In vitro measurement of patellofemoral force after three types of knee replacement

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Using a new, non-invasive method, we measured the patellofemoral force (PFF) in cadaver knees mounted in a rig to simulate weight-bearing. The PFF was measured from 20° to 120° of flexion before and after implanting three designs of knee prosthesis.

Medial unicompartmental arthroplasty with a meniscal-bearing prosthesis and with retention of both cruciate ligaments caused no significant change in the PFF. After arthroplasty with a posterior-cruciate-retaining prosthesis and division of the anterior cruciate ligament, the PFF decreased in extension and increased by 20% in flexion. Implantation of a posterior stabilised prosthesis and division of both cruciate ligaments produced a decrease in the PFF in extension but maintained normal load in flexion.

There was a direct relationship between the PFF and the angle made with the patellar tendon and the long axis of the tibia. The abnormalities of the patellar tendon angle which resulted from implantation of the two total prostheses explain the observed changes in the PFF and show how the mechanics of the patellofemoral joint depend upon the kinematics of the tibiofemoral articulation.

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Patellofemoral problems are a common cause of failure after total knee replacement (TKR). Patellar resurfacing implants have often shown loosening or wear of their polyethylene surfaces^{1,2} and large increases in anterior patellar strain have been reported after TKR,^{3,4} suggesting

that joint replacement may have an adverse effect on the mechanics of the extensor mechanism of the knee.

We have devised a non-invasive method for measuring the patellofemoral force in a cadaver knee subjected to life-like loads in a laboratory rig.⁵ We now describe the use of that technique to measure changes in the magnitude of the patellofemoral force after the implantation of three designs of knee prosthesis into cadaver joints.

Materials and Methods

Non-invasive measurement of the patellofemoral force.

Three forces act on the patella in the sagittal plane (Fig. 1a). The quadriceps tendon force (QTF) and the patellar tendon force (PTF) are tensile; the patellofemoral force (PFF) is compressive. Equilibrium requires that, if only these three forces act on the patella, they must be coplanar and concurrent at the centre of force of the patella (P). In principle, our method of measurement of the PFF is to replace it with a measured tensile force (TF), applied at point P through a cable attached to the anterior aspect of the bone (Fig. 1b). The tension in the cable is increased until patellar lift-off occurs, at which point TF is deemed to have replaced PFF exactly (Fig. 1c). A detailed description of the testing rig, the method of determining the centre of pressure of the patella and the direction and magnitude of PFF have been published previously.⁵

Experimental procedure. Eight fresh human cadaver knees were taken at post mortem and quick frozen to -20°C. Immediately before testing each knee was thawed to room temperature and prepared by resecting the skin, subcutaneous fat and muscles. The lower part of the extensor mechanism, including the quadriceps tendon and the retinaculae, was kept intact. The tibia and femur were sectioned about 20 cm from the joint line and intramedullary rods cemented into the marrow cavities of both bones. The femoral rod was used to fix the specimen to the testing rig (Fig. 2). To test the knee at flexion angles between 20° and 80°, the femur was fixed at 45° to the horizontal; for flexion angles between 100° and 120°, the femur was fixed vertically. These positions facilitated applications of the QTF to the patella by hanging a 0.05 kg weight from the tibial rod at a distance of approximately 45 cm from the joint line. This load applied a flexing moment of up to

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Figure 1a – The three forces which act on the patella in the sagittal plane are the patellar tendon force (PTF), the quadriceps tendon force (QTF) and the patellofemoral force (PFF). For equilibrium, the forces must intersect at a single point (P), the centre of force. Figure 1b – A force (TF) is applied to the anterior aspect of the patella at P and gradually increased until the patella just lifts off. Figure 1c – The measured force TF is deemed to have exactly replaced the PFF.

2.2 Nm which was balanced by tension in a cable attached at one end to the quadriceps tendon and at the other to a turnbuckle. The angle of knee flexion was controlled by adjusting the length of this cable.

Each knee was tested in six positions from 20° to 120° of flexion. The angle of flexion was measured with a protractor and the TF and QTF were measured with strain-gauged proving rings in the two cables.

At each position of flexion, the angle in the sagittal plane between the patellar tendon and the long axis of the tibia was measured by a protractor. When the tendon inclined anteriorly from its tibial insertion, the angle was positive.

