Femoral articular geometry and patellofemoral stability

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**Article info**

**Abstract**

**Background:** Patellofemoral instability is a major cause of anterior knee pain. The aim of this study was to examine how the medial and lateral stability of the patellofemoral joint in the normal knee changes with knee flexion and measure its relationship to differences in femoral trochlear geometry.

**Methods:** Twelve fresh-frozen cadaveric knees were used. Five components of the quadriceps and the iliotibial band were loaded physiologically with 175 N and 30 N, respectively. The force required to displace the patella 10 mm laterally and medially at 0°, 20°, 30°, 60° and 90° knee flexion was measured. Patellofemoral contact points at these knee flexion angles were marked. The trochlea cartilage geometry at these flexion angles was visualized by Computed Tomography imaging of the femora in air with no overlying tissue. The sulcus, medial and lateral facet angles were measured. The facet angles were measured relative to the posterior condylar datum.

**Results:** The lateral facet slope decreased progressively with flexion from 23° ± 3° (mean ± S.D.) at 0° to 17 ± 5° at 90°. While the medial facet angle increased progressively from 8° ± 8° to 36° ± 9° between 0° and 90°. Patellar lateral stability varied from 96 ± 22 N at 0°, to 77 ± 23 N at 20°, then to 101 ± 27 N at 90° knee flexion. Medial stability varied from 74 ± 20 N at 0° to 170 ± 21 N at 90°.

There were significant correlations between the sulcus angle and the medial facet angle with medial stability ($r = 0.78$, $p < 0.0001$).

**Conclusions:** These results provide objective evidence relating the changes of femoral profile geometry with knee flexion to patellofemoral stability.

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**Keywords:** Patellofemoral joint, Patellofemoral stability, Femoral articular geometry, Biomechanics test

1. Introduction

Patellar “stability” has been quantified according to classical mechanical principles by measuring the magnitude of force that opposes its displacement from its initial position of equilibrium [1,2]. This is stability as an objective mechanical measurement and, although it may be related, it is different from the spectrum of subjective “instability” that presents as a clinical problem. A reduction in objective stability as a result of injury or pathology and the changing stability with flexion may result in clinical patellofemoral instability in a particular part of the flexion cycle. In order to avoid confusion, the reader should note that this paper is concerned solely with objective measurement of mechanical “stability” in-vitro; the clinical usage to describe subjective sensation in-vivo is not the subject of this paper.

Objective data that characterises the factors which govern the stability of the patellofemoral joint helps us understand the behaviour of this joint. The patellofemoral joint relies on both its geometry and its soft tissue stabilisers working in concert so that it...
can function with optimal mobility and stability. The difficulty in understanding this is reflected in the radiographic indices, the various surgical techniques and variable results for correction of patellar instability [3–11].

The stability of the patellofemoral joint at different flexion angles has been described [2] and it is known that the ‘skyline’ profile of the patellar contact area on the femoral groove changes with knee flexion [12]. It has been suggested that the increased stability of the patellofemoral joint with increasing knee flexion is as a result of a deepening sulcus with knee flexion [13]. It has been shown that surgically flattening the lateral condyle can reduce the lateral stability of the patellofemoral joint [14]. Similarly, flattening the anterior trochlea and subsequently deepening it with a trochleoplasty caused decreases and then increases of patellar lateral stability, respectively [15]. A review of clinical data found that the trochlear sulcus angle was related to clinical instability [16]. This evidence suggested the hypothesis that the force needed to displace the patella away from its stable articulation in the trochlear groove would correlate to the slope of the lateral or medial trochlear facet on which it is moving. Thus, the aim of this study was to examine how the medial and lateral stability of the patellofemoral joint in the normal knee changes with knee flexion and to measure its relationship to changes in femoral trochlear geometry.

2. Methods

Twelve fresh-frozen cadaveric knees with no prior history of knee surgery or disease were used in this study (mean age 64 Standard Deviation (S.D. = 15). These were obtained from the International Institute for the Advancement of Medicine (Jessup, PA, USA). The Institute undertook screening and consent for the use of the knees for research. Ethical permission for the study was obtained from the Riverside Research Ethics Committee. The knees were stored at −20 °C and thawed a day prior to experimentation.

