Acta orthop. scand. 48, 511–516, 1977

LOAD BEARING CHARACTERISTICS OF THE PATELLO-FEMORAL JOINT

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This study of normal patello-femoral biomechanics defines some functional specifications which may be useful in future total knee prosthesis design. Serial lateral X-rays of 15 fresh knees and their patellar mechanisms at several flexion angles provided definition of the direction of the resolved patello-femoral forces. Assuming that the patella acts as a frictionless pulley, the magnitude of the patello-femoral forces during several activities was calculated using data from Morrison (1970) and Smidt (1973). It ranged between 421 and 3420 newtons for the various activities and for isometric exercise. A methylene blue contact print technique was used to measure the bearing areas. These data indicate that between 13 and 38 per cent of the patellar surface bears joint loadings. Patello-femoral contact stresses were calculated to range from 1.28 to 12.6 N/mm^2. A 696 newton man climbing stairs would, for example, generate a patello-femoral force of 1754 newtons and would experience patello-femoral contact stresses between 3.73 and 6.87 N/mm^2. Stress values were equal to or in excess of anticipated tibial-femoral stresses. The high patello-femoral load values, the small bearing surfaces, and the consequent significant stress magnitudes indicate the need for caution in development of a patello-femoral joint prosthetic replacement.

Key words: biomechanics; knee joint; knee joint forces; patella; prosthesis

Accepted 16.v.77

Most available total knee joint prostheses treat only the tibial-femoral portion of the knee joint. Residual pathology at the patello-femoral articulation may detract from the expected relief of pain after these procedures. Rational design considerations in the development of a prosthesis which addresses this problem require knowledge not only of the anatomy of the joint, but also of the functional stresses normally present across the bearing surfaces. Although considerable information is available regarding the anatomy, biomechanics, and pathology of the patello-femoral articulation, we were unable to find adequate published experimental values for patello-femoral contact areas and for contact stresses at

This paper was presented in part at the 1975 meeting of the Orthopaedic Research Society in San Francisco, California.

this articulation during normal activities. This study was initiated to define these quantities which may be useful in the development and pre-clinical testing of prostheses designed to treat this articulation.

MATERIALS AND METHODS

Quantification of patello-femoral contact stresses depends upon the force normal to the touching articular surfaces and upon the area of contact itself. Fifteen fresh grossly normal cadaver lower limbs were minimally dissected. Metallic markers were placed at the centers of the insertions of the patellar tendon on the tibia and patella, at the center of the insertion of the quadriceps mechanism on the superior pole of the patella, and at the center of the quadriceps musculotendinous junction, approximately 6 cm above the patella. The extensor mechanism was manually pre-tensioned and lateral roentgenograms made at 15-degree increments from full extension to full flexion.

The knee joint angle, α, was determined by drawing lines along the central axes of the tibia and femur (Figure 1 A). Lines connecting the two metallic markers in the patellar tendon and in the quadriceps aponeurosis, respectively, defined an angle, β (Figure 1 B), the patellar

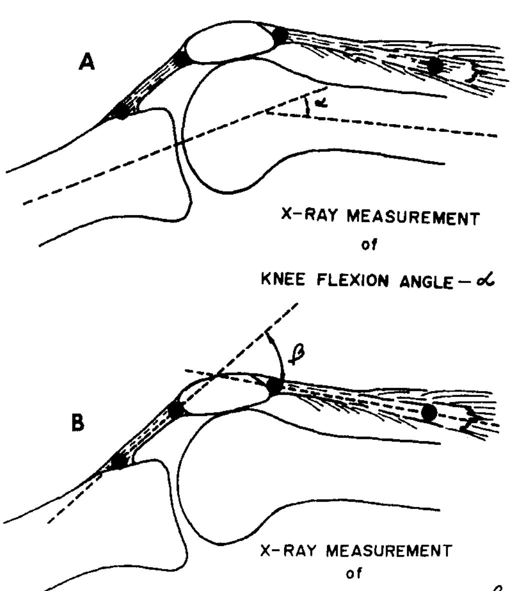

Figure 1. Radiographic method of determination of knee flexion angle, α, and patellar mechanism angle, β.

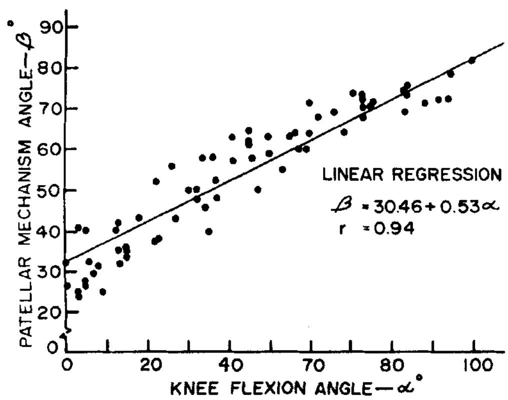

Figure 2. The relationship between the knee flexion angle and the patellar mechanism angle.

mechanism angle. The relationship between α and β (Figure 2) was plotted and subjected to exponential, power series, and linear regression analysis. The latter provided the best data fit and this relationship was used in the remainder of the study.

