The influence of malrotation of the femoral component in total knee replacement on the mechanics of patellofemoral contact during gait

AN IN VITRO BIOMECHANICAL STUDY

C. Verlinden, P. Uvin, L. Labey, J. P. Luyckx, J. Bellemans, H. Vandenneucker

From University Hospital Leuven, Lubbeek, Belgium

Malrotation of the femoral component is a cause of patellofemoral maltracking after total knee arthroplasty. Its precise effect on the patellofemoral mechanics has not been well quantified. We have developed an in vitro method to measure the influence of patellar maltracking on contact. Maltracking was induced by progressively rotating the femoral component either internally or externally. The contact mechanics were analysed using Tekscan. The results showed that excessive malrotation of the femoral component, both internally and externally, had a significant influence on the mechanics of contact. The contact area decreased with progressive maltracking, with a concomitant increase in contact pressure. The amount of contact area that carries more than the yield stress of ultra-high molecular weight polyethylene significantly increases with progressive maltracking. It is likely that the elevated pressures noted in malrotation could cause accelerated and excessive wear of the patellar button.

Despite the overall high success rates of total knee replacement (TKR), complications involving the patellofemoral joint remain an important cause of persistent pain and failure, and may require revision surgery. Clinical studies have shown that altered kinematics and increased contact pressure at the patellofemoral joint contribute to complications and failure.

The increase in patellofemoral contact pressures after TKR with patellar resurfacing has been documented in several in vivo and cadaver studies. It has been reported that these pressures, although measured under non-physiological loads, might even exceed the yield strength of the ultra-high molecular weight polyethylene (UHMWPE) patellar component.

Many authors have analysed the factors which influence patellar (mal)tracking, resulting in altered patellofemoral kinematics and excessive loads. Malrotation of the femoral component is the major cause of maltracking and can occur with the most commonly used surgical techniques.

It is therefore possible that femoral malrotation negatively influences the area of patellofemoral contact and contact pressures, potentially endangering the integrity of the prosthesis, but we are not aware of any data documenting and quantifying this effect. The aim of this study was to investigate the effect of malrotation of the femoral component on the patellofemoral contact area and pressure in total knee arthroplasty by employing in vitro simulation of a full gait cycle using the latest design of knee simulator. We wished to address the following questions: Does malrotation of the femoral component alter the position of patellar contact? Does this result in a higher contact pressure and a reduced area of contact? Does femoral malrotation induce a risk concerning the long-term survival of the prosthetic components? Is excessive femoral external rotation potentially less harmful than excessive internal rotation, as frequently supposed in the literature?

Materials and Methods

A right Genesis II Oxinium femoral component, size 5 (Smith & Nephew, Memphis, Tennessee), was cemented on to a specially designed fixture. A matching 32 mm diameter patellar button of the onset type was fixed on to a steel holder using bone cement. The femoral fixture and the patellar holder both fitted into the stations of a Prosim knee joint wear simulator (Simulation Solutions, Stockport, United Kingdom).

During the experiments the patella was covered with a pressure-sensitive film (K-Scan 4000, pixel size 1.613 mm², measurement range 9000 PSI (Tekscan, South Boston, Massachusetts)), using double-sided tape. Care was taken to make sure that the pressure sensor covered the entire area of contact of the patellar button and did not
move with respect to the patella during a simulated gait cycle. For all tests the same patellar button and Tekscan sensor were used, and the sensor remained attached to the patella continuously. The sensor was calibrated and conditioned according to the manufacturer’s guidelines.

The femoral fixture (Fig. 1) was specially designed for these experiments. It allows positioning of the femoral component in seven different angles of axial rotation: neutral alignment (with the femoral component parallel to the epicondylar axis); 2.5° of internal or external rotation with respect to neutral alignment; 5° of internal or external rotation; and 7.5° of internal or external rotation. A test of the accuracy and repeatability of the device showed that it was accurate to within 1.1° and precise to within 1.3°.

The knee joint wear simulator was used to simulate load and movement between the patella and the femur during gait. The flexion-extension, distal-proximal and rotational movements of the patella were derived from Jenny et al.⁹ and Hsu et al.¹⁹ who showed how it moves as a function of the angle of knee flexion. For these three degrees of freedom, displacement control was applied. The contact force between the femoral component and the patellar insert was taken from Ward and Powers.²⁰ The mediolateral constraints on the normal patella were simulated by two linear compression springs, with a spring constant of 8 N/mm, that resisted movement along this axis (Fig. 1). Complete blocking of mediolateral movement was also simulated as a worst-case of maltracking. Tilting of the patella was locked. The input curves for the flexion angle, rotation and contact force as a function of percentage of the gait cycle are shown in Figure 2.

After fixation of the components, they were mounted in the wear simulator. The patella and femur were aligned mediolaterally with the femoral component in neutral orientation and fully extended. The patella was then allowed to slide freely into the trochlear groove by manually applying a patellofemoral contact force. In this configuration, the compression springs were mounted in such a way that they did not exert a net force on the femur. This assembly was then kept for all other tests with the malrotated femoral components.