The skin and subcutaneous tissue were removed. The deep fascia, retinaculae and iliotibial band (ITB) were preserved. The femur and tibia were cut approximately 20 and 15 cm above and below the knee, respectively. The head of the fibula was transfixed to the tibia by two bone screws to maintain its anatomical position and then the distal part excised. Using polymethylmethacrylate and after preparation of the medullary cavity, an intramedullary sleeve and a rod were cemented into the femur and tibia respectively. The sleeve in the femur was aligned to the central femoral axis by use of rubber spacers and an outrigger alignment rod. A polyethylene socket was cemented into the patella, centred over the median ridge and 10-mm deep to the anterior (superficial) surface. This was taken to be the geometric centre of the patella. The quadriceps was separated into six components: rectus femoris (RF), vastus intermedius (VI), vastus lateralis longus (VLL), vastus lateralis obliquus (VLO), vastus medialis longus (VML), and vastus medialis obliquus (VMO).

The stability rig was composed of two parts. The fixed part was attached to the base of an Instron materials testing machine (Instron Ltd., Buckinghamshire, England) and the moving part was a three-degree of freedom mounting attached to the Instron load cell (Figure 1). The knee was mounted sideways (lateral aspect upwards) in the fixed part of the stability rig by locating the cemented femoral sleeve onto a rod on a femoral mounting device. The knee was aligned such that the anatomic axis of the femur was perpendicular to the load cell axis. The femoral sleeve allowed the rotation of the femur to be adjusted until the most posterior parts of the femoral condyles were aligned vertically in a distal–proximal view; when this was achieved it

Figure 1. The experimental setup. The figure shows a right knee mounted in a rig secured to the Instron base plate with its lateral aspect upwards. The components of the quadriceps were loaded with cables passed over pulleys and attached to weights. The three-degree of freedom mounting is attached to the load cell of the Instron machine at the top and is coupled to the patella by snap fit into a polyethylene patellar socket cemented into the patella. The knee was flexed to the desired angle and a rod was placed anterior to the tibial rod to maintain this position but allow for the tibia to rotate during testing. The femur was rigidly fixed and the load cell moves up and down by 10 mm from the neutral position to test for lateral and medial stability (see inset).
was locked to the mounting device. The lower end of the three-degree of freedom mounting was snap-fitted into the polyethylene patellar socket to form a ball and socket joint (Figure 1). The mounting allowed for patellar mobility in a sagittal plane: anterior–posterior, proximal–distal translations and flexion–extension rotation and the ball–patellar socket joint allowed for unimpeded tilt and rotations to occur when the patella was displaced medially and laterally.

The components of the quadriceps were each loaded with hanging weights using cables and pulleys. A total load of 175 N was applied to the quadriceps groups [1] and 30 N was applied to the iliotibial band [17,18]. This was done according to the directions and physiological cross-sectional areas of the muscles [19] relative to the femoral axis: VLL 14° lateral and 0° anterior; VLO 35° lateral and 33° posterior; VML 15° medial and 0° anterior; VMO 47° medial and 44° posterior; RF + VI 0° lateral and 0° anterior and ITB 0° lateral and six degrees posterior. The quadriceps tension distribution was: RF + VI 35%, VLL 33%, VLO nine percent, VML 14%, and VMO nine percent.

The knee was tested at 0°, 20°, 30°, 60° and 90° of tibiofemoral flexion. It was flexed against the muscle tension and at the chosen test angle a vertical rod was placed anterior to the tibial rod to block knee extension. At each test angle, the patella was displaced cyclically 10 mm laterally and medially at 100 mm/min from its stable neutral position. The fourth load versus displacement curve was recorded. The mechanical stability of the patella at each flexion angle and test condition was determined.

The medial stability was defined as the ‘restraining force’ recorded by the load cell when the patella was displaced medially by 10 mm and similarly, lateral stability was the ‘restraining force’ required to displace the patella 10 mm laterally from its stable neutral position [2]. The ‘restraining force’ is a measure of the resultant force of all the tissues acting on the patella, which resists its movement away from the position of equilibrium.

At the conclusion of testing while the knee was still loaded within the rig, the contact of the patella to the femur at the tested joint angles was determined visually through a lateral parapatellar incision. Pins were embedded on the side of the lateral condyle to mark these points. The soft tissues around the femur were removed. Care was taken not to damage the articular cartilage and to keep it moist with normal saline. Computed Tomography (CT) scans were performed on the femoral specimens. The image files were loaded into a three-dimensional imaging software to reconstruct and orientate a 3D model of the femur (Figure 2). The femur was aligned with the anatomical axis horizontal and perpendicular to the axial plane and the posterior condyles parallel to the sagittal plane. This is analogous to the orientation of the femur in the testing rig, viewed end-on from distal to proximal. The femur was then virtually flexed until the pin which corresponded to the patellofemoral contact of the flexion angle of interest was uppermost (most anterior) in the