A defect of almost all cadaver studies in which load-bearing is simulated by tension in the quadriceps tendon is that the loads applied must be relatively small to avoid the otherwise common mishap of rupture of the connection between the slippery tendon and the cable.⁶ In our study, however, the model was used to make comparisons and, even if deformation of the soft tissues and articular surfaces under higher loads altered the ratio QTF:PFF:PTF, such alterations would have been expected to be similar in all the experimental situations. The magnitudes of the PFF measured in our model were comparable with those of several other in vitro studies in which pressure-sensitive films, electrical force transducers and tendon-force measurements had been used.⁵ The patellar tendon angles measured in our rig were the same as those given for the normal joint by Van Eijden, De Boer and Weijs.⁷

The prostheses. The prostheses were implanted using standard surgical methods and components of the appropriate size through a medial capsular incision which was then repaired. The patella was not resurfaced in any specimen. The three types of prosthesis were as follows:

1) The unicompartamental meniscal-bearing prosthesis (Oxford Knee; Biomet Ltd, Bridgend, UK). The prosthesis was used to replace the medial compartment of the joint. The implant incorporates a free meniscal bearing which is

Fig. 2
Diagram of the experimental rig.

Table I. The ratio of the QTF measured after arthroplasty to that measured before for the three designs of knee prosthesis (means of four specimens)

Flexion angle (degrees)	Unicompartmental		Posterior-cruciate-retaining		Posterior stabilised	
	Mean	SD	Mean	SD	Mean	SD
20	0.98	0.09	0.96	0.11	1.03	0.12
40	0.97	0.13	0.89	0.15	0.94	0.11
60	1.04	0.08	0.89	0.13	0.93	0.12
80	0.97	0.13	0.96	0.09	0.92	0.13
100	0.98	0.11	1.03	0.13	0.94	0.10
120	1.03	0.10	1.05	0.11	0.96	0.12

Table II. The ratio of the PFF measured after arthroplasty to that measured before for the three designs of knee prosthesis (means of four specimens)

Flexion angle (degrees)	Unicompartmental		Posterior-cruciate-retaining		Posterior stabilised	
	Mean	SD	Mean	SD	Mean	SD
20	1.02	0.08	0.83*	0.10	0.89*	0.10
40	1.04	0.11	0.86*	0.11	0.94	0.12
60	0.98	0.11	0.91	0.13	0.96	0.14
80	0.96	0.08	1.03	0.07	0.94	0.12
100	1.03	0.09	1.21*	0.09	1.04	0.11
120	1.02	0.02	1.24*	0.08	0.95	0.14

* statistically significant ($p < 0.05$) difference between results before and after arthroplasty (paired t -test)

unconstrained in the sagittal plane and depends for antero-posterior stability upon the retention of both cruciate ligaments.⁸ This prosthesis was implanted into the first four knees.

2) The posterior-cruciate-retaining condylar prosthesis (AGC Knee; Biomet Ltd, Bridgend, UK). This prosthesis is also relatively unconstrained, the tibial plateau having a surface which is flat in the sagittal plane except for a raised lip at its anterior margin.⁹ It is designed to allow retention of the posterior cruciate ligament. It was implanted into each of the first four knees after removal of the unicompartmental implant and section of the anterior cruciate ligament.

3) The posterior stabilised total condylar prosthesis (Insall Burstein II; Zimmer, Swindon, UK). This prosthesis is partially constrained in the anteroposterior plane by the concave form of its tibial plateau. In flexion, the femoral component engages a central tibial peg which forces it to roll to the back of the tibial plateau.¹⁰ It was implanted into the second four knees after section of both cruciate ligaments.

The patellar retinaculae. The unicompartmental and the posterior-cruciate-retaining prostheses were tested with the medial retinaculum repaired and with the lateral retinaculum remaining intact. During the experiments, the point of patellar lift-off was easily determined in all these specimens. In two of the four posteriorly stabilised implants, however, repair of the medial retinaculum caused the patella to be held so firmly in the intercondylar groove that it could not be lifted off even in the unloaded state. Both retinaculae had to be divided before the testing procedure could be undertaken satisfactorily.

Statistical analysis. Each specimen acted as its own control and the measurements before and after knee replace-

ment were compared using the paired Student's t -test.

Intraobserver error. All the measurements were made by one observer (RKM). An intraobserver error study reported previously has shown that the errors in measuring the forces were about 1%.⁵

Results

The PFF and QTF after implantation were expressed as the ratios of the forces measured in the same specimen before implantation (i.e., 1 = no change after knee replacement).