The location of patellar contact, its area and the contact pressure were measured dynamically for each configuration during 20 consecutive gait cycles at a rate of 100 frames per second, resulting in a total of 2000 frames for each angle of rotation. The maximum area of contact was determined as the total area in mm² over the entire patella where pressure was recorded during peak loading. The contact pressure was defined as the average pressure recorded on the entire patella at peak load. Data concerning the contact area, the average contact pressure, the location of the centre of pressure and the contact force were gathered and exported as text files. Microsoft Office Excel 2007 (Microsoft, Redmond, Washington) was used to analyse and process the data. One-way analysis of variance (ANOVA) with Bonferroni-Dunn correction was used to evaluate the significance of the measured differences. The Tekscan sensor underwent a total of 280 cycles. After completion of the experiment, a series of tests were carried out with a total of 50 cycles in different configurations which all showed reproducible results to ensure that no inaccuracy or damage of the sensor due to shear stress or other factors had occurred during the experiment.

Fig. 1
Lateral and frontal view of the experimental arrangement with the femoral component mounting piece (top), which allowed correct and reproducible rotation of the femoral component, and the patellar button (below). All degrees of freedom are indicated; tilt was locked. Mediolateral soft tissue restraints were mimicked by linear springs (ML, mediolateral; DP, distal-proximal; PFJRF, patellofemoral joint reaction force).

Fig. 2
Input curves for the wear simulator.
Results

Differences in the location of patellar contact between the neutral position and both internal and external rotation were noted for each malrotation configuration (Fig. 3). During the entire cycle the centre of force, which is a good indicator of contact location, shifted laterally for each configuration of internal malrotation, whereas it moved medially for each configuration of excessive external rotation. The contact area was maximal in neutral configuration. Malrotation of the femoral component, in both internal and external configurations, induced a smaller patellofemoral contact area. The experimental arrangement with inhibited mediolateral movement of the patella revealed lower values in the contact area, with the same tendency towards lower contact areas in the malrotated configurations. This is illustrated in Figure 4 and Table I. The patellofemoral contact pressure was minimal in the neutral configuration. Malrotating the femoral component either internally or externally, progressively increased the pressure. Both experimental arrangements revealed the same tendency, with higher values in the more excessive rotational configurations with blocked mediolateral movement (Fig. 5, Table II). The level of significance of the differences in contact area and pressure between all the malrotation configurations in the experiments utilising mediolateral springs, is shown in Table III. Where mediolateral movement was blocked, measurements revealed a significant difference between every malrotation configuration and the neutral arrangement, but showed that the differences between the progressive malrotation configurations were not significant.

The contact pressure for each configuration of malrotation (Table II), is an average value over the entire contact area at peak loading. Tekscan recordings (Fig. 3) and evidence from the literature show that the contact pressure is not spread homogeneously over the patella but has focal areas of higher pressure. In order to quantify these differ-
ences and their evolution with progressive malrotation, we analysed the extent of the area that carries more than a certain threshold pressure, which we defined as 20 MPa. In the experiment where the mediolateral patellar movement was controlled with springs, the relative amount of highly pressurised area significantly increased with progressive malrotation (p < 0.001). When the mediolateral movement of the patella was blocked, malrotation did not initially increase this amount. However, with 5° or more of malrotation there was a rapid significant increase (p < 0.001). Figure 6 and Table IV illustrate the evolution of the percentage of the total contact area carrying more than 20 MPa.

**Discussion**

All our experiments were carried out with the same femoral component (Genesis II, Smith & Nephew). It is highly likely that experiments using components with different designs could show other results.

Although Fuchs et al\(^2\) were not able to demonstrate that malrotation of the femoral component influences patellar tracking, we could clearly show a significant influence on patellofemoral contact area and pressure. The patellofemoral contact area decreases after TKA, with a concomitant increase in contact pressure.\(^2\)\(^,\)\(^2\)\(^2\) An altered pattern of patellofemoral kinematics, as is seen in maltracking due to malrotation of the femoral component, can lead to an even smaller contact area and hence to higher patellofemoral contact pressure. This study is to our knowledge the first to document and quantify this.

In our experiment the mounting piece allowed malrotation of the femoral component to up to 7.5°. We tested the reproducibility of adjusting the angle of malrotation on the mounting piece and found an average deviation of 0.8°.

However, in vivo the position of the patella and its tracking are not determined exclusively by rotation of the femoral component. The presence of restraining collateral ligaments is highly important and must therefore be introduced in any study concerning patellar tracking. However, there has been no study of the exact values of these restraining mediolateral forces. In order to mimic these soft-tissue constraints, we used a pair of springs (stiffness k = 8200 N/m) positioned at each side of the patella. In order to compare our data we repeated the experiment when the patella was fixed and the mediolateral movement blocked, but as this constraining force was independent of the angle of flexion and varies dynamically in vivo with the degree of flexion, our investigation was not a true copy of an in vivo gait cycle. However, owing to the dynamic nature of the soft-tissue restraining forces, it is probable that at any point of an in vivo gait cycle the contact characteristics would be within the range of our two experimental set-ups.