If quadriceps force, F_Q, is assumed to be sustained equally by the quadriceps aponeurosis, patella, and the patellar tendon; and if the patellar mechanism angle, β, is known, the normal patellar contact force, the force of the patella against the distal femur, F_{pf}, may be calculated to the equation, $F_{pf} = 2 F_Q \sin \beta/2$ (Figure 3).

Morrison (1970 a and b) has reported maximum quadriceps forces and associated knee flexion angles for five activities—level walking,

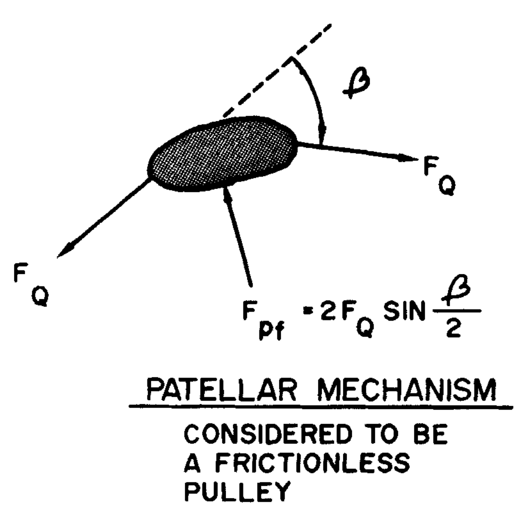

Figure 3. Method of calculation of patello-femoral force.

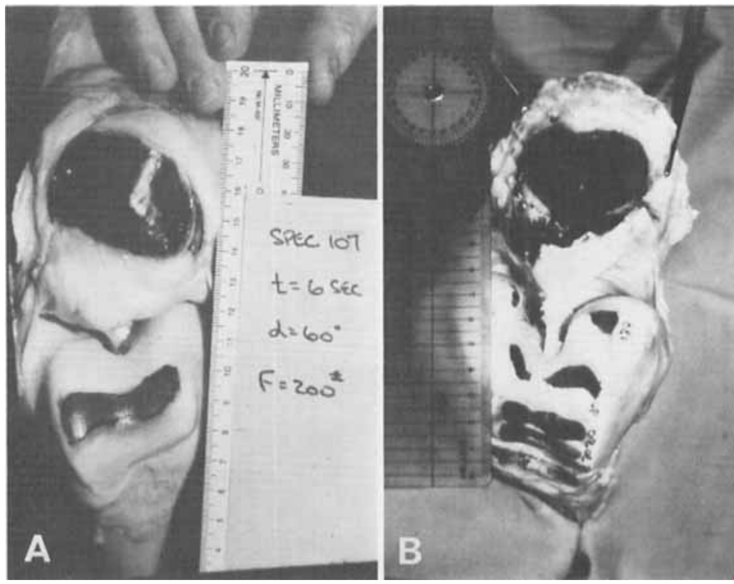

Figure 4. Typical photographs of specimens. A. A single contact area at 30° knee flexion. B. The number of contact area determinations that were possible for one specimen is shown.

walking up and down a one to six ramp, and going up and down stairs. Similar isometric quadriceps forces at defined knee flexion angles have been reported by Smidt (1973). These values were used to calculate the patello-femoral contact forces in Table 1.

In order to determine contact stress, direct measurement of the patello-femoral contact areas was necessary. For our 15 fresh specimens, the patella was surgically released from its medial and lateral proximal capsular attachments. It was painted with methylene blue or red instrument ink. A desired knee flexion angle was selected. The patella was returned to its normal position for the selected angle and secured in position with manual tension on the quadriceps aponeurosis. A force of 25 kg was then manually applied normal to the patello-femoral contact point and coaxial with the predetermined patello-femoral force vector, F_{pf}. The resulting contact print (Figure 4) of the force bearing area was photographed.

A lens to object distance of 18 inches was used for all photographs. A centimeter ruler was positioned level with the area of interest and included in all photographs. Multiple exposures were made of each specimen to be sure that the contact areas of interest would be perpendicular to the focal axis. The resulting slides were enlarged until the centimeter markings on the ruler could be superimposed on centimeter graph paper. The contact areas were traced and the areas enclosed by the tracings were determined by planimetry.

