Acta Orthop Scand *57*, 41–46, 1986

Joint forces in extension of the knee

Analysis of a mechanical model

A two-dimensional model of the tibio-femoral joint was constructed by using the results of cadaver knee dissections together with radiographic landmarks on healthy knees loaded at various angles of flexion. The tibio-femoral compressive force during isometric knee extension had the same magnitude as the patellar tendon force. The tibio-femoral shear force changed direction from posterior at full flexion to anterior when the knee was extended, indicating that high forces may arise in the anterior cruciate ligament in the straight knee. Women developed some 20 per cent higher knee joint forces than men for the same extending muscular moment, since women's patellar tendon moment arms were found to be shorter. The model presented may be used for quantifying tibio-femoral forces during knee extending activities.

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Most earlier mechanical knee models have been limited to giving the tibio-femoral forces only for one particular situation or course (e.g. Lindahl & Movin 1967, Morrison 1968, Smidt 1973, Bishop 1977, Kelley et al. 1978, Ellis et al. 1979 and 1984, Wahrenberg et al. 1978, Lindbeck 1983, Ekholm et al. 1984). The aim of our study was to define a general mechanical model of the knee. It was developed from a morphological investigation of cadavers combined with a determination of radiographic landmarks in healthy individuals.

Material and methods

Cadaver knees of 10 men and 10 women (Table 1) without severe arthrosis were dissected. The patella was cut off from the extensor apparatus. The thickness of the patellar tendon was measured at its origin. The measured distances were used as landmarks on the radiographs.

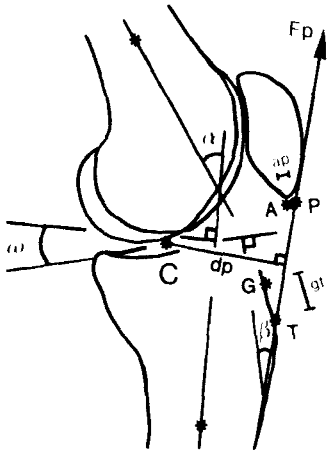
Table 1. Anthropometrical and morphological data from dissected cadavers. Distances in mm. Mean (SD)

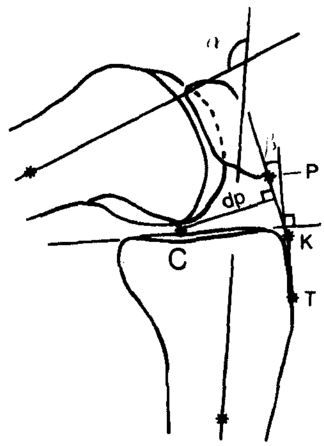
	Women n=10	Men n=10
Height (m)	1.62 (0.06)	1.74 (0.05)
Femoral epicondyle width	84.4 (3.6)	93.7 (4.7)
Patellar height	43.6 (3.1)	48.2 (2.8)
Patellar width	45.4 (3.9)	49.5 (4.0)
Distance ap (Figure 1)	2.7 (0.7)	3.1 (0.8)
Distance gt (Figure 1)	8.2 (1.8)	11.0 (2.2)

Twenty healthy subjects, 10 men (mean age 27 years, height 1.82 m, and weight 75 kg) and 10 women (mean age 23 years, height 1.67 m, and weight 59 kg), without knee pathology volunteered for the radiographic study. Lateral radiographic films (30×40 cm) were taken of the loaded knee with the subjects standing at different knee flexion angles (α, Figure 1). One anteroposterior and one lateral projection were obtained with the knee straight and the quadriceps activated. The film-focus distance was 155 cm and the knee centre-film distance was 15 cm, giving a magnification correction factor of 1.11.

A curve between the contours of the two femoral condyles was drawn, giving a "mean femoral condyle". This technique was used because the condyles are not completely congruent. The corresponding procedure was performed regarding the tibial plateau. The contact point (Figures 1 and 2) was defined as the midpoint at the shortest distance between the mean femoral and tibial condyles. The distance between the medial and lateral femur epicondyles (femoral epicondyle width) was measured on the radiograph and directly on the knee with callipers. The long axis of the femur was determined from the radiographs as the line between two shaft midpoints located 75 and 150 mm from the mean femoral condyle surface. The tibial long axis was similarly determined. The knee flexion angle was defined as the angle between the two bone axes.

