustrating the mechanics of the quadriceps leg raise task (adapted from Grood et al., 1984). Quadriceps force (F_q) is applied to lift the lower leg under the action of gravity alone. θ represents the angle of knee flexion. W is the weight of the lower leg and L is the distance from the knee joint axis to the center of the mass of the lower leg.

at the knee.

The cosine of the knee flexion angle increases most rapidly near 90° and varies little near full extension. Thus, the initial increase of the quadriceps force results primarily from an increase in cos θ. In the range of 20-60°, the ratio of cos θ to the effective moment arm of the knee varies little. Consequently, the quadriceps force needed to extend the leg in this region remains relatively constant (Fig. 6, solid, dashed, and dotted lines between 20-60° of flexion). As the knee approaches full extension, the quadriceps force increases rapidly. The increase in quadriceps force near full extension may be due to the fact that the posterior capsule begins to tighten in this region (O'Connor et al., 1990). It may also be caused by contact of the ACL with the roof of the intercondylar notch as the knee approaches full extension, which presumably causes the ACL to tighten, thereby requiring much higher quadriceps force (Norwood and Cross, 1977).

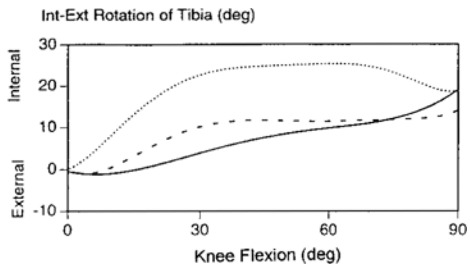
4.2 Relative displacements of the tibia

Kurosawa et al. (1985), Reuben et al. (1989) and Rovick et al. (1991) all measured the three-dimensional displacements of the femur relative to the tibia in vitro during a simulated squat-to-stand activity. They all reported that the most prominent motions were internal rotation of the femur and posterior displacement of the femoral origin as the knee flexed from full extension to 120°. In these experiments, the femur rotated as much as 20° internally relative to its position at full extension as the knee moved from 0° to 90°. Furthermore, the lateral condyle of the femur translated more than 15 mm as the knee flexed from full extension to 90°. Unfortunately, quantitative comparisons between these results and those obtained in the present study are not possible since the pattern of quadriceps force applied in each set of experiments is different. However, qualitative comparisons may be made.

For the loading conditions given by Fig. 6, Fig. 9 describes the changes in the position and orientation of the tibia relative to its position and orientation at full extension of the knee (the reference configuration). The most prominent motion of the bones was internal rotation and anterior displacement of the tibia relative to the femur as the knee flexed from full extension to 90° (Fig. 9(a) and (e), compare solid, dashed, and dotted lines).

The tibia rotated internally by as much as 25° in one of the specimens tested. In all specimens, the peak value of internal tibial rotation exceeded 15°. Furthermore, in each specimen the tibia only rotated internally as the knee was flexed from full extension (Fig. 9(a), compare solid, dashed, and dotted lines). These results indicate that this particular motion of the tibia was consistent and repeatable among the knees tested.

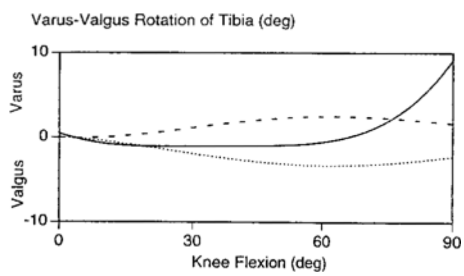
All specimens exhibited anterior translation of the tibial origin relative to the femoral origin as the knee flexed from full extension to 90°. The peak value of anterior translation of the tibial origin relative to the femoral origin in one speci men reached 15 mm as the knee was flexed to 90° (Fig. 9(e)). In each of the other two specimens, anterior translation of the tibial origin remained below 10 mm. In one specimen, the tibia actually translated posteriorly as the knee began to flex from full extension. With increasing knee flexion, however, the tibia then was translated anteriorly. These results are consistent with those reported by Kurosawa et al. (1985), who showed that the



