



## Basic kinematics and biomechanics of the patello-femoral joint Part 1 : The native patella

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The patello-femoral joint (PFJ) enhances our ability of knee flexion and extension and is assumed to have evolved through men's ability of having adopted a bipedal gait. This articulation between patella and femur is relatively complex and displays intricate biomechanical behaviour. Forces in the patello-femoral joint are a function of the quadriceps force, and the angle of flexion of the knee. They are highly dependent on the distance between the patello-femoral joint and the centre of gravity, which explains why different activities despite equivalence in tibio-femoral angle may exert wide variations in patello-femoral reaction forces and contact pressures. During normal daily activities the PFJ becomes exposed to force values between  $0.5$  to  $9.7 \times$  body weight, whilst sporting activities create force values that approach up to  $20 \times$  body weight. Those forces are considerable and it is therefore not surprising that the PFJ is particularly susceptible to degenerative disease especially if its mechanical equilibrium is disturbed through injury, muscle weakness and congenital or developmental abnormalities. The clinician must be aware of biomechanical and kinematic specifics, the high patello-femoral load values, small patellofemoral contact areas, and resultant high stress magnitudes when trying to remedy such abnormalities.

**Keywords :** patellofemoral joint ; kinematics ; biomechanics.

### INTRODUCTION

Any clinician contemplating treatment for conditions affecting the patello-femoral joint must

possess some basic knowledge of anatomy, biomechanics and kinematics of the knee and the locomotor system. Without such understanding it is difficult to appreciate the implications associated with the various surgical and conservative treatment modalities. This is of particular importance when considering surgical remedies in the treatment of patello-femoral disorders (e.g. localised cartilage defects, chondromalazia) especially if associated with patella mal-tracking, as changes in the relationship between patella and femur may significantly alter the distribution of forces and any overcorrection may potentially hasten the development of degenerative disease. Similar concerns exist with regard to the treatment of the patella in total knee arthroplasty, where surgically imposed changes through resurfacing may have significant effects on performance and behaviour of the patello-femoral joint (1,11). Complications arising from patella resurfacing are still considerable and analysis of retrieved patellar components and the significant

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failure rate of metal backed patella designs continue to underscore the extreme mechanical environment in which these implants are expected to perform (8,12,42).

### **Kinematics of the patello-femoral joint**

Kinematics of knee joint characterises the relative motion that exists between femur, tibia and patella (53). The patella is a sesamoid bone implanted within the tendon of the extensor mechanism. Arguably its most important function is its role in facilitating extension of the knee by increasing the efficacy of the quadriceps muscle (26). This is achieved through the patella's function as a fulcrum, thus anteriorly displacing the line of pull and increasing the moment arm of the quadriceps muscle force in relation to the centre of rotation of the knee. The patella has shown to enhance the force of extension by as much as 50% throughout the entire range of motion (53). The patella also facilitates improved distribution of patello-femoral compressive force on the femur through an increase in contact area during flexion (Fig. 1). In addition the patella acts as a guide for the extensor mechanism by centralising the divergent pull from the four muscles of the quadriceps and transmitting these forces to the patella tendon. Together with the anatomical shape of the patello-femoral articulation this protects the extensor apparatus from dislocating.

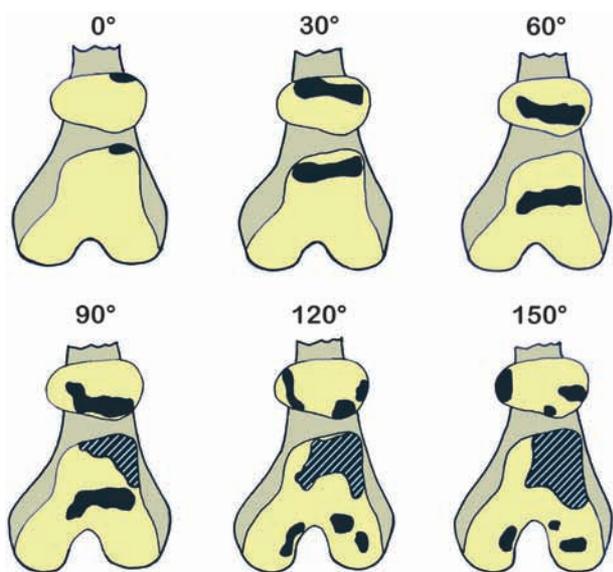
When the knee is extended, the tightened quadriceps pulls the patella upwards until the upper border reaches beyond the femoral trochlea groove. So long as the line of gravity falls behind the center axis of the knee joint when standing upright, the quadriceps must contract to neutralize the rotatory effect of gravity on the knee, which would otherwise force the knee into flexion. As soon as the line of gravity falls within or in front of the knee, as seen in full extension or hyperextension, the quadriceps becomes relaxed. The quadriceps apparatus, being oblique in its angulation towards the patella and patellar tendon, creates a line of pull with an outward directed horizontal component when contracted. The angle between the line of pull and the patella tendon is often referred to as the Q-angle,

which is responsible for a tendency of the patella to slip outward over the lateral femoral condyle creating a lateral force vector (18). To offset this propensity the lateral condyle projects farther forward whilst the fibres of vastus medialis which secure the patella medially extend farther distally compared to those of the vastus lateralis (53). The reversal of the 'screw home mechanism' during the initial 30° of knee flexion, essentially derotates the tibia, leading to a reduction in Q-angle and lateral force vector (7,19,23,40,41). In the coronal and axial views, this sideways component (lateral vector), which reduces with knee flexion, is balanced by the reaction occurring on the slope of the femoral trochlea (Fig. 2) (60). Hence the patella is most vulnerable during the initial degrees of knee flexion, when its engagement into the trochlea may still be incomplete whilst the effect of the Q-angle, albeit reduced, remains present.