The forces in the various parts of the patellar tendon were presumed to act as one single force vector from the patellar origin (point P) to the tibial insertion (T) (Figure 1). For 120° knee angle, where the tendon is curved, an extra point K (Figure 2) was chosen at a distance corresponding to half the patellar tendon thickness in front of the anterior border





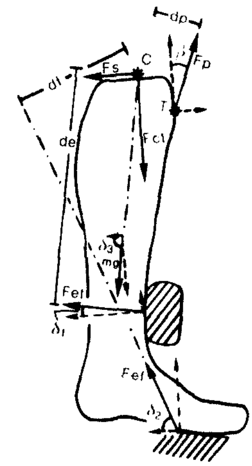


Figure 1. The anatomical landmarks of the morphological and radiographic studies.
The patellar tendon moment arm (dp) is the perpendicular distance between the patellar tendon force (Fp) and the centre of the tibio-femoral contact area (C). Fp acts through the proximal patellar tendon midpoint (P) with a distance "ap" from the patellar apex (A) and through the distal midpoint (T) with distance "gt" from the posterior tendon border (G). The tibial plateau slopes backwards (ω mean for women 7.2°, and for men 9.2°) in relation to the tibial long axis. α = knee angle. β = angle between Fp and a line perpendicular to the tibial plateau.

Figure 2. At 120° knee angle (α) the patellar tendon is curved. Point K was located where the tendon changed its direction. For explanation of other symbols, see Figure 1 and its legend.

Figure 3. Free body diagram of the lower leg.
The magnitude of the patellar tendon force (Fp) and the tibio-femoral shear (Fs) and compressive force (Fct) can be calculated (see equations in the text) if the external forces (Fet, Fef and mg), their angles in relation to the tibial plateau ($\delta 1$, $\delta 2$ and $\delta 3$) and their moment arms (de and df) have been determined.

of the tibial plateau. The direction of the patellar tendon force (Fp) was defined as the straight line between points K and P. The patellar tendon moment arm (dp) was the perpendicular distance between Fp and the tibio-femoral contact point (C).

Mechanical analysis

Point C was chosen as the origin for the calculation of moments. For the tibia in a free body diagram and in a static situation (Figure 3), the flexing knee load moment (Mf) equals the extending knee muscular moment (Me). These moments about C may be determined:

$$Mf = Me = (de \times Fet) + (df \times Fef) + (dm \times mg) \quad (1)$$

Fp is given by:

$$Fp = Mf / dp \quad (2)$$

From Figure 3, we get the vector equation:

$$\bar{F}p + \bar{F}ct + \bar{F}s + \bar{F}et + \bar{F}ef + \bar{m}g = 0 \quad (3)$$

Projecting (3) in the normal (or Fct) direction of the tibial plateau;

$$Fct = Fp \times \cos\beta + Fet \times \sin\delta 1 + Fef \times \sin\delta 2 - mg \times \sin\delta 3 \quad (4)$$

in tangential (or Fs) direction;

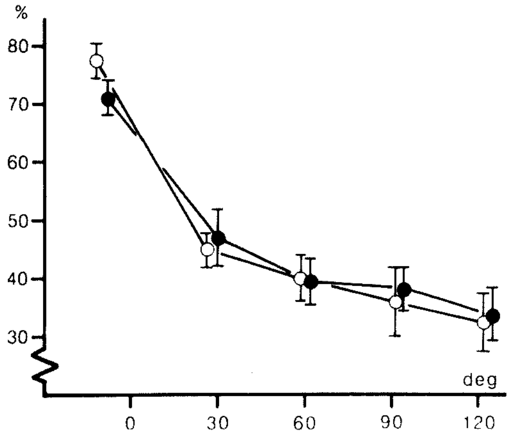
$$Fs = Fp \times \sin\beta - Fet \times \cos\delta 1 - Fef \times \cos\delta 2 - mg \times \cos\delta 3 \quad (5)$$

In the example in Figure 3, the external forces act on the anterior border of tibia and on the sole of the foot, but it is obvious that these forces may act anywhere on the "free body".