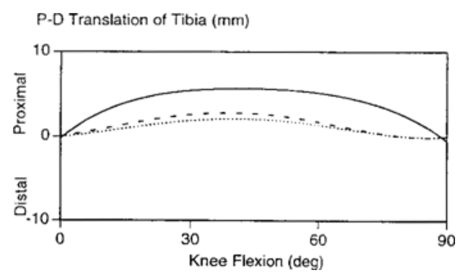
(a) Internal-external rotation of the tibia : internal rotation (positive) ; external rotation (negative)

lateral condyle of the femur translates posteriorly as the knee is flexed from full extension. In this case, posterior translation of the femoral origin relative to the tibial origin is equivalent to anterior translation of the tibial origin relative to the femoral origin. This phenomenon is characteristic of femoral roll-back at the knee (Garg and Walker, 1990).

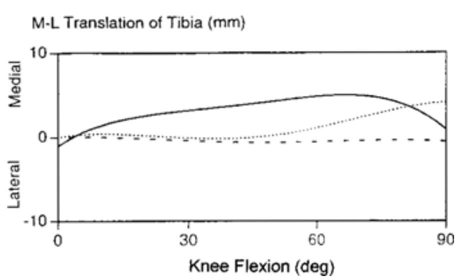
The remaining rotations and translations of the tibia are generally small in comparison with the internal-external rotations and anterior-posterior translations of this bone during a leg raise task. These less prominent motions of the tibia are also more variable, a finding which agrees with the observations reported by Rovick et al. (1991).



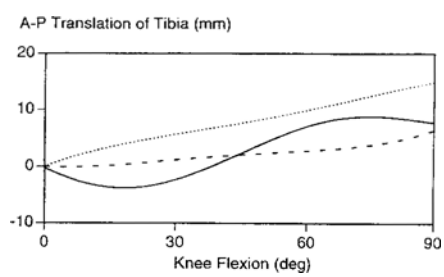
(b) Varus-valgus rotation of the tibia : varus of the tibia (positive) ; valgus of the tibia (negative)



(d) Proximal-distal translations of the tibia : distal translation (positive) ; proximal translation (negative)



(c) Medial-lateral translations of the tibia : medial translation (positive) ; lateral translation (negative)



(e) Anterior-posterior translations of the tibia : anterior translation (positive), posterior translation (negative)

Fig. 9 Displacements of the tibia measured relative to the femur for the quadriceps leg raise task. The displacements of the tibia were measured relative to its position and orientation when the knee was placed in full extension (i. e., the reference configuration of the leg). Translations of the tibia were found by calculating the displacement of the origin of the tibial reference frame with respect to the origin of the femoral reference frame, and were then expressed in the tibial reference frame. Rotations of the tibia were found by calculating the orientation of the tibial reference frame relative to the orientation of the femoral reference frame, and were then expressed in the tibial reference frame. Shown are the measured rotations and translations of the tibia relative to the femur for the three specimens tested in this study : specimen 1 (solid line), specimen 2 (dashed line), and specimen 3 (dotted line).

Varus-valgus rotations of the tibia remain less than 10° throughout the range of knee flexion (Fig. 9(b), solid, dashed, and dotted lines). Furthermore, medial-lateral and proximal-distal translations of the tibia were no more than 5 mm for all of the specimens tested (Figs. 9(c) and (d), solid, dashed, and dotted lines). For two of the specimen knees, proximal-distal translations of the tibial origin remained almost zero throughout the range of knee flexion (Fig. 9(d), solid, dashed, and dotted lines).

4.3 Relative displacements of the patella

As knee flexion increases, the patella shifts *laterally*, usually after a slight shift toward the medial side of the knee (Nagamine et al., 1995 ; van Kampen et al., 1987). Peak values of patellar shift as measured by van Kampen et al. (1987) ranged from 5 mm to as much as 15 mm. Results given by Nagamine et al. (1995), however, appear to be much less variable, with patellar shift remaining below 8 mm for all 11 specimens tested in that study.