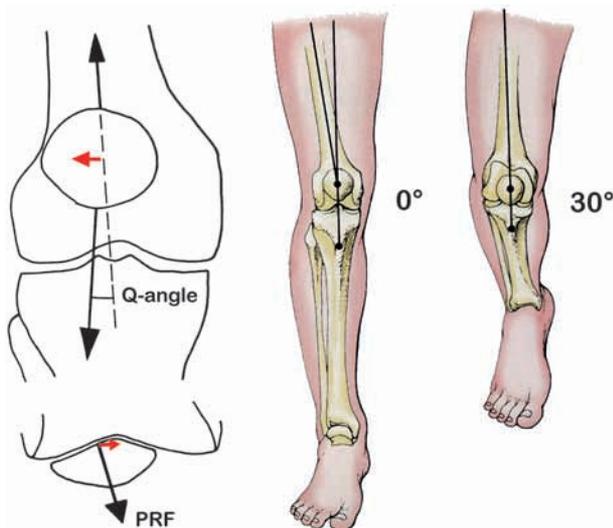
### **Patello-femoral contact areas**

The patella is usually out of contact with the trochlea groove in full extension. Depending on the length of the patellar tendon, the patella is drawn into the trochlea from a slight lateral position and gains contact with the femur between 10° to 20° (Fig. 1) (20,51). The contact begins with the inferior margin of the patella and moves proximally as flexion proceeds (Fig. 3) (36). Beyond 30° the patella settles into the deepening trochlea groove where it is further stabilised by the quadriceps and patellar tendon force.

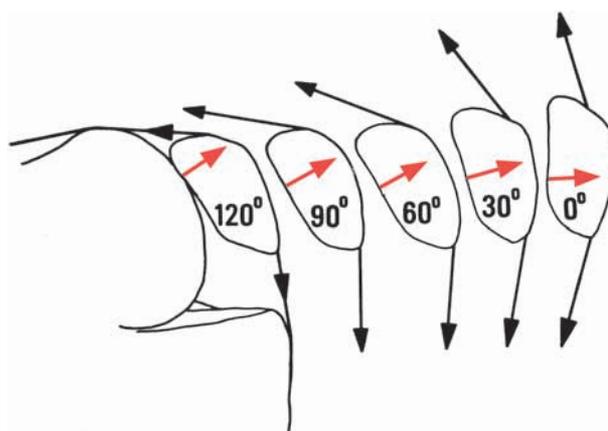
The patello-femoral contact area extends from the medial margin of the medial facet to the lateral margin of the lateral facet as a broad band of contact moving from distal to proximal (2,19,22,25,29). Between 30° to 60° of flexion the contact is across the centre, at 90° of flexion the contact moves towards the superior pole, and beyond 90° the patella is astride the medial and lateral condyles, forming two separate contact areas (Fig. 4). During flexion the patella maintains a lateral shift as well as a subtle degree of rotation around a longitudinal axis, positioning the medial facet more posterior (47,56,58). In full flexion the lateral femoral condyle is completely covered by the lateral patella



**Fig. 1.** — Patello-femoral (solid area) and tendo-femoral (shaded area) contact areas at various degrees of knee flexion.



**Fig. 2.** — Internal rotation of tibia during flexion neutralises Q-angle and reduces lateral patello-femoral vector. In the coronal and axial views, the sideways component is balanced by the reaction occurring on the slope of the femoral trochlea (Adapted from Walker (60), courtesy of Charles C Thomas Publishers Ltd, Springfield, Illinois).\*

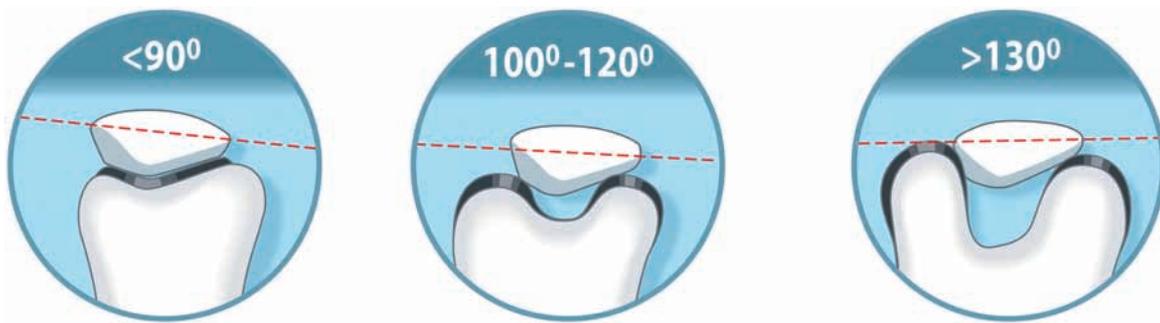


**Fig. 3.** — Contact points between patella and femur move from proximal to distal on the femur and from distal to proximal on the patella during knee flexion (adapted from Walker (60), courtesy of Charles C Thomas Publishers Ltd, Springfield, Illinois).\*

facet whilst the medial condyle is almost completely uncovered being merely in contact with the odd facet. Ficat and Hungerford have described this movement of the patella in the coronal plane during flexion as a ‘gentle curve with its concavity facing laterally’ (19). Studies of the tracking pattern of the natural patella have confirmed that the patella rotates as much as 12° to 15° in relation to the femur, with most of the rotation occurring beyond 50° of knee flexion (31,47,49,56,57). Furthermore the patella tilts about a medio-lateral direction in the axial plane, being influenced by knee flexion, the degree of internal or external rotation and the varus/valgus alignment of the tibio-femoral joint (56,57). Similarly, the patella undergoes medially directed displacement by as much as 5 mm in the coronal plane, with most of the displacement occurring during the initial 30° of knee flexion.