Figure 2. Images files were loaded into a three-dimensional imaging software to reconstruct and orientate a 3D model of the femur.
sagittal image window of the software. In this manner, tangential images of the ‘skyline’ profile of the femoral trochlear groove for the patellofemoral contact at 0°, 20°, 30°, 60° and 90° knee flexion were obtained. Since the CT scan of the femur was performed in air with no overlying tissue, the articular contour could be visualized and measurements made. Using these images, the sulcus angle, medial and lateral facet angles for each of the flexion angles were measured (Figure 3). The sulcus angle was defined as the angle that subtended the deepest part of the trochlear groove and the most superficial points of the lateral and medial condyles in that particular profile image of the femur. The lateral or medial facet slope was the slope of the facet relative to the posterior condylar datum in the plane of the ‘skyline’ section. The medial limb of the sulcus angle at the 0° contact point on the medial aspect was usually the femoral bone above the medial condylar articular surface. This is because the distal pole of the patella articulates with the proximal limit of the lateral articular surface on the lateral condyle but is still non-articular on the medial side.

2.1. Statistical analysis

The distributions of the data were checked for normality. Changes of either patellar restraining force or trochlear geometry were examined, across the range of flexion angles, using one-way analyses of variance (ANOVA), with Bonferroni post-tests to calculate levels of significance at each angle of knee flexion. Relationships between patellar restraining force and local trochlear facet slope were examined using the Pearson’s correlation coefficient. Statistical significance was taken to be p < 0.05. In order to examine the tendency for the patella to dislocate while the knee was flexing, in relation to the trochlear geometry, the Pearson correlation was used to relate the change in restraining force to the change in facet angle, both medial and lateral directions, between successive angles of knee flexion.

3. Results

The normality tests indicated a good approximation to normal distribution for all data. The sulcus angle did not change significantly (p = 0.09), that is from anteroproximal to posterodistal, varying from 155° ± 7° (mean ± S.D.) at 0° flexion, to 147° ± 10° at 90° flexion (Figure 3). The lateral facet slope decreased progressively from 23° ± 3° at 0° flexion to 17 ± 5° at 90° (p = 0.0003). The post hoc tests showed that the lateral facet slope was significantly larger at 0° than at 30° to 90°. The medial facet angle increased progressively from 8° ± 8° to 36° ± 9° between 0° and 90° knee flexion (p < 0.0001; Figure 4).

Patellar lateral stability, expressed as restraining force at 10 mm lateral displacement, varied from 96 ± 22 N at 0° knee flexion, to 77 ± 23 N at 20° knee flexion, then to 101 ± 27 N at 90° (p < 0.01 for 20° versus 0° and 90°; p < 0.05 for 30° versus 0° and 90°; Figure 4). When displaced medially, the restraining force increased (p < 0.0001) with knee flexion, from 74 ± 20 N at 0° flexion to 170 ± 21 N at 90° flexion (Figure 5). For medial stability the change with knee flexion angle was much larger than for lateral stability. This pattern (Figure 5) reflects the variations of their respective facet angles, with the medial facet angle changing more than the lateral facet angle (Figure 4).

There were significant correlations between the sulcus angle and the medial facet angle with medial stability. As the sulcus angle deepened (smaller sulcus angle) the medial stability increased (r = −0.45, p = 0.0003) (Figure 6). As the slope of the medial facet increased the force required to displace the patella medially increased (r = 0.78, p < 0.0001; Figure 7). The lateral restraining force did not correlate significantly with the sulcus angle (r = −0.24, p = 0.06), but did correlate with the lateral facet angle (r = 0.28, p = 0.03; Figures 5 & 6).

The change in medial restraining force between successive knee flexion angles did not correlate significantly with change in the medial facet angle (p = 0.6). However, there was a significant correlation (r = 0.48, p = 0.001) between the change in lateral restraining force and the change in the lateral trochlear facet angle, between successive knee flexion angles (Figure 8).

![Figure 3. The lateral, medial facet angles and sulcus angle for each of the flexion angles were measured.](image-url)
Figure 4. The sulcus angle (above) and the medial and lateral facet angle (below) at the tested knee angles (mean ± S.D., n = 12). The error bars for the facet angles are shown in one direction only for clarity.