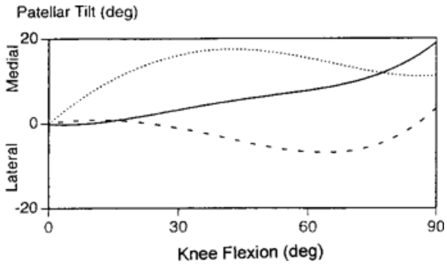
Figure 10 shows the changes in the position and orientation of the patella relative to its position and orientation with the knee at full extension (i. e., the reference configuration). The most consistent and reproducible motions of the patella are flexion as well as anterior and distal translations of the patellar origin relative to the origin of the femoral reference frame (solid, dashed, and dotted lines in Figs. 10(c), (e), and (f)). These results suggest, in agreement with findings reported by others, that the motion of the patella is relatively stable throughout the range of knee flexion. Nagamine et al. (1995) reported that patellar tracking is relatively independent of the amount and direction of the applied quadriceps force. This is because the anatomic configuration of the patellofemoral joint allows the patella to sit firmly inside the femoral groove at all angles of flexion, except as the knee approaches full extension.

Consistent with results reported by van Kampen et al. (1987) and Nagamine et al. (1995), patellar flexion *lags* knee flexion throughout the range of joint motion (Fig. 10(c), solid, dashed, and dotted lines). At 90° of knee flexion, peak values of patellar flexion range from 50-60°. Corresponding values reported by van Kampen et al. (1987) approached 70°. For two of the specimens tested in this study, the patella shifts *medially* as the knee flexes toward 90°. This motion was much smaller and more variable in the third specimen, however (Fig. 10(d), compare solid, dashed, and dotted lines). Peak values of patellar shift remain below 10 mm, which compares well with the data reported by Nagamine et al. (1995).

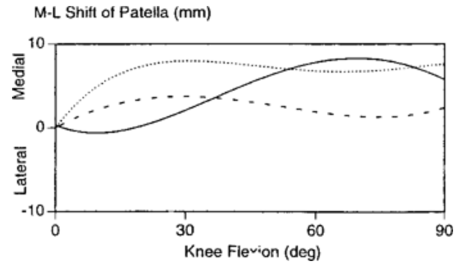
There is considerable variability in the measurements of patellar tilt for the three knees tested here (Fig. 10(a), compare solid, dashed, and dotted lines). In two of the specimens, the patella tilts medially as the knee flexes from full extension to 90°. In the third specimen, the patella tilts laterally relative to its position at full extension of the knee. The peak magnitude of this rotation is also greater than that reported by van Kampen et al. (1987) and Nagamine et al. (1995). Patellar tilt approaches 20° in two of the specimens tested in this study. The literature data show that patellar tilt remains below 10°.

These is also some discrepancy between measurements of patellar rotation in this study and data reported by others. Patellar rotation generally remains small throughout the range of knee flexion (Fig. 10(b), solid, dashed, and dotted lines). The patellae of two specimens rotate medially as the knee is flexed to 90°. These results are consistent with the findings of van Kampen et al. (1987) and Nagamine et al. (1995). In the third specimen, however, the patella rotates laterally (Fig. 10(b)). For all of the knees tested, however, peak values of patellar rotation remain below 10°, which is consistent with data reported by van Kampen et al. (1987) and Nagamine et al. (1995).

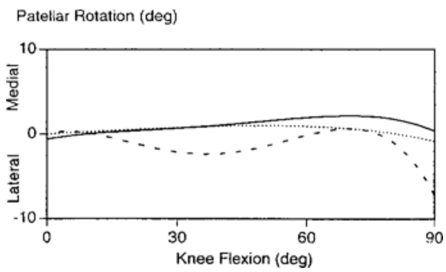
Differences between patellar tilt and patellar rotation obtained in this study and those reported by van Kampen et al. (1987) and Nagamine et al. (1995) are explained by the fact that the quadriceps forces applied to the cadaveric knees were different. Nagamine et al. (1995) and van Kampen et al. (1987) measured the tracking