In the transverse or axial plane, as seen on skyline radiographs, the patella is perfectly congruent with the trochlea, assuring its medial/lateral stability. Longitudinal sectioning of the patella, as performed by Krakow and Hungerford, has confirmed that the patella adopts almost a flat surface in the sagittal plane making it perfectly unconstrained as far as its anatomical form is concerned (34). They concluded that the length of the patellar tendon and the angle between the patellar tendon and the

**FIGURES** (\* where necessary permission has been obtained from publisher).



**Fig. 4.** — Bifurcation of patello-femoral contact area beyond  $100^\circ$  of knee flexion. The odd facet of the patella only makes contact with the medial femoral condyle beyond  $125^\circ$  to  $135^\circ$  of knee flexion.

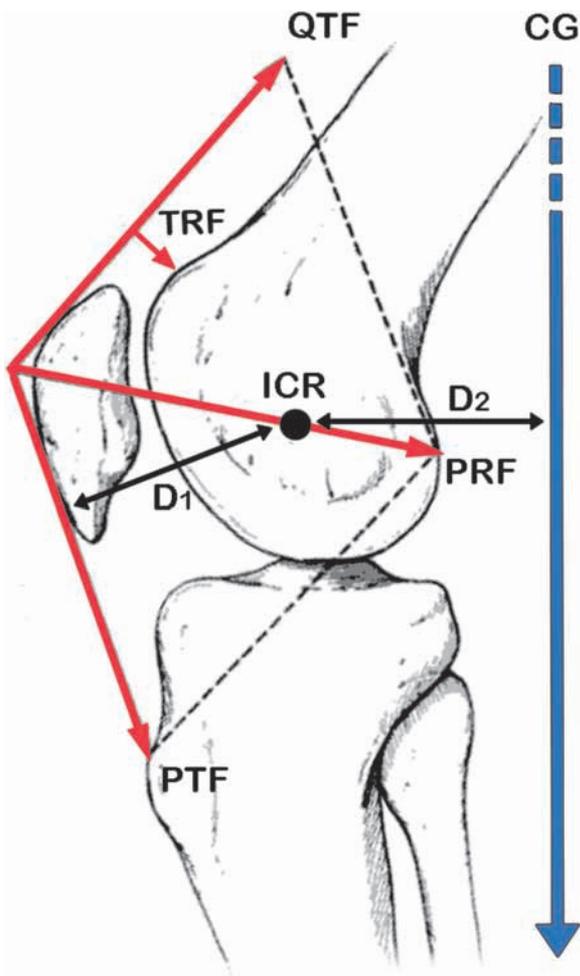
quadriceps tendon determine the load bearing area of the patella. The patella form, therefore, provides stability against lateral subluxation but does not impede the patella from rocking around its transverse axis to the point at which the resultant of the patello-femoral joint reaction force is perpendicular to the contact surface.

As with the location of patello-femoral contact areas, the size of the contact areas is highly dependent on knee position. From  $20^\circ$  to  $60^\circ$  of flexion the average contact area increases linearly from around  $150 \text{ mm}^2$  to  $480 \text{ mm}^2$  (2,28,39,55). It then remains almost constant up to about  $90^\circ$  of flexion after which a linear reduction occurs (25,28,39). At  $120^\circ$  the contact area will have dropped to  $360 \text{ mm}^2$  (55). Due to the drastically changed contact pattern beyond  $100^\circ$ , when the patella leaves the trochlea straddling the intercondylar notch, contact areas may fall well below  $100 \text{ mm}^2$  at full flexion (Fig. 4) (25). Matthews *et al* have shown that on average only 19% of the patella bearing surface is engaged at  $30^\circ$  of flexion, 29% at  $60^\circ$ , 28% at  $90^\circ$ , and 13% at  $120^\circ$  (39). These values compare favourably with those obtained by Huberti *et al* who measured percentage values of the patello-femoral contact area of between 20.5% to 32.2% (28).

### Biomechanics

Wide attention has been given to define the force transmission in the patello-femoral joint. For ease of calculation it is sufficiently accurate to consider these forces to lie in a sagittal plane. Reaction

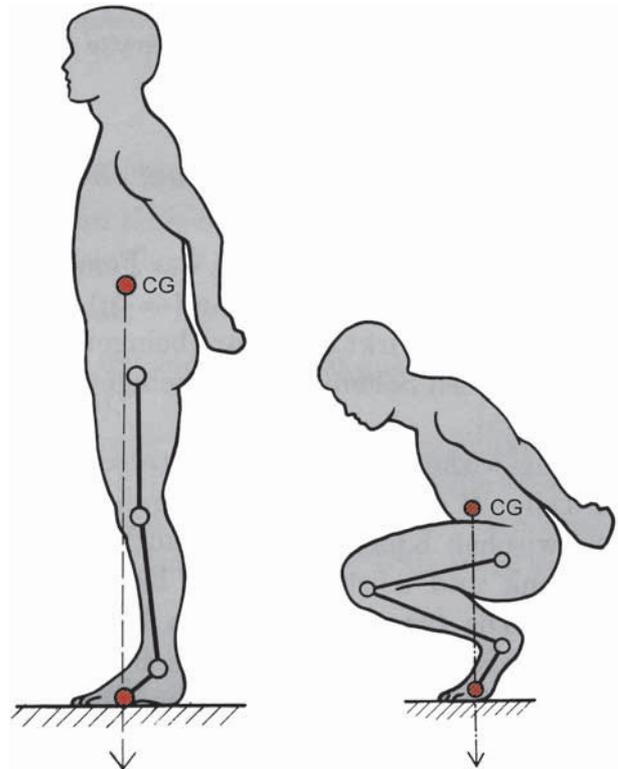
forces are equal and opposite forces that exist between adjacent bones at a joint and caused by the weight and inertial forces of the two segments. The term patello-femoral compressive force (PCF), representing the sole load acting on the patella, and patello-femoral reaction force (PRF) may be used interchangeably, although it is conceivable that the resultant (reaction) force produced by the quadriceps mechanism at different angles of flexion may be broken into normal (compressive) and tangential force components. For the purpose of this review however, we may assume equality between PCF and PRF, with the latter acting perpendicular to the articulating surface of the patella and equal and opposite to the resultant of the patellar tendon and quadriceps force, based on the 'Parallelogram of Forces' (Fig. 5) (60). With increasing flexion the line of reaction moves upwards leading to an increase in PRF for the following reasons. As the angle between the patella tendon and the quadriceps becomes more acute the resultant force vector increases. With knee flexion effective lever arms of femur and tibia increase, requiring greater quadriceps power to resist the flexion moment of the body weight. Close to extension the PRF is only about  $1/3$  of the quadriceps force, whilst beyond  $60^\circ$  of flexion the patella force is about  $1 \frac{1}{4}$  times the quadriceps force (60). The line of pull between the quadriceps and patella tendon when viewed in the coronal plane is affected by the Q-angle, giving a resultant force in the lateral direction (Fig. 2) (19). In the axial plane this is balanced by a reaction force which is inclined inwards. Walker conceded