ANOVA 3-way variance-analysis and Student's unpaired *t*-test were used for the statistical analyses of the data. The significance levels chosen were consistently $p<0.01$.

Results

There was no difference between the dissected and *in vivo* groups regarding the measure-



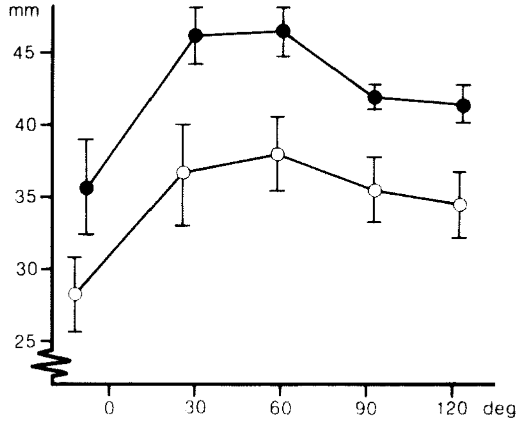
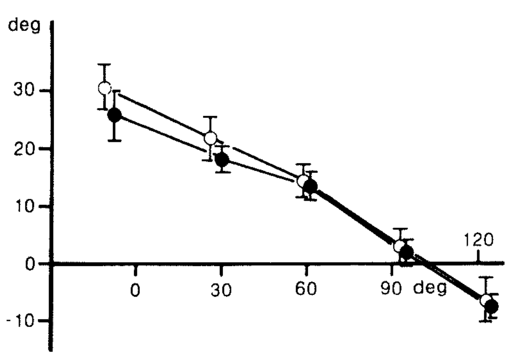


Figure 4. The location of the tibio-femoral contact point (C, y-axis) at various knee flexion angles (x-axis). The 95 per cent confidence intervals of the means are drawn for men (filled circles) and women (open circles). Y-axis shows the location in per cent of the sagittal tibial plateau length. 100 per cent = the anterior border of the plateau.

Figure 5. The patellar tendon moment arm (dp, y-axis) at various knee flexion angles (x-axis).
The 95 per cent confidence intervals of the means are drawn for men (filled circles) and women (open circles). Women have statistically significant shorter moment arms than men throughout the whole range of motion.

ments given in Table 1. Point C moved anteriorly on the plateau when the knee was extended (Figure 4). At the last 30° of knee extension point C was more displaced than above 30° knee flexion angle. The displacement of C during knee extension from 120° knee angle to straight was about 40 per cent of the tibial plateau sagittal length, and similar for both sexes. Regarding the localization of point C, there was a difference between the straight knee and all the other knee angles, but also a difference between 30° and 90 as well as 120°.

The maximum length of the patellar tendon moment arm was found at 30–60° knee flexion angle, and the shortest at the straight knee (Figure 5). The same pattern was seen for both sexes. The magnitude of angle β increased during the knee extension movement (Figure 6).



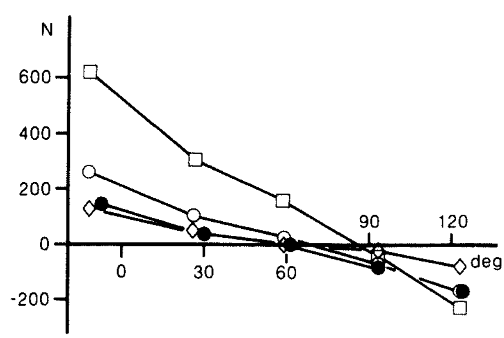


Figure 6. The magnitude of the angle between the patellar tendon and the perpendicular line of the tibial plateau (β, y-axis) at various knee flexion angles (x-axis). The 95 per cent confidence intervals of the means are shown for men (filled circles) and women (open circles). Note the intersection of the zero-line at about 100° knee angle.