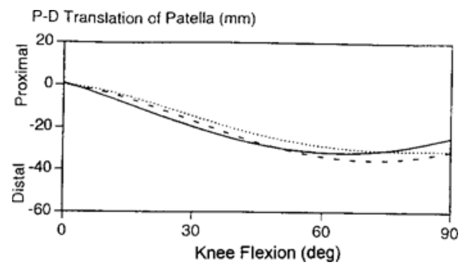
(a) Patellar tilt : medial tilt (positive) ; lateral tilt (negative)



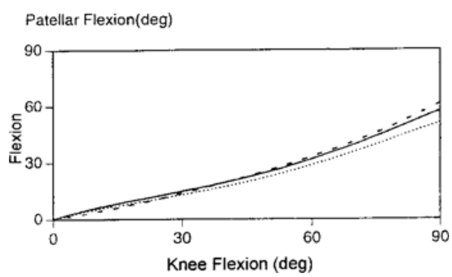
(d) Medial-lateral shift of the patella : medial shift (positive) ; lateral shift (negative)



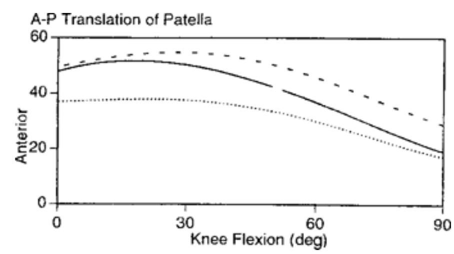
(b) Patellar rotation : medial rotation (positive) ; lateral rotation (negative)



(e) Proximal-distal translation of the patella : distal translation (positive) ; proximal translation (negative)



(c) Patellar flexion-extension : flexion (positive) ; extension (negative)



(f) Anterior-posterior translation of the patella : anterior translation (positive) ; posterior translation (negative)

Fig. 10 The corresponding displacements of the patella measured relative to the femur for the quadriceps leg raise task. The displacements of the patella were measured relative to its position and orientation when the knee was placed in full extension (i. e., the reference configuration of the leg). Translations of the patella were found by calculating the displacement of the origin of the patellar reference frame with respect to the origin of the femoral reference frame, and were then expressed in the femoral reference frame. Patellar rotations were found by calculating the orientation of the patellar reference frame relative to the orientation of the femoral reference frame, and were then expressed in the femoral reference frame. Shown are the measured rotations and translations of the patella relative to the femur for the three specimens tested in this study: specimen 1 (solid line), specimen 2 (dashed line), and specimen 3 (dotted line).

patterns of the patella during *passive* flexion-extension of the knee. In these experiments, the applied quadriceps force was relatively small (less than 100 N). Although Nagamine et al. (1995) found that the neither the magnitude nor the direction of the quadriceps force affected the movements of the patella, one would expect that changes in the applied quadriceps force would have a greater effect on patellar rotation and patellar tilt which are more highly variable between the specimens. This is because patellar rotation and patellar tilt are highly influenced by the direction of force applied to the patella by the patellar ligament, which in turn depends upon quadriceps force.

4.4 Moment arm of the knee-extensor mechanism

The moment arm of the knee-extensor mechanism when plotted against the angle of knee flexion exhibits a characteristic bell shape. For the three specimens tested in this study, the moment arm peaks somewhere in the 40-60° region of flexion. It then decreases rather rapidly on either side of this peak (Fig. 11, solid, dashed, and dotted lines). It is readily apparent that there is considerable variation in the moment arms for the three specimens used in this study. Figure 11 shows that the moment arms for two of the specimens differ by more than 20 mm at one particular angle of the knee (compare solid and dotted curves at around 40° of flexion). These results suggest that anatomical differences play a large role in determining the moment arm of the knee-extensor mechanism.