Although Aglietti et al. (1975) have reported that in a limited number of experiments on two specimens patello-femoral areas were only slightly affected by increases in quadriceps force, we were concerned that due to the deformability of articular cartilage, our manually applied patello-femoral load of 25 kg was nonphysiologic, and that our measured contact areas might be smaller than those experienced by the knee during normal activities. Accordingly, prepared specimens were fixtured in an Instron testing machine where loads between 118 and 1472 newtons were sequentially applied to the patella. These experiments demonstrated that a small but significant increase in measured areas results from the higher loads applied to the patella along the F_{pf} vector line. A similar series of experiments determined that the time period of loading caused no significant differences in measured areas over a 6 second to 2 minute span. Subsequent specimens were loaded to 893 newtons at various knee flexion angles for 6 seconds. Total patellar articular areas were determined by the same technique.

RESULTS

The average and range for total patello-femoral contact areas at all measured knee flexion angles are presented in Figure 5 A. The percentage of the average total patellar articular surface in contact with the femur is recorded. The influence

Table 1. Patello-femoral and tibial-femoral contact forces, areas, and stresses compared.

Quadriceps force data from	Activity	Knee joint angle (degrees)	Patellar mechanism angle (degrees)	Quadriceps force (Newtons)	Tibial-femoral contact force (Newtons)	Tibial-femoral contact areas (millimeters²)	Tibial-femoral contact stress (N/mm²)
Morrison	Walking	15	38	647	2109	660	3.2
(1970 a, 1970 b)	Up ramp	30	46	785	2786	560	5.0
	Down ramp	15	38	1923	2786	660	4.2
	Upstairs	45	54	1923	2963	560	5.3
	Downstairs	60	62	1687	2668	560	4.8
Smidt (1973)	Isometric	5	33	1393	1294	–	–
	Quadriceps	15	38	1776	1648	660	1.9
	Contracture	30	46	2168	2040	560	3.6
		45	54	2433	2325	560	4.1
		60	62	2530	2462	560	4.4
		90	78	2717	2698	560	4.8

Quadriceps force data from	Activity	Knee joint angle (degrees)	Patellar mechanism angle (degrees)	Quadriceps force (Newtons)	Patello-femoral contact force (Newtons)	Patello-femoral contact areas (millimeters²) Low load	Average	High load	Patello-femoral contact stress (N/mm²) Low load	Average	High load
Morrison	Walking	15	38	647	422	170	230	320	1.3	1.9	2.5
(1970 a, 1970 b)	Up ramp	30	46	785	618	210	260	410	1.5	2.5	2.8
	Down ramp	15	38	1923	1265	170	230	320	3.9	5.5	7.4
	Upstairs	45	54	1923	1756	260	330	480	3.7	5.3	6.9
	Downstairs	60	62	1687	1746	340	380	440	4.0	4.5	5.2
Smidt (1973)	Isometric	5	33	1393	795	–	–	–	–	–	–
	Quadriceps	15	38	1776	1158	170	230	320	3.6	5.0	6.8
	Contracture	30	46	2168	1697	210	260	410	4.2	6.6	8.0
		45	54	2433	2207	260	330	480	4.6	6.7	8.7
		60	62	2530	2609	340	380	440	6.0	6.8	7.8
		90	78	2717	3424	270	370	510	6.8	9.4	12.6