Figure 7. Tibio-femoral shear force (Fs) during isometric knee extension at various knee angles (x-axis) for women (unfilled symbols) and for men (filled circle). Square = external force against tibia (Fet) of 100 N and its moment arm (de) of 0.4 m. Circle = Fet of 100 N and de of 0.2 m. Diamond = Fet of 50 N and de of 0.2 m. Positive values indicate that tibia tends to slide anteriorly in relation to femur.

β intersected the zero-line at about 100° knee angle.

The mechanical analysis and the equations presented here were put into practice. Consider an isometric knee extension against a resistance applied to the anterior side of the distal leg, with the subject lying on one side giving mg=0, the resistance acts with a force (Fet, Figure 3) perpendicular to the tibial long axis. The flexing load moment (Mf) about point C may be calculated if the magnitude of Fet and its moment arm (de) to point C have been determined. From equation (1), we get Mf = Fet × de. From equations (2), (4) and (5), we get Fp = Mf / dp, Fct = Fp × cosβ + Fet × sinδ1, and Fs = Fp × sinβ – Fet × cosδ1.

The maximum calculated Fp was found at the straight knee because of a shorter patellar tendon moment arm. Women had shorter moment arms than men, and therefore the calculated forces in women were consistently about 20 per cent higher. Fct did not change dramatically over the various knee angles and the magnitude was about the same as for Fp. When the knee extended with a moment of 40 Nm, with Fet = 100 N and de = 0.4 m, we found for women that Fp and Fct were about 1100 N at knee angles of 30–120°. For the straight knee these forces were higher, about 1230 N and 1400 N respectively. Fs changed sign between 50 and 90° flexion angle, depending on the magnitudes of Fet and de (Figure 7). Fs was around -200 N at 120° knee angle and as much as 600 N anteriorly shearing when the knee was straight.

Discussion

Our results concerning point C mainly agreed with those of earlier investigators (Zuppinger 1904, Lindahl & Movin 1967), but we found a tibio-femoral rolling motion also at flexion angles beyond 30°. Lindahl & Movin (1967) found a displacement of point C not exceeding 10 mm and Walker & Hajek (1972) a displacement of about 10–15 mm. The corresponding distance in our study was about 40 per cent of the sagittal tibial plateau length (Figure 4) or more than 20 mm, when the knee extended from 120° to straight. The sliding and rolling tibiofemoral motion is an important mechanical event that was not sufficiently considered in the early design of knee joint endoprostheses and seems to explain the unfavourable results of hinge prostheses that do not allow for this motion. Arthroplasty based on simulation of normal knee joint motion and biomechanics offers a natural chance for good results (Gunston 1971, Walker et al. 1982).

Using a permutation technique, Nissan (1980) showed that an accurate anterio-posterior location of point C is most important in knee biomechanical analysis. If point C were moved backward 10 mm in the present model, the calculated forces in the patellar tendon would decrease about 22 per cent. If point C were moved 10 mm anteriorly, this force would increase up to 40 per cent.

Smidt (1973) estimated the patellar and hamstring tendon forces by using the instant centre of motion as the moment point. In our study, the moment point chosen was the centre point of contact (C) between the condyles. The same biomechanical approach has been used by several authors (Lindahl & Movin 1967, Haffajee et al. 1972, Bishop 1977, Ellis et al. 1979). It is an advantage to choose point C as the moment point because the tibio-femoral compressive force acts through it and thus gives rise to a zero moment. The shear force is balanced by structures with short or zero moment arms to C (e.g. cruciates, menisci, collateral ligaments and joint capsule) and therefore causes a small or zero moment about this point. Consequently, the moments caused by the joint compressive and shear forces can be omitted. In Smidt's (1973) model these moments were not discussed, and this omission may give incorrect values of the force magnitudes acting in the knee.

The β angle magnitude is of great mechanical interest (Figure 6). The intersection of the zero-line at about 100° knee angle indicates that the patellar tendon pulls the tibia backwards in relation to the femur above this knee angle (negative β), and has the opposite effect at straighter knee angles (positive β). The magnitudes of β at various knee angles have not been reported earlier, as far as we know.