Both the shape and the magnitude of the moment arm of the knee-extensor mechanism is dominated by the perpendicular distance between the screw axis and the line of action of the patellar ligament force (compare solid, dashed, and dotted lines in Fig. 11 and Fig. 12). The angle between these two lines varies from about 40° near full knee extension to about 85° at large angles of flexion (Fig. 13, compare solid, dashed, and dotted lines). When the sine of this angle is evaluated, however, the result varies from 0.65 to a value that is very close to 1.0. This means that for most of the joint range of motion, the sine of the angle between the screw axis and the line of action of the patellar ligament force does not sigmticatly influence the magnitude of the knee-extensor moment arm. This conclusion is supported by the results of Figs. 11 and 12, which show that the moment arm of the knee-extensor mechanism and the perpendicular distance between the screw axis and line of action of the patellar ligament force are nearly equal at all angles of knee flexion.

The fact that the angle between the screw axis and the line of action of the patellar ligament force remains, for the most part, above 60° is consistent with the view that the major movements of the tibia relative to the femur occur in the sagittal plane. If the motion of the tibia rela-

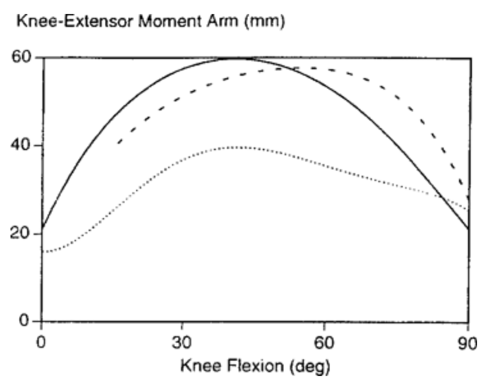


Fig. 11 Moment arm of the knee-extensor mechanism calculated using the measured relative displacement of the bones. Shown are the experimental data obtained from three specimens: specimen 1 (solid line), specimen 2 (dashed line), and specimen 3 (dotted line). Notice that the knee-extensor moment arms for the three specimens tested differ by as much as 2 cm.

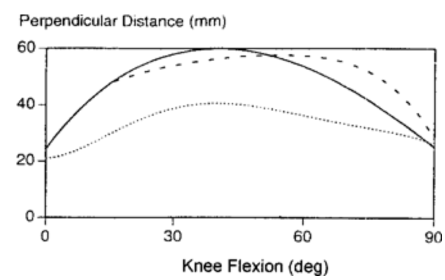


Fig. 12 The perpendicular distance between the line of action of the patellar ligament force and the line representing the direction of the screw axis of the tibia relative to the femur for the three specimen knees: specimen 1 (solid line), specimen 2 (dashed line), and specimen 3 (dotted line). For each specimen, the perpendicular distance between the patellar ligament force and the screw axis is approximately equal to the magnitude of the knee-extensor moment arm shown in Fig. 5.

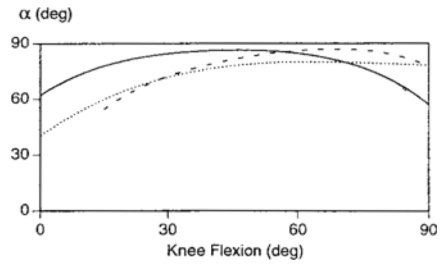


Fig. 13 The twist angle (α), defined as the angle between the line of action of the patellar ligament force and the direction of the screw axis of the tibia relative to the femur, for the three cadaveric knees : specimen 1 (solid line), specimen 2 (dashed line), and specimen 3 (dotted line). Sine of the twist angle (α) is close to unity at nearly all angles of knee flexion during the quadriceps leg raise task. This explains why the perpendicular distance given in Fig. 12 is nearly equal to the magnitude of the knee-extensor moment arm shown in Fig. 5 for each specimen.

tive to the femur remained in the sagittal plane, the angle between the screw axis and the line of action of the patellar ligament force would always be equal to 90°. This is because the screw axis of the tibia relative to the femur would then always be directed perpendicular to the plane defined by the line of action of the patellar ligament force and the long axis of the tibia. Therefore, only in this special case would the moment arm of the knee-extensor mechanism always be equal to the perpendicular distance between the screw axis and the line of action of the applied force, since the sine of the angle between these two lines remains equal to one.