Figure 5. The restraining force for 10 mm medial and lateral displacement versus knee flexion angle (mean ± S.D., n = 12).
4. Discussion

Instability and malalignment of the patellofemoral joint are major causes of knee pain in young people [20–22] and are among the dominant causes of dissatisfaction following knee arthroplasty [23–25]. Instability or stability of this mechanism is dependent upon a complex interaction of various structures. The extensor mechanism, the retinacular restraints, the articular geometry and limb alignment are amongst these factors, but their individual contributions are difficult to quantify because of this interaction. For example, there is a dynamising effect of the vastus medialis obliquus muscle on the medial patellofemoral ligament [17,18].

This study presents data to demonstrate the effects of the geometry of the profile of the femoral trochlea on the stability of the patellofemoral joint in various knee flexion angles. It was hypothesized that the critical angle would be the individual slopes of the facets as this would be the slope in which the patella must climb when it is forcibly displaced from its stable position. The results support this view for the medial side, with a strong correlation between the slope of the medial facet and medial stability. On the lateral side, the clinical observation that dislocation occurs early in flexion was borne out by the significant drop in stability seen between full extension and 20° of flexion where the facet angle is lower. However, because of the small range of lateral trochlear facet angles, the correlation with lateral restraining force was weak. The medial facet increased by four times from eight degrees to 36° in the flexion range tested, whereas the lateral facet angle decreased by only approximately a third, from 23° to 17°. Generally, the graph of facet angles (Figure 4) and restraining forces (Figure 5) with knee flexion angles share a similar pattern for the medial and lateral sides.

As the knee flexes, the angle between the quadriceps pull and the patellar tendon in the sagittal plane closes and so the resultant vector of the patellofemoral joint reaction force increases [26]. This pulls the patella against the trochlea, making it

![Figure 6. Medial stability versus sulcus angle (below) (r = −0.45, p = 0.0003) and lateral stability versus sulcus angle (above) (r = −0.24, p = 0.06).](image_url)
more difficult to displace the patella from side to side. This would cause the medial and lateral restraining force to increase as the knee is progressively flexed. On the medial side, data from this study is consistent with this and is in agreement with previous work [2]. For the lateral side, the increase in stability from 20° to 0° flexion provides valuable insight into the clinical problem of patellar instability. This change in stability at near full extension has been ascribed to the medial retinaculum tightening in the last 20° of extension, since its disruption caused the most decrease in lateral stability in this range of flexion [14,27]. After 20°, the lateral stability only increases moderately taking into consideration the increasing patellofemoral joint reaction force described earlier. This may be because there is an opposing effect of the reduction in lateral facet angle as the knee is flexed. This decreasing pattern with knee flexion has been previously recognised in a more detailed anatomic study using the epicondylar axis as the datum of measurement [19].

The weak correlation between the lateral facet angle and the lateral restraining force implies that for patellar lateral displacement the soft tissues are a more dominant factor or that the interaction between the two is more complicated. There was a stronger correlation between the change in the lateral stability versus the change in the lateral facet angle from one joint angle to the next (Figure 8): for every degree reduction in lateral facet angle, there was three newton reduction in lateral restraining force from one joint angle to the next. A previous study investigated the effect on lateral joint stability when the lateral femoral condyle was flattened with an osteotomy [14]. The average lateral facet slope at the contact zone at 20° knee flexion was 22° (S.D. = 5) and this was flattened to 0°. On average the reduction in lateral stability was between 50 and 60 N. Our findings agree with this, confirming that the lateral trochlea plays a major role in restraining the patella in full extension. This observation has practical implications in trochleoplasty and prosthetic design.
This study has limitations inherent with in vitro work. The knees were obtained from elderly individuals who were free from clinical disease and although good cartilage thickness and the absence of gross arthritic changes were verified in our imaging studies, some degree of age related changes in the joints and soft tissues is unavoidable. Moreover, although care was taken to load the quadriceps muscle according to physiological directions, this is a necessary simplification of the situation in vivo. Much higher muscle forces are present in-vivo but loading of the individual components necessitated lower tensions. From experience, higher tensions caused tearing or avulsion of the muscle. The muscle tension in the experiment was also constant at all angles tested and this is not the case in the natural knee as higher tensions are generated at higher flexion angles in weight-bearing activity.

These results provide objective evidence to support the hypothesis relating the changes of cartilaginous femoral profile of the trochlea in different parts of knee flexion to patellofemoral stability. They go some way to explaining the vulnerability of the patellofemoral joint near full extension, and suggest that variability of patellofemoral stability may at least in part be due to individual variation in femoral geometry. These objective mechanical studies give substantially more insight into the complexities of patellofemoral function than inference from purely anatomic studies and reinforce the need for all elements of the mechanism to be measured when dealing with the subjective, clinical symptoms of instability of this challenging mechanism.

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References