**Fig. 5.** — Free-body diagram of the patello-femoral joint showing the ‘Parallelogram of Forces’. The patella-femoral reaction force (PRF) is the resultant vector of the quadriceps tendon strain force (QTF) and the patellar tendon strain force (PTF). The tendo-femoral reaction force (TRF = force between quadriceps tendon and trochlea) is also shown. CG = centre of gravity. ICR = instant centre of rotation.

that in the midrange of flexion up to 60°, the condition for lateral stability of the patella is namely that the angle of inclination of the lateral trochlea groove is larger than the Q-angle (60,61).

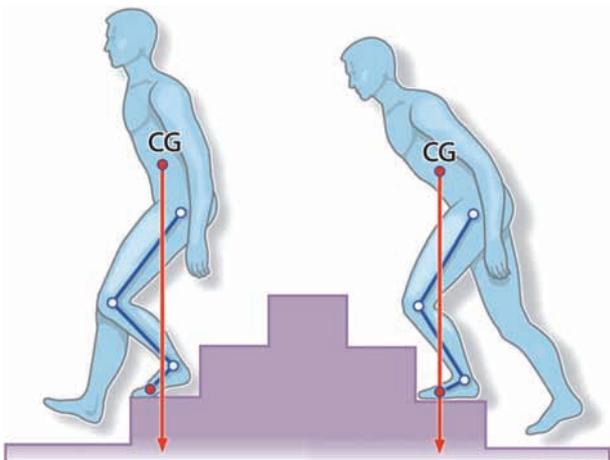
The most important variable in the calculation of static forces is the distance between the line of body weight (centre of gravity) and the patello-femoral joint. Changes of posture in the sagittal plane (leaning forward or backward) will alter this distance and lead to substantial differences in static force transmission, whilst changes in the coronal plane



**Fig. 6.** — Moment arms acting on the patello-femoral joint are depending on body position and the distance between the centre of gravity (CG) and the patello-femoral joint (Adapted from Bandi (5,6), courtesy of Hans Huber Verlag, Bern, Switzerland).\*

will exert little influence (Fig. 6) (5,14). In full extension the centre of gravity falls anterior to the knee, moment arms become 0, hence no forces acting on the patello-femoral joint (5). Whenever the line of body weight is moved posterior and away from the patello-femoral joint, muscle activity and tension in the patellar ligament will increase to maintain position, resulting in higher patello-femoral compressive forces.

During normal activities requiring flexion under load, hip flexion is also present, thus bringing the centre of gravity forward and shortening the femoral lever arm. This relationship is exemplified by the skier, who, by leaning backwards on his skis, has to increase his quadriceps force to prevent a fall. This in turn dramatically increases PRF potentially leading to quadriceps tendon or anterior cruciate ligament rupture (6). On the other hand the patient with quadriceps weakness can rise from a

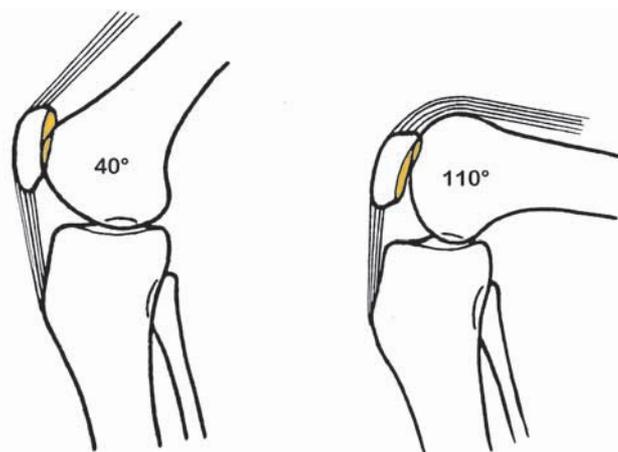
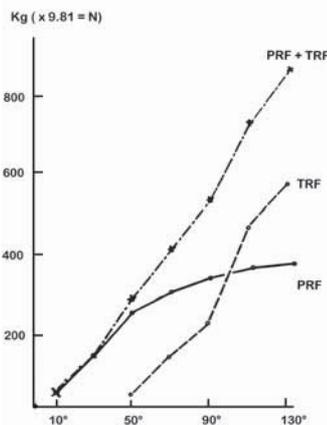


**Fig. 7.** — During stair ascend the centre of gravity (CG) is positioned almost above the patello-femoral joint, hence moment arm of femur and tibia are relatively short and the patello-femoral reaction force (PRF) is low. During stair descend the CG is positioned further posterior to the patello-femoral joint, creating longer moment arms and a subsequent increase in PRF.