5. Discussion

The fact that the angle between the screw axis and the line of action of the patellar ligament force is not always 90° (i. e., it varies anywhere from 60-85° in the three specimens tested) confirms what is already well-known about the real knee. It shows that the tibia undergoes movements other than flexion-extension, most noticeably internal-external rotation about its long axis. Movements of the tibia relative to the femur

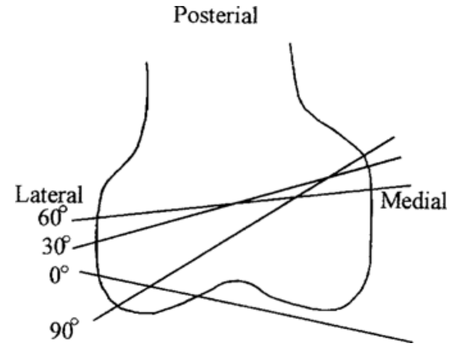


Fig. 14 Frontal plane view of the screw axes during the leg raise task for specimen 1. The screw axes are represented relative to the femoral coordinate system as a projection on the frontal plane.

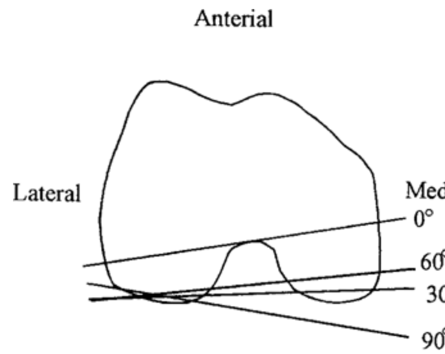


Fig. 15 Transverse plane view of the screw axes during the leg raise task for specimen 1. The screw axes are represented relative to the femoral coordinate system as a projection on the transverse plane.

that are not contained in the sagittal plane reflect a change in the *direction* of the screw axis (Figs. 14 and Fig. 15). Specifically, they indicate that the screw axis does not remain perpendicular to the plane containing the line of action of the patellar ligament force and the long axis of the tibia. If the angle between these two lines approached zero, the patellar ligament force would not be able to produce rotation of the tibia relative to the femur ; any force applied by the patellar ligament would then merely translate the tibia along its screw axis. In this sense, the value of the sine of the angle between the screw axis and the patellar ligament force represents the *efficiency* with which the patellar ligament force is able to exert torque on the tibia and rotate this bone about its screw axis.

To understand the variation with knee flexion angle of the moment arm of the knee-extensor mechanism, it is necessary first to explain the variation with knee flexion angle of the perpendicular distance between the screw axis of the tibia relative to the femur and the line of action of the patellar ligament force. The perpendicular distance between the screw axis of the knee and the line of action of the patellar ligament force is a function of the *location* within the knee of the screw axis of the tibia relative to the femur. The location of the screw axis is in turn determined by the interaction between two properties : (i) the geometry of the articulating surfaces of the femur, tibia, and patella at the knee, and (ii) the magnitude and direction of the forces applied by the muscles spanning the knee. Quantifying the interaction between these properties for a specific activity will enable an understanding of the variation with knee flexion angle of the moment arm of the knee-extensor mechanism.

In conclusion, the three dimensional displacements of the tibia and patella relative to the femur were measured in three cadaveric knees during a quadriceps leg raise task. The result has been used to estimate the location of the finite screw axis. The use of the screw axis appears to be a useful and accurate approach for calculating the moment arm. The procedure described in this paper also provides a promising method for measuring the total relative motion between two body segments without making limiting assumptions concerning the geometry of the actual joint.

Acknowledgments

The author would like to express his appreciation to Dr. Marcus Pandy of The University of Texas at Austin and to the Biomechanics Laboratory of The University of Texas Medical Branch at Galveston for their assistance in this study.

References

Ahmed, A. M. and Burke, D. L., 1983, "In-vitro Measurement of Static Pressure Distribution in Synovial Joints-Part Ⅰ : Tibial Surface of the Knee," *J. Biomech. Engng.*, Vol. 105, pp 216～225.