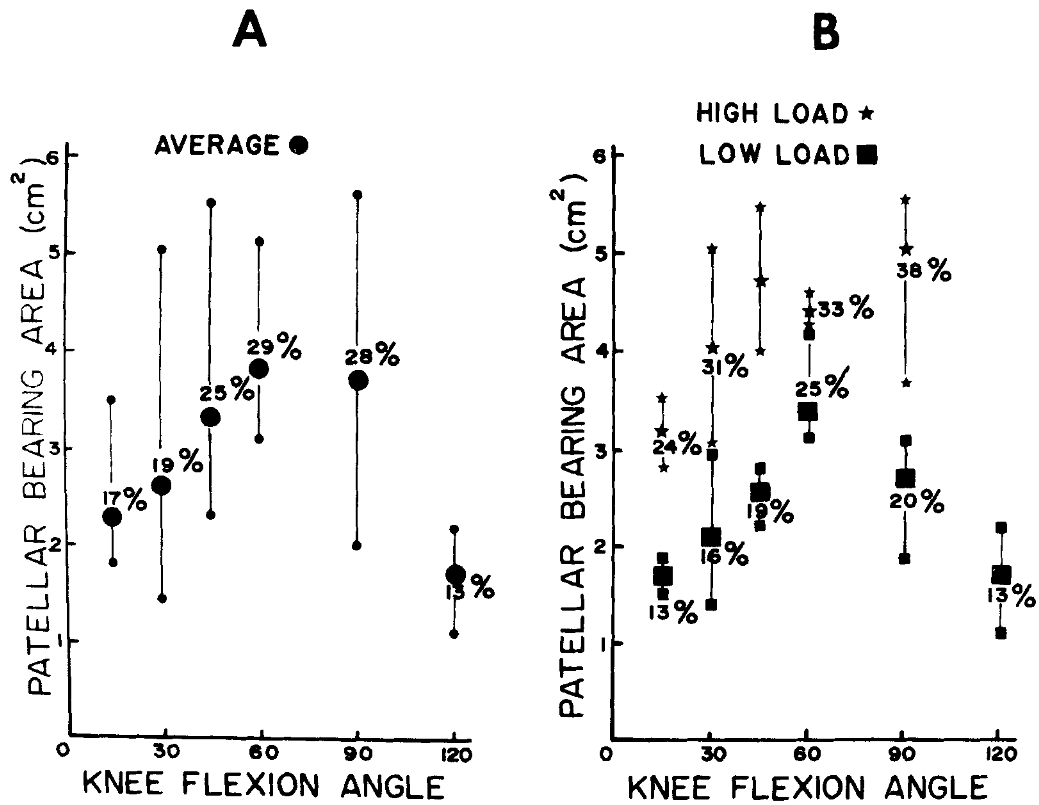

Figure 5. A. The averages and ranges of experimentally determined contact areas from all specimens at all knee flexion angles. B. The experimentally determined effect of the patello-femoral loading conditions.

of "low" and "high" patello-femoral loading conditions is presented in Figure 5 B.

An overall summary of our experimental data and contact stress calculations is presented in Table 1. Quadriceps forces are from the reports of Morrison (1907 a and b) and Smidt (1973). The tibial-femoral contact areas are from the report of Kettelkamp & Jacobs (1972). The patello-femoral forces were calculated from the equation: $F_{pf} = 2F_Q \sin \beta/2$ where F_{pf} is the patello-femoral force, F_Q is the quadriceps force and β is the patellar mechanism angle determined as previously described. Patello-femoral contact areas as measured are reported here, as well as the averages of those measured under "low load" experiments, "high load" experiments and the overall average for all experiments. Patello-femoral contact stresses were calculated from data by the equation: contact stress

$$= \frac{\text{force moment to surface}}{\text{surface area}}$$

and are presented as they were influenced by the above loading conditions.

DISCUSSION

Our experimental results and calculations from available data indicate that patello-femoral *forces* are generally less than tibial-femoral forces (Morrison 1970 a, b) during normal activities and during maximum isometric quadriceps contracture (Smidt 1973). We observed that the patella rarely touches the articular surface of the femur at full knee extension. Patello-femoral contact *areas* were less than total tibial-femoral contact areas for all degrees of knee flexion under all loading conditions (Kettelkamp & Jacobs 1972). It is of interest that our values for patello-femoral contact areas are in generally good agreement with those reported by Aglietti et al. (1975) using a casting technique employing the injection of acrylic cement into the joint with simulated quadriceps forces from zero to 5400 newtons.

Bearing surface contact stress is of great importance in prosthetic joint design. These studies indicate that the patello-femoral contact stresses are similar to those of the tibial-femoral joint during normal activities. Under all loading conditions, and for all knee flexion angles, patello-femoral stresses were considerably greater than tibial-femoral stresses during isometric quadriceps contracture. A recent report by Walker & Erkman (1975) proposes a load bearing role for the menisci with a total stressed area of up to 12 cm^2 observed in normal knees. If a load bearing function of the menisci is confirmed, then the patello-femoral contact stresses would be proportionately much greater than tibial-femoral stresses.

A prosthesis to treat the patello-femoral as well as the tibial-femoral articulations of the knee must necessarily be designed to function under these anatomically imposed high stress conditions.

High patello-femoral load values, small patello-femoral bearing surfaces, and the resultant high stress magnitudes indicate the need for caution in the design and development of a patello-femoral component for a total knee joint replacement prosthesis.

REFERENCES

Aglietti, P., Insall, J., Walker, P. S. & Trent, P. (1975) A new patella prosthesis. *Clin. Orthop.* **107**, 175–187.

Kettelkamp, D. B. & Jacobs, A. W. (1972) Tibiofemoral contact areas—determination and implications. *J. Bone Jt Surg.* **54**-A, 349–356.

Morrison, J. B. (1970) The mechanism of the knee joint in relation to normal walking. *J. Biomech.* **3**, 51–61.

Morrison, J. B. (1970) The mechanics of muscle function in locomotion. *J. Biomech.* **3**, 431–451.

Smidt, G. I. (1973) Biomechanical analysis of knee flexion and extension. *J. Biomech.* **6**, 79–92.

Walker, P. S. & Erkman, M. J. (1975) The role of the menisci in force transmission across the knee. *Clin. Orthop.* **109**, 184–192.

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