chair by leaning forward, bringing the centre of gravity closer to the knee. Similarly there are significant differences in PRFs in ascending and descending stairs (Fig. 7). Predicted force values for stair ascend range from 1.8 to 2.3 × BW, compared to those for stair descent which are ranging from 2.9 to 6 × BW (4,45). The increased values on descending are due to the centre-of-gravity being moved further

backward behind the patello-femoral joint in order to maintain balance. Force transmission in the patello-femoral joint is therefore dependent on the relationship between the centre of gravity of the body and the knee flexion angle and calculations should not be based simply on the length of the femur and the position of the hip joint alone (5). To demonstrate the calculation of PRF we may use a simplified model of a person standing with both knees flexed to 45° and with half the body weight (0.5 × BW) being transferred through each knee (Fig. 5). In this position the centre of gravity is approx. 42 mm posterior, and the patella tendon 30 mm anterior to the centre of rotation of the knee. Let us further assume that there is no frictional loss at the patello-femoral interface and quadriceps and patellar tendon forces are equal. The patellar tendon force (PTF) = (0.5 × BW) × 42 mm / 30 mm = 0.7 × BW. The parallelogram of forces can now be scaled to obtain a value of approx. 0.8 × BW for the resultant patello-femoral force.

In 1911 Fick first recognised that the quadriceps tendon started to abut onto the proximal aspect of the femoral trochlea in mid range of knee flexion (20). As a result the compressive forces become divided between the tendo-femoral and patello-femoral contact areas (6,21,24,44). This phenomenon later described by Goymann & Müller as the ‘turn-round’ of forces, represents an elegant way of main-



**Fig. 8.** — Patello-femoral reaction force (PRF) plotted against the tendo-femoral reaction force (TRF). Calculated values for PRF show noticeable tail-off beyond 50° of knee flexion due to the turn-round phenomenon of the quadriceps tendon against the femoral trochlea (adapted from Bandi<sup>6</sup>, courtesy of Hans Huber Verlag, Bern, Switzerland).\*

Table I. — Static measurements of patello-femoral joint reaction forces in relation to knee flexion angle.  
(Values should be viewed with due regard to the complexity of the problem and with the knowledge of the assumptions which must necessarily be made in obtaining them)

Knee flexion angle	Average PF contact area (mm <sup>2</sup> )	Percentage of total contact area <sup>o</sup>	Tendo-femoral compressive force (Newtons)	Peak PRF (Newtons)	Peak PRF (body weight)	Peak PF contact pressure (N/mm <sup>2</sup> )
0°	140	10	0	0	-	-
10°	200	15	0	100	0.2	0.5
30°	280	20	0	300	0.5	1.1
50°	320	23	250	860	1.2	2.7
70°	450	32	1300	1810	2.7	4.0
90°	350	25	2200	2860	4.2	8.1
110°	260	19	4500	3300	4.8	12.7
135°	130	9	5800	7500	12.9	57.7

<sup>o</sup> based on an average total contact area of 1340 mm<sup>2</sup>.

taining relatively constant unit load under a mechanical situation where total load is increasing (Fig. 8) (24,25). The efficacy of the turn-round of the divided forces is dependent on the length and the altitude of the patella (24). The 'turn-round' of forces takes effect between 50° to 90° of flexion (5,21,25). According to measurements obtained by Hehne the contact area of the quadriceps tendon is significantly larger compared to the contact area of the patello-femoral joint (27). At 90° the quadriceps contact area measures approx. 1 to 2 times, at 120° 2 to 3 times, and at 140° 3 to 4 times that of the patello-femoral contact area (Fig. 1).

Huberti *et al* calculated average tendo-femoral contact forces at 120° of approximately 550N, whilst patello-femoral contact forces at the same degree of knee flexion measured on average 1600N, indicating a ratio of 1:3 between tendo-femoral and patello-femoral contact force (28). This may at least to some extent explain the higher frequency of chondromalacia in patella alta, as tendo-femoral contact may be eliminated or substantially decreased, creating an increase in patello-femoral reaction force (3,28,30,35).

Static measurements of patello-femoral reaction forces (PRF) have been reported as showing an almost linear increase in force values up to 110° after which they decline (Table I). The first such measurements were performed by Burckhardt in 1924, who disregarded the load sharing function of

the quadriceps tendon, hence his results have been flawed (10). Furmaier in 1953 and Bandi in 1972 made appropriate adjustments in their force calculations incorporating tendo-femoral contact forces (5,21). Calculated PRF values range from 0 × BW at 15° to 12.9 × BW at 135° (5,21,28,38,46,51).

Patello-femoral reaction forces (PRF) during activities vary greatly and are essentially dependent on the type of activity performed (Table II). Predicted force values range from 0.6 × BW for level walking to 7.7 × BW for jogging, and 20 × BW for jumping (9,13,16,17,32,33,43-45,48,52,59,62,63). For isokinetic exercise Kaufman *et al* found that PFR peaked at around 70° to 75° of knee flexion. Calculated values are dependent on exercise speed and ranged from 3.4 to 6.8 × BW. Ericson and Nisell noted that both patello-femoral and tendo-femoral reaction forces during cycling were generally lower when compared to those generated through daily and most other sporting activities (Fig. 9) (17). The magnitude of joint forces was almost independent of body weight, but increased with work load and reduced saddle height. Tendo-femoral reaction force rose to 295N at 108° of knee flexion. Anterior knee pain during cycling may henceforth be associated with compression of the supra-patellar bursa or medial para-patellar plica at higher knee flexion angles if the saddle position is kept low and knee flexion angles are subsequently increased throughout all stages of the revolu-

Table II. — Patello-femoral joint reaction forces for various activities. (Values should be viewed with due regard to the complexity of the problem and with the knowledge of the assumptions which must necessarily be made in obtaining them)