In our study, tilting of the patella was blocked, whereas it is known that changes in patellar tilt alter with a malrotated femoral component,\(^2\)\(^2\) but it is not clear whether the changes in the angle of patellar tilt alter the patellofemoral contact pressure. A similar experiment with a modified
Patellar tilt might give different results. If the patella were able to tilt in conformity with the trochlear groove in a malrotated configuration, a lower contact pressure might result. However, if patella tilt had occurred during our experiment there would still be an increase in pressure with a malrotated femoral component, given that recent work by Besier et al.\(^{23}\) indicates that even in vivo with normal patellar tilting, malrotation induces a smaller contact area.

Most measurements concerning the contact area and pressure in relation to the yield strength of UHMWPE have been carried out in the tibiofemoral joint. We believe that, because of its clinical importance, it is also necessary to give attention to the dynamics and kinematics of the patellofemoral joint. An increase in contact pressure at the patellofemoral joint after TKR may lead to greater wear of UHMWPE, because this augmented contact pressure, defined as the patellofemoral joint reaction force divided by the area of contact between the patella and the surface of the trochlear, is the primary cause of wear, failure and loosening of the patellar component.\(^{6,9,12}\)

Although the pressure recorded did not reach the 20 MPa threshold during the normal gait cycle, the UHMWPE patellar button could still be at risk. The contact pressures we describe in Table II are average values, resulting from a force distribution over the whole area of patellar contact. Our Tekscan data revealed that certain parts of the patellofemoral contact area carry a greater load, and therefore experience greater pressures, which do exceed the yield stress of UHMWPE. If we take into account the fact that in vivo the pressures are most likely to be within the range between our two experimental set-ups, the results clearly demonstrate the possible danger of a malrotated femoral component. Although the total contact area decreases with progressive malrotation, the percentage of the area which is highly loaded rapidly increases. Severe mal-tracking does not generate an average contact pressure above the yield strength, but large parts of the contact area experience higher loads, resulting in more than 25% of the area, where the pressure easily exceeds 20 MPa, being at risk. Excessive contact pressure is the main cause of component wear.

<table>
<thead>
<tr>
<th>Table III</th>
<th>P values indicating the level of significance of the measured differences between configurations of malrotation in the experiment using medial and lateral springs. The right side of the diagonal line contains data concerning contact area; the left side contains data concerning contact pressure ((p &lt; 0.001))</th>
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<tbody>
<tr>
<td></td>
<td>Internal rotation</td>
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<tr>
<td>ML* springs</td>
<td>7.5</td>
</tr>
<tr>
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<td>0.81</td>
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<tr>
<td>7.5</td>
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* ML, mediolateral

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wear, and therefore it is possible that patellar maltracking induces focal accelerated wear of the patellar button, which can present clinically as instability or pain and may result in loosening or failure of the component. In the lower ranges of flexion, as in level walking, the lower patellofemoral forces, the more central contact areas and the clinical results all indicate that failure of the UHMWPE would not commonly occur in this range of movement. However, in the higher flexion ranges, seen in ascending and descending stairs, rising from a chair and squatting, are more likely to be damaging loads to the patellar component. Although the lower ranges of flexion would be more frequently used, the average activities of daily life would result in a combination of both lower and higher ranges. In future, the patellofemoral joint must be included in biomechanical studies of the knee to examine the long-term effect on wear of elevated contact pressures in the joint.

In our investigation we only studied contact pressures. Shear stress was not assessed. However, as we applied the same axial load in every malrotation configuration with the same friction coefficient, the elevated contact pressures which we noted with malrotation also imply an increase in shear stress.

The results show that both internal and external malrotation of the femur progressively reduce the contact area. As a consequence, the contact pressure in the patellofemoral joint gradually increases with increasing malrotation. Although there is a relative increase in pressure of 2.5% between the neutral configuration and the most severe angles of malrotation, when mediolateral movement was allowed, and of over 50% in the fixed mediolateral position, these average pressures never exceed the yield strength of UHMWPE. Our study simulated a flexion arc between 0° and 60° as in a normal gait cycle, and by using the same input curves for every configuration, did not take into account a possible increase in patellofemoral load due to malrotation. Takeuchi et al. have shown that contact pressures at angles of flexion > 60° can exceed the compressive yield strength of UHMWPE, with up to more than 40 MPa at angles above 60°.

Maltracking of the patella after TKA in vitro can result from an error in rotational alignment of the femoral component during surgery. Traditionally, internal rotation has always been considered to be more harmful than excessive external rotation. Our results indicate that both internally and externally malrotated femoral components generate significantly increased contact pressures in the patellofemoral joint. The main factor in achieving a correct pattern of patellar tracking, even with the aid of navigation systems, is the ability of the surgeon to determine the rotation of the femoral component correctly.