Quadriceps tendon force. There was no significant change in the QTF after implantation of any of the prostheses (Table I).

Patellofemoral force. Measurements of the PFF after knee replacement are given in Table II and Figure 3. Implantation of the unicompartmental prosthesis produced no significant change in the PFF, but after implanting the posterior-cruciate-retaining implant, it was significantly lower than that in the intact knee at 20° and 40° of flexion and significantly higher at 100° and 120° of flexion. With the posterior stabilised implant, the PFF was significantly lower at 20° of flexion but there were no other significant changes.

Patellar tendon angle. The angles between the patellar tendon and the tibial axis are given in Table III and Figure 4. The unicompartmental prosthesis reproduced the normal pattern of patellar tendon angle. After implanting the posterior-cruciate-retaining implant, the angle of the patellar tendon was less than normal at 20° and 40° of flexion and greater than normal at 100° and 120° of flexion. With the posterior stabilised implant, the angle was less than normal at 20°, 40° and 60° of flexion, but normal at higher degrees of flexion.

Fig. 3

The mean change ($\pm$ SEM) in the PFF with the three designs of implant. The PFF after arthroplasty is expressed as a percentage of the PFF before arthroplasty.

Table III. The patellar tendon angle before and after arthroplasty with the three designs of prosthesis (means of four specimens)

Flexion angle (degrees)	Before arthroplasty		Unicompartamental		Posterior-cruciate-retaining		Posterior stabilised	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
20	15	2.9	17	2.3	10*	3.3	8.9*	1.9
40	12.5	3.0	13.0	2.7	9.2*	3.1	4.9*	1.8
60	6.3	3.8	6.5	2.7	5.5	3.7	2.2*	1.7
80	0.8	1.5	1.8	2.1	2.8	3.1	-2.2	2.6
100	-6.5	3.8	-7.1	1.7	1.0*	1.4	-9.5	4.4
120	-11.7	7.0	-12.0	2.3	1.2*	2.5	-11.0	2.2

* results significantly different ($p < 0.05$) from normal

Fig. 4

The mean ($\pm$ SEM) patellar tendon angle in the normal knee and with the three designs of implant.

Fig. 5

The magnitude of the patellofemoral force (PFF) varies inversely with the angle between the patellar tendon (PT) and the quadriceps tendon (QT), the patellar mechanism angle.

Discussion

Direct measurement of the compressive patellofemoral force is difficult in the intact joint because of the need to place the measuring device within the joint;¹¹⁻¹⁴ this difficulty is compounded by knee replacement. Several authors, however, have reported large increases in anterior patellar strain after arthroplasty.^{3,4} In our study, the PFF was measured indirectly by determining the equivalent tensile load

applied to the anterior aspect of the patella, with the advantage that the patellofemoral joint was not invaded. This technique has been validated in cadaver knees and found to agree better with theoretical estimates of patellofemoral force than did invasive methods of measurement.^{5,15}

The method of loading the joint, allowing six degrees of freedom to the tibia, is most likely to produce a close simulation of load-bearing in life.¹⁶ The magnitudes of the loads used were much less than occur in life. The forces are therefore expressed non-dimensionally in the belief that their proportions are largely independent of their absolute values.¹⁷

The different patellar tendon angles which were observed after each arthroplasty correlated with, and may explain, the changes in the PFF. The unicompartmental replacement caused no alteration in the patellar tendon angles and no change in PFF. Both the tricompartmental implants caused the tendon angle to decrease near extension and with both, the PFF was less than normal near extension. Only the posterior-cruciate-retaining implant caused the tendon angle to increase in flexion and produced an abnormally high PFF in flexion.

Mechanical analysis. The magnitude of the PFF varies directly with the QTF and inversely with the angle between the quadriceps tendon and the patellar tendon, the patellar mechanism angle¹⁸ (Fig. 5). The more acute the patellar mechanism angle, the greater is the PFF as a proportion of the QTF. Figure 6 shows that, in all positions of the joint, anterior displacement of the tibia relative to the femur diminishes, and posterior displacement increases the patellar tendon angle.

In the normal knee, anterior and posterior displacement is controlled mainly by the cruciate ligaments, the tibial plateau being almost flat and providing little constraint in the anteroposterior plane. These ligaments, acting like a

Fig. 6

In extension in the normal knee (discontinuous line) the patellar tendon is inclined anteriorly from its tibial insertion and therefore the patellar tendon angle is positive. After both types of total arthroplasty (solid line), the tibia subluxes forwards in extension, thus reducing the patellar tendon angle.