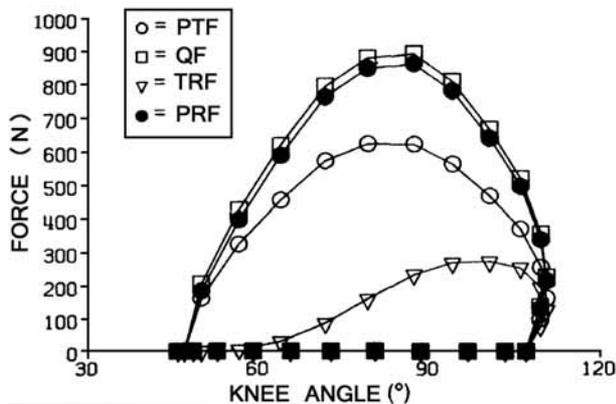
Author	Year	Activity	Body Weight (kg)	Knee flexion Angle	Peak PRF (N)	Peak PRF (xBW)
Reilly & Martens	1972	Level walking	70	10°	334	0.5
Morra & Greenwald	2006	Walking gait	-	15°	420	0.6
Bresler & Frankel	1950	Level Walking	71	20°	840	1.2
Ericson & Nisell	1987	Cycling	71	83°	905	1.3
Nisell	1985	Lifting (12.8 kg box)	77	90°	1600	2.2
Andriacchi <i>et al</i>	1980	Stair ascent	71	65°	1500	2.1
		Stair descent	71	60°	4000	5.7
Morra & Greenwald	2006	Stair ascent	-	45°	1760	2.5
Reilly & Martens	1972	Stair walking	85	55°	2500	3.3
Smidt	1973	Isometric quads contraction	82	75°	2127	2.6
Morra & Greenwald	2006	Rising from a chair	-	90°	1950	2.8
Ellis <i>et al</i>	1979	Rising from a chair	-	120°	-	3.1
Kelley <i>et al</i>	1978	Rising from a chair	-	90°	3800	5.5
Kaufman <i>et al</i>	1991	Isokinetic exercise	81	70°	-	5.1
Huberti & Hayes	1984	Isometric extension		90°	4600	6.5
Nisell	1985	Isometric extension	72	90°	6900	9.7
Dahlqvist <i>et al</i>	1982	Ascending from squat		140°		6.0
		Descending from squat		140°		7.6
Reilly & Martens	1972	Squatting	85	130°	6375	7.6
Winter	1983	Jogging	72	50°		7.7
Wahrenberg <i>et al</i>	1978	Kicking	76	100°	5800	7.8
Smith <i>et al</i>	1972	Jumping	-	-		20
Nisell	1985	Quadriceps tendon rupture			10900-18300	14.4-24.2
Zernicke <i>et al</i>	1977	Patellar tendon rupture	-	90°		25

tion (15,50). Cycling should be considered a preferable activity for most patients recovering from knee surgery, but especially for those who are obese. Proper attention however should be given to appropriate adjustment of work load and saddle height..

When considering the magnitude of patello-femoral compressive force it has to be remembered that this force acts through an area which varies with knee flexion (37,54,60). Henceforth an increase in PRF does not necessarily assume an increase in patello-femoral pressure. Patello-femoral contact area is small close to full extension of the knee, indicating that patello-femoral pressure is higher for the same PRF magnitude. From this follows,

that the patello-femoral pressure, near extension may be relatively high although the compressive force appears comparatively low (44).

To obtain a rough estimate of the resulting contact pressures in the patello-femoral joint, the mean pressure is calculated by dividing patello-femoral force values by the patello-femoral contact area. Patello-femoral contact pressure values are dependent on activity and knee flexion angle, and range from 1.28 to 12.6 N/mm<sup>2</sup> (28,29). Accordingly a 696 Newton man climbing stairs would generate a patello-femoral compression force of 1754N equivalent to 2.5 × BW and experience patello-femoral pressures between 3.73 and 6.87 N/mm<sup>2</sup> (39).



**Fig. 9.** — Patello-femoral joint forces during cycling motion. Patello-femoral reaction force (PRF), patellar tendon strain force (PTF), quadriceps strain force (QF), and tendo-femoral reaction force (TRF) plotted against the knee flexion angle during a complete revolution (Redrawn and adapted from Ericson & Nisel (17), with permission of the American Physical Therapy Association).\*

Women have smaller knees and hence shorter patellar tendon moment arms than men. Subsequently the PRF increases by up to 20% for the same knee extending moment which would explain the somewhat higher frequency of patello-femoral disorders in females (44).

Denham and Bishop have shown that the patello-femoral reaction force exceeds the tibio-femoral reaction force in angles above 25°. At near full knee flexion these values rose to almost 150% of the forces passing through the tibio-femoral joint (14). It is therefore not surprising that the patello-femoral articulation is covered by a deep and deformable layer of hyaline cartilage. With 4 to 6 mm in depth, this cartilaginous cover is the thickest to be found in the body, and designed to protect the richly innervated subchondral bone in such a way that the pain threshold is not surpassed (19).

Articular cartilage carries viscoelastic properties enabling it to adapt to the changing surface contours whilst the patella moves along its irregular pathway. At the same time surface deformation under load will lead to a subsequent increase in pressure transmitting area. This process is time dependent and pressure values will thus be different for short-term and long-term loading. The above

mentioned values concerning patello-femoral pressure should hence be regarded as reference values only, as pressure transmitting areas of the patello-femoral joint increase with increasing load and duration of loading (5,22,28,39,55). This may explain why peak stresses of up to 20 times body weight can be tolerated without causing lasting damage as they are applied over a relatively large area for only very short periods of time, whilst long term application of such loads would invariably lead to cartilage breakdown.