Ahmed, A. M., Burke, D. L., and Yu, A., 1983, "In-vitro Measurement of Static Pressure Distribution in Synovial Joints-Part Ⅱ : Retropatellar Surface," *J. Biomech. Engng.*, Vol. 105, pp. 226～236.

Ahmed, A. M., Burke, D. L., and Hyder, A., 1986, "Effect of Tibial Prerotation on the Ligamentous Response of the Flexed Knee to Passive Anterior Shear," *Trans. Ortho. Res. Soc.*, Vol. 11, pp. 127

Ahmed, A. M., Hyder, A., Burke, D. L., and Chan, K. H., 1987, "In-vitro Ligament Tension Pattern in the Flexed Knee in Passive Loading," *J. Orthop. Res.*, Vol. 5, pp. 217～230.

Blankevoort, L., Huikes, R., and de Lange, A., 1988, "The Envelop of Passive Knee Joint Motion," *J. of Biomechanics*, Vol. 24, pp. 1019～1031.

Craig, J. J., 1986, "Introduction to Robotics : Mechanics & Control," Addison-Wesley

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.*, Vol. 64-A, pp. 258～264

Garg, A. and Walker, P. S., 1990, "Prediction of Total Knee Motion Using a Three-Dimensional Computer-Graphic Model," *J. Biomechanics*, Vol. 23, pp. 45～50

Grood, E. S., Suntay, W. J., Noyes, F. R., and Butler, D. L., 1984, "Biomechanics of the Knee-Extension Exercise : Effect of Cutting the Anterior Cruciate Ligament," *J. Bone Jt. Surg.*, Vol. 66-A, pp. 725～734.

Grood, E. S., Stowers, S. F., and Noyes, F. R., 1988, "Limits of Movement in the Human Knee," *J. Bone Jt. Surg.*, Vol. 70-A, pp. 88～97.

Heegard, J., Leyvraz, P. F., Curnier, A., Rakotomanana, L. and Huiskes, R., 1995, "The Biomechanics of the Human Patella During Passive Knee Flexion," *J. Biomechanics*, Vol. 28, pp. 1265～1279.

Heegard, J., Leyvraz, P. F., van Kampen, A., Rakotomanana, L., Rubin, P. J., and Blankevoort, L., 1994, "Influence of Soft Structures on Patellar Three-Dimensional Tracking," *Clinical orthopaedics and related research*, Vol. 299, pp. 235 ~243

Hefzy, M., Jackson, W. T., Saddemi, S. R., and Hsieh, Y. F., 1992, "Effects of Tibial Rotations on Patellar Tracking and Patello-Femoral Contact Areas," *J. Biomed. Eng.*, Vol. 14, pp. 329 ~343.

Hirokawa, S., Solomonow, M., Lu, Y., Lou, Z., and DAmbrosia 1992, "Anterior-Posterior and Rotational Displacement of the Tibia Elicited by Quadriceps Contraction," *Am. J. Sports Med.*, Vol. 20, pp. 299~306.

Hirokawa, S., 1991, "Three-Dimensional Mathematical Model Analysis of the Patellofemoral Joint," *J. Biomechanics*, Vol. 24, pp. 659~671.

Hunt, K. H., 1978, "Kinematic Geometry of Mechanisms," Clarendon Press, Oxford, England

Kim, S., 1996, "A Three-Dimensional Dynamic Musculoskeletal Model of the Human Knee Joint," Ph. D. Dissertation, Mechanical Engineering Dept. The University of Texas at Austin, TX, U. S. A

Kinzel, G. L., Hall, A. S. Jr., and Hillberry, B. M., 1972, "Measurement of the Totaal Motion Between Two Body Segments- Ⅰ. Analytical Development," *J. Biomechanics*, Vol. 5, pp. 93 ~105.

Kinzel, G. L., Hillberry, B. M., and Hall, A. S. Jr., 1972, "Measurement of the Totaal Motion Between Two Body Segments- Ⅱ. Description of Application," *J. Biomechanics*, Vol. 5, pp. 283 ~293.

Koh, T. J., Grabiner, M. D., and De Swart, R. J., 1995, "In Vivo Tracking of the Human Patella.," *J. Biomechanics*, Vol. 25, pp. 637~643.