The patellar tendon moment arm estimated by various authors is illustrated in Figure 8.

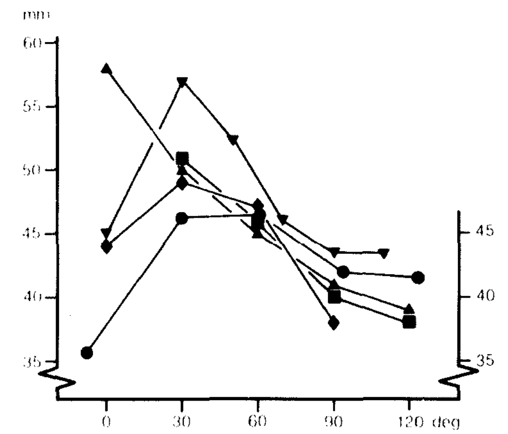


Figure 8. Patellar tendon moment arm estimated by others compared to our results (circle). Triangle apex up = Kaufer (1971). Triangle apex down = Bandi (1972). Square = Haxton (1945). Diamond = Smidt (1973).

The pattern of change is quite similar. The main difference between our results and others is seen at straighter knee angles, where we suggest the moment arm to be shorter than has earlier been reported. This, together with the physiological muscular length-tension relationship, may be one explanation why knee extending strength decreases during the last 30–40° of knee extension (Williams & Stutzman 1959, Lieb & Perry 1968, Lindahl & Movin 1968, Lindahl et al. 1969, Haffajee et al. 1972, Smidt 1973). It may also explain the difficulty in acheiving the last few degrees of knee extension in clinical practice.

Women had significantly shorter moment arms (dp) than men (Figure 5). Therefore, the forces arising in female knees are higher than in men's for the same knee moment. In walking, the knee load moment is generally proportional to body weight. Usually, women weigh less than men and their knee load moments will therefore be less. The contact area of the compressive forces is probably less for women in general due to their smaller knees. A woman's knee joints are exposed to about the same load moment as those of a man of the same weight, but her knee joint forces will be higher (smaller knees – shorter patellar tendon moment arms). The stresses on the cartilage occasioned by compressive forces and, probably, smaller contact areas will also be greater. The conclusion is that overweight women are exposed to higher knee joint stresses than men of corresponding weight. This may partly explain why gonarthrosis is more frequent in women than in men and why overweight women are especially vulnerable.

High anterior cruciate ligament tension may arise during knee rehabilitation (Brask et al. 1984). Lindahl & Movin (1967) and Paulos et al. (1981) found high anterior cruciate ligament forces during the last 30° of extension, with the maximum at the extended knee joint. At knee angles 0–45°, Arms et al. (1984) found that the strain increased considerably due to the quadriceps contraction. At knee flexion angles greater than 60°, they found a tension decrease. The theoretical results of our model agree well with these studies. In the postoperative rehabilitation of patients with a repaired anterior cruciate ligament, we can tell from our model that if the quadriceps muscle is isometrically exercised with the knee flexed to 60° or more, anterior tibial shear forces (Figure 7) will not arise if the external moment arm (de) is 20 cm or less. In this situation the anterior cruciate ligament will not be tensed. In the straight knee however, the anterior shear force may rise to around 200 N and 600 N (de=20 cm and de=40 cm respectively, Fet=100 N), causing high forces acting on the anterior cruciate ligament.

After patellectomy the patellar tendon moment arm becomes shorter and the action angle of the patellar tendon force changes, pulling the tibia more posteriorly in relation to the femur. A shorter moment arm decreases the knee extending strength proportionally to the length of the moment arm. Knowing how forces act in and around the knee joint makes it possible to estimate the mechanical effects of a patellectomy and other derangements of the knee.

Acknowledgements

This project was supported by the Karolinska Institute, the Swedish Medical Research Council (5720) and the Swedish Work Environment Fund (84/0520). We thank Lars Lindbeck for help with the biomechanical analysis.

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