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#### REFERENCES

1. **Abraham W, Buchanan JR, Daubert H, Greer III RB, Keefer J.** Should the patella be resurfaced in total knee arthroplasty? Efficacy of patellar resurfacing. *Clin Orthop Relat Res* 1988; 236: 128-134.
2. **Aglietti P, Insall JN, Walker PS, Trent PS.** A new patella prosthesis: Design and application. *Clin Orthop Relat Res* 1975; 107: 175-187.
3. **Aglietti P, Buzzi R, Insall JN.** Disorders of the patellofemoral joint. In: Insall JN, Windsor RE, Scott N, Kelly MA, Aglietti P (eds). *Surgery of the Knee*. 2<sup>nd</sup> ed, Churchill Livingstone, New York, 1993, pp 241-385.
4. **Andriacchi TP, Andersson GBJ, Fremier RW, Stern D, Galante JO.** A study of lower limb mechanics during stair climbing. *J Bone Joint Surg* 1980; 62-A: 749-757.
5. **Bandi W.** [Chondromalacia patellae and arthritis of the patello-femoral joint.] (in German). *Helv Chir Acta*, 1972; Suppl. 11: 1-70.
6. **Bandi W.** *Die retropatellaren Kniegelenksschäden*. [Retropatellar derangement of the knee], 2<sup>nd</sup> ed, Verlag Hans Huber, Bern, 1982. (In German).
7. **Bousquet G, Dejour H.** [The anatomical and physiological basis in the conception and development of knee arthroplasty.] (in French). *Acta Orthop Belg* 1973; 39: 40-55.
8. **Boyd AD, Ewald FC, Thomas WH, Poss R, Sledge CB.** Long-term complications after total knee arthroplasty with or without resurfacing of the patella. *J Bone Joint Surg* 1993; 75-A: 674-681.

9. **Bresler B, Frankel JP.** The forces and moment in the leg during level walking. *Trans Am Society of Medical Engineers* 1950 ; 72 : 27-36.
10. **Burckhardt H.** [Development of loose bodies and mechanics of the knee joint.] (in German). *Brunns Beitr Klin Chir* 1924 ; 130 : 163-214.
11. **Burnett RSJ, Boone JL, McCarthy KP, Rosenzweig S, Barrack RL.** A prospective randomised clinical trial of patellar resurfacing and nonresurfacing in bilateral TKA. *Clin Orthop Relat Res* 2007 ; 464 : 65-72.
12. **Collier JP, McNamara JL, Surprenant VA, Jensen RE, Surprenant HP.** All-polyethylene components are not the answer. *Clin Orthop Relat Res* 1991 ; 273 : 198-203.
13. **Dahlqvist NJ, Mayo P, Seedhom BB.** Forces during squatting and rising from a deep squat. *Eng Med* 1982 ; 11 : 69-76.
14. **Denham RA, Bishop RED.** Mechanics of the knee and problems in reconstructive surgery. *J Bone Joint Surg* 1978 ; 60-B : 345-352.
15. **Dickson Jr. TB.** Preventing overuse cycling injuries. *Phys Sportsmed* 1985 ; 13 : 116-123.
16. **Ellis MI, Seedhom BB, Amis AA, Dowson D, Wright V.** Forces in the knee joint whilst rising from normal and motorised chairs. *Eng Med* 1979 ; 8 : 33-40.
17. **Ericson MO, Nisell R.** Patellofemoral joint forces during ergometer cycling. *Phys Ther* 1987 ; 67 : 1365-1369.
18. **Ficat P.** *Pathologie Fémoro-Patellaire.* [Pathology of the patello-femoral joint], Masson & Cie, Paris, 1970.
19. **Ficat P, Hungerford DS.** *Disorders of the Patello-femoral Joint.* Williams & Wilkins, Baltimore, 1977.
20. **Fick R.** *Handbuch der Anatomie und Mechanik der Gelenke.* Vol. III : Spezielle Gelenk- und Muskelmechanik. Gustav Fischer Verlag, Jena, 1911.
21. **Furmaier A.** [Contribution to the mechanics of patella and knee joint.] (in German). *Arch Orthop Unfallchir* 1953 ; 46 : 78-90.
22. **Goodfellow JW, Hungerford DS, Zindel M.** Patellofemoral joint mechanics and pathology. *J Bone Joint Surg* 1976 ; 58-B : 287-290.
23. **Goodsir J.** On the mechanism of the knee joint. In : Turner W (ed). *The Anatomical Memoirs of John Goodsir.* Vol. 2. Adams and Charles Black, Edinburgh, 1868, pp 231-245.
24. **Goymann V, Haasters J, Heller W.** [Current studies on the biomechanics of the patella.] (in German). *Z Orthop* 1974 ; 112 : 623-625.
25. **Goymann V, Müller HG.** New calculations of the biomechanics of the patellofemoral joint and its clinical significance. In : Ingwersen OS, Van Linge B, Van Rens ThJG, Rösingh GE, Veraart BEEMJ, Le Vay D (eds). *The Knee Joint : Recent Advances in Basic Research and Clinical Aspects.* American Elsevier Publishing, New York, 1974, pp 16-21.
26. **Haxton H.** The function of the patella and the effects of its excision. *Surg Gyn Obst* 1945 ; 80 : 389-395.
27. **Hehne HJ.** *Das Patellofemoralgelenk.* [The patellofemoral joint], Enke, Stuttgart, 1983.
28. **Huberti HH, Hayes WC.** Patellofemoral contact pressures : The influence of Q-angle and tendofemoral contact. *J Bone Joint Surg* 1984 ; 66-A : 715-724.
29. **Hungerford DS, Goodfellow JW.** [Patello-femoral contact areas and their relationship to chondromalacia.] (in German). *Z Orthop* 1975 ; 113 : 784-786.
30. **Insall JN, Goldberg V, Salvati E.** Recurrent dislocation of the high-riding patella. *Clin Orthop Relat Res* 1972 ; 88 : 67-69.
31. **Kaltwasser P, Uematsu O, Walker PS.** The patellofemoral joint in total knee replacement. *Trans Orthop Res Soc* 1987 ; 12 : 292.
32. **Kaufman KR, An KN, Litchy WJ, Morrey BF, Chao EYS.** Dynamic joint forces during knee isokinetic exercise. *Am J Sports Med* 1991 ; 19 : 305-316.
33. **Kelley DL, Dainis A, Wood GK.** Mechanics and muscular dynamics of rising from a seated position. In : Komi PV (ed). *International Series on Biomechanics. Proceedings of the 5<sup>th</sup> International Congress on Biomechanics.* Biomechanics V-B. University Park Press, Baltimore, 1976, pp 127-134.
34. **Krackow KA, Hungerford DS.** Anatomy and kinematics of the normal knee. In : Hungerford DS, Krackow KA, Kenna RV (eds). *Total Knee Arthroplasty. A Comprehensive Approach.* Williams & Wilkins, Baltimore, 1984, pp 5-19.
35. **Lancourt JE, Christini JA.** Patella alta and patella infera : The etiological role in patella dislocation, chondromalacia, and apophysitis of the tibial tubercle. *J Bone Joint Surg* 1975 ; 57-A : 1112-1115.
36. **Leszko F, Sharma A, Komistek RD et al.** Comparison of in vivo patellofemoral kinematics for subjects having high-flexion total knee arthroplasty implant with patients having normal knees. *J Arthroplasty* 2010 ; 25 : 398-404.
37. **Maquet P, Van de Berg A, Simonet J.** Femoro-tibial weight bearing areas. *J Bone Joint Surg* 1975 ; 57-A : 766-771.
38. **Maquet P.** *Biomechanics of the Knee.* Springer, Berlin, 1984.
39. **Matthews LS, Sonstegard DA, Henke JA.** Load bearing characteristics of the patello-femoral joint. *Acta Orthop Scand* 1977 ; 48 : 511-516.
40. **Menschik A.** [Knee joint mechanics : Screw-home.] (In German). *Z Orthop* 1975 ; 113 : 388-400.
41. **Meyer H.** [Knee joint mechanics.] (in German). *Arch Anat Physiol Wiss Med* (Müllers Archiv) 1853 ; 497-547.
42. **Moreland JR.** Mechanisms of failure in total knee arthroplasty. *Clin Orthop Relat Res* 1988 ; 226 : 49-64.
43. **Morra EA, Greenwald AS.** Patello-femoral replacement polymer stress during daily activities : A finite element study. Presented at the 56<sup>th</sup> Annual Meeting of the AAOS, Chicago, 2006.
44. **Nisell R.** Mechanics of the knee : A study of joint and muscle load with clinical applications. *Acta Orthop Scand* 1985 ; 56 (Suppl. 216) : 1-42.
45. **Paul JJ.** Force actions transmitted in the knee of normal subjects and by prosthetic joint replacements. In : The