In flexion in the normal knee (discontinuous line) the patellar tendon is inclined posteriorly and therefore the patellar tendon angle is negative. After arthroplasty with the posterior-cruciate-retaining implant (solid line) the tibia subluxes posteriorly, thus making the patellar tendon angle positive.

Fig. 7

Parallelograms of force demonstrating that changes in the magnitude and/or the direction of the PTF and PFF need not change the magnitude or direction of QF.

four-bar linkage, cause the femur to roll backwards in flexion and forwards in extension.¹⁹ Simultaneously, in the normal knee, the patellar tendon angle changes from about $+20^\circ$ in full extension to 0° at 90° and to -15° in full flexion.⁷ The different patterns of patellofemoral loading in the three implants may be explained by the degree to which they succeeded in reproducing the effects of the cruciate mechanism.

With the unicompartmental implant, the freely mobile bearing allowed the intact cruciate ligaments to reproduce the normal movements between the femur and the tibia throughout the range of flexion, and the tendon angles and PFF were normal in all the positions tested. Observations in the laboratory²⁰ agree with radiological studies of this implant in patients which showed that the bearings move posteriorly in flexion and anteriorly in extension.²¹

The posterior-cruciate-retaining prosthesis was implanted, as it is designed to be used, into joints which lacked the anterior cruciate ligament. Our observations suggest that in extension the tibia subluxed forwards (Fig. 6) reducing the patellar tendon angle, and in flexion it subluxed backwards increasing the tendon angle.

Anterior tibial subluxation in extension is readily explained as the result of the anterior pull of the patellar tendon which is unresisted in the absence of the anterior cruciate ligament. The effect on the patellofemoral force is similar to that of anterior displacement of the tibial tubercle, the procedure designed by Maquet to reduce the compressive load on the patella.²²

The observed posterior subluxation of the tibia in flexion suggests that the retained posterior cruciate ligament was ineffective. Posterior-cruciate-retaining implants are designed in the belief that the intact posterior ligament will continue to function physiologically even in the absence of its partner, the anterior cruciate ligament. Our observations

suggest that this expectation may not be well-founded. If the anterior cruciate ligament is ruptured, the posterior ligament is free to adopt a more vertical position, rendering it relatively less effective in controlling the antero-posterior position of the femur on the tibia. Paradoxical movements like those which we observed in the laboratory have also been described in patients when unconstrained prostheses were implanted into anterior-cruciate-deficient knees.²³

The posterior stabilised implant is designed to function in the absence of both cruciate ligaments. Like the previous implant the prosthesis caused a decrease in the patellar tendon angle in extension with a consequent fall in the PFF, suggesting that its articular surfaces offered too little resistance to anterior tibial subluxation. In flexion, the polyethylene tibial peg which engages the femoral component prevented posterior subluxation of the tibia, reproducing the normal pattern of patellar tendon angles beyond 90° and, consequently, providing normal loading of the patellofemoral joint. In two specimens lateral release was necessary, suggesting that the non-anatomical shape of the femoral component overstretched the retinaculæ and induced an increased PFF from tension in these lateral structures. Patellar fractures have been reported as a complication of the use of this prosthesis.²⁴

Despite the alterations in the patellar tendon angle and PFF, the QTF remained normal. An abnormal location of the femur on the tibial plateau has only a slight effect on the geometrical relationship between the quadriceps tendon and the femur. The patellar mechanism angle can therefore increase or decrease and the PFF and the PTF can decrease or increase without changing the quadriceps force (Fig. 7).

Conclusions. The cruciate mechanism is essential to the maintenance of physiological function of the patellofemoral joint. If both cruciate ligaments can be retained, the normal mechanics of the patellofemoral joint can be restored by an implant which allows unconstrained translational movements of the femur on the tibia.

In the absence of its anterior component, the cruciate mechanism becomes ineffective and an unconstrained implant may allow paradoxical movements of the femur on the tibia in the sagittal plane. The assumption that retention of the posterior cruciate ligament will restore normal 'roll-back' of the femur may not be well founded. Loss of the physiological roll-back of the femur induces abnormally high patellofemoral forces in flexion.

In the absence of a functioning cruciate mechanism, a mechanical interlock which prevents posterior subluxation of the tibia as the knee flexes can successfully avoid abnormally high patellofemoral forces in flexion, provided that the release of the retinaculæ is undertaken when required.

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