Kurosawa, H., Fukubayashi, T., and Nakajima, H., 1980, "Load-Bearing mode of the Knee Joint : Physical Behavior of the Knee Joint with and Without the Menisci," *Clin. Orthop. Rel. Res*. Vol. 149, pp. 283~290.

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*, Vol. 18, pp. 487~499.

LaFortune, M. A., Cavanagh, P. R., Sommer, H. J. Ⅲ, Kalenak, A., 1992, "Three-Dimensional Kinematics of the Human Knee During Walking," *J. Biomechanics*, Vol. 25, pp. 347~357.

Lewis, J. L., Lew, W. D., and Schmidt, J., 1988, "Description and Error Evaluation of an in Vitro Knee Joint Testing System," *ASME J. Biomech. Engng*. Vol. 110 pp. 238~248

Nagamine, R., Otani, T., White, S. E., McCarthy, D. S., and Whiteside, L. A., 1995, "Patellar Tracking Measurement in the Normal Knee," *J. Orthopaedic Research*, Vol. 13, pp. 115~122

Nisell, R, 1985, "Mechanics of the Knee," *Acta Orthop. Scand.*, Supplementum 216, Vol. 56, 1 ~42.

Norwood, L. A. and Cross, M. J., 1977, "The Intercondylar Shelf and the Anterior Cruciate Ligament," *Am. J. Sports Med.*, Vol. 5, pp. 171 ~176

Reuben, J. D., Rovick, J. S., Schrager, R. J., Walker, P. S., and Boland, A. L., 1989, "Three-Dimensional Dynamic Motion Analysis of the Anterior Cruciate Ligament Deficient Knee Joint," *Amer. J. Sports Med.*, Vol. 17, pp. 463 ~471.

Rovick, J. S., Reuben, J. D., Schrager, R. J., and Walker, P. S., 1991, "Relation Between Knee Motion and Ligament Length Patterns," *Clinical Biomechanics.*, Vol. 6, pp. 213~220.

Smidt G. L., 1973, "Biomechanical Analysis of Knee Flexion and Extension," *J. Biomechanics*, Vol. 6, pp. 79~92.

Sommer Ⅲ, H. J., 1992, "Determination of First and Second Order Instant Screw Parameters from Landmark Trajectories," ASME *J. of Mechanical Design*, Vol. 114, pp. 274~281.

Spoor, C. W. and Veldpaus, F. E., 1980, "Rigid Body Motion Calculated from Spatiaal Coordinates of Markers," *J. Biomechanics*, Vol. 13, pp. 391~393.

Spoor, C. and van Leeuwen, J., 1992, "Technical Note : Knee Muscle Moment Arms from MRI and from Tendon Travel," *J. Biomechanics*, Vol. 25, pp. 201~206.

Suh, C. H. and Radcliffe, C. W., 1978, "Kinematics and Mechanisms Design," John

Wiley and Sons.

van Dijk, R., Huiskes, R., and Selvik, G., 1979, "Roentgen Stereophotogrametric Methods for the Evaluation of the Three-Dimensional Kinematic Behaviour and Cruciate Ligament Length Patterns of the Human Knee Joint," *J. Biomechanics*, Vol. 12, pp. 727～731

van Kampen, A. and Huiskes, R., 1990, "The Three-Dimensional Tracking Pattern of the Human Patella," *J Orthopaedic Research*, Vol. 8, pp. 372～382.

van Kampen, A., 1987, "The Three-Dimensional Tracking Pattern of the Patella," Ph. D. thesis, University of Nijmegen. The Netherlands.

Veress, S. A., Lippert, F. G., Hou, M. C. Y., and Takamoto, T., 1979, "Patellar Tracking Patterns Measurement by Analytical X-Ray Photogrammetry," *J. Biomechanics*, Vol. 12, pp. 639～650.

Wu, G. and Cavanagh, P. R., 1995, "ISB Recommendations for Standardization in the Reporting of Kinematic Data," *J. Biomechanics*, Vol. 28, pp. 1257～1261.