- Medical Engineering Working Party (ed). *Total Knee Replacement*. Mechanical Engineering Publications Limited, London, 1974, pp 126-131.
46. **Perry J, Antonelli D, Ford W.** Analysis of knee-joint forces during flexed-knee stance. *J Bone Joint Surg* 1975 ; 57-A : 961-967.
47. **Reider B, Marshall JL, Ring B.** Patellar tracking. *Clin Orthop Relat Res* 1981 ; 157 : 143-148.
48. **Reilly DT, Martens M.** Experimental analysis of the quadriceps muscle force and patello-femoral joint reaction force for various activities. *Acta Orthop Scand* 1972 ; 43 : 126-137.
49. **Rhoads DD, Nobel PC, Reuben JD, Mahoney OM, Tullos HS.** The effect of femoral component position on patella tracking after total knee arthroplasty. *Clin Orthop Relat Res* 1990 ; 260 : 43-51.
50. **Schindler OS.** Synovial plicae of the knee. *Current Orthopaedics* 2004 ; 18 : 210-219.
51. **Smidt GL.** Biomechanical analysis of knee flexion and extension. *J Biomech* 1973 ; 6 : 79-92.
52. **Smith AJ.** *A study of force on the body in athletic activities with particular reference to jumping*. PhD thesis. England, Leeds, 1972.
53. **Steindler A.** *Kinesiology of the Human Body under Normal and Pathological Conditions*. 2<sup>nd</sup> print. Charles C Thomas, Springfield, 1964.
54. **Townsend PR, Rose RM, Radin EL, Raux P.** The biomechanics of the human patella and its implications for chondromalacia. *J Biomech* 1977 ; 10 : 403-407.
55. **Trent PS.** The patello-femoral joint. In : Walker PS (ed). *Human Joints and their Artificial Replacement*. Charles C. Thomas, Springfield, 1977, pp 181-182.
56. **Van Kampen A, Huiskes R, Blankevoort L, Van Rens ThJG.** The three-dimensional tracking pattern of the patella in the human knee joint and its effects on surgical intervention. In : Müller W, Hackenbroich W (eds). *Surgery and Arthroscopy of the Knee*. Springer, Heidelberg, 1988, pp 434-445.
57. **Van Kampen A, Huiskes R.** The three-dimensional tracking pattern of the human patella. *J Orthop Res* 1990 ; 8 : 372-382.
58. **Veress SA, Lippert FG, Hou MCY, Takamoto T.** Patellar tracking patterns measurement by analytical x-ray photogrammetry. *J Biomech* 1979 ; 12 : 639-650.
59. **Wahrenberg H, Lindbeck L, Ekholm J.** Knee muscular moment, tendon tension force and EMG during a vigorous movement in man. *Scand J Rehab Med* 1978 ; 10 : 99-106.
60. **Walker PS.** *Human Joints and their Artificial Replacement*. Charles C. Thomas, Springfield, 1977.
61. **Walker PS.** Contact areas and load transmission in the knee. In American Academy of Orthopaedic Surgeons : *Symposium on Reconstructive Surgery of the Knee*. Mosby Company, Saint Louis, 1978, pp 26-36.
62. **Winter DA.** Moments of force and mechanical power in jogging. *J Biomech* 1983 ; 16 : 91-97.
63. **Zernicke RF, Garhammer J, Jobe FW.** Human patellar-tendon rupture. A kinetic analysis. *J Bone Joint Surg* 1977 ; 59-A : 179-183.