Anteroposterior positioning of the tibial component and its effect on the mechanics of patellofemoral contact

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ABSTRACT:

The biomechanics of the patellofemoral joint can become disturbed during total knee replacement by alterations induced by the position and shape of the different prosthetic components. The role of the patella and femoral trochlea has been well studied. We have examined the effect of anterior or posterior positioning of the tibial component on the mechanisms of patellofemoral contact in total knee replacement. The hypothesis was that placing the tibial component more posteriorly would reduce patellofemoral contact stress while providing a more efficient lever arm during extension of the knee.

We studied five different positions of the tibial component using a six degrees of freedom dynamic knee simulator system based on the Oxford rig, while simulating an active knee squat under physiological loading conditions. The patellofemoral contact force decreased at a mean of 2.2% for every millimetre of posterior translation of the tibial component. Anterior positions of the tibial component were associated with elevation of the patellofemoral joint pressure, which was particularly marked in flexion > 90°.

From our results we believe that more posterior positioning of the tibial component in total knee replacement would be beneficial to the patellofemoral joint.

Introduction

Anterior knee pain remains a common complaint after total knee replacement (TKR). Studies have reported an incidence of anterior knee pain in TKR both with and without patellar resurfacing of about 6.2% and 25%, respectively.[1,2] Factors such as the position of the trochlear groove, the rotational position of the femoral and tibial components and the position of the joint line have an important influence on patellofemoral joint biomechanics.[3,4] However, little is known about the influence of anteroposterior positioning of the tibial tray.

From a biomechanical point of view, a more posterior position of the tibial component would theoretically lead to an improved lever mechanism for extension of the knee, and might reduce patellofemoral contact stress (Fig. 1⇓).

Posterior translation of the tibial component should be comparable to the effect obtained by the Fulkerson and Maquet procedures,[4,5] where anterior displacement of the tibial tuberosity improves the mechanical advantage of the quadriceps and reduces the force in the patellofemoral joint. Hence, these procedures are sometimes used for the treatment of refractory anterior knee pain.[3,6,7]

In TKR where the tibial dimensions fall between two component sizes, the surgeon has the option to select the smaller baseplate and place it some distance away from the anterior margin of the tibia, thereby establishing a more posterior position. The tibia is shifted anteriorly relative to the prosthetic components, providing a ‘Maquet’-type effect on the tibial tuberosity and extensor mechanism thereby unloading the patellofemoral joint.

The recent availability of validated dynamic knee simulators, as well as advances in the analysis of contact force and contact area, have made it possible to quantify such an effect.

In this study we used a third-generation, custom-made, dynamic Oxford knee simulator under physiological loading conditions during a deep knee squat, to study the influence of the position of the tibial component on the patellofemoral contact force, contact area and contact pressure. Our hypothesis was that a more posterior placement of the component would reduce patellofemoral contact stress while providing a more efficient lever arm during extension of the knee.

Material and methods

A dynamic knee simulator system based on the Oxford rig was customized and adapted for this study to simulate and record the dynamic loads during a single knee squat under physiologic loading conditions as described below (Fig. 2). The simulator consists of two custom rigid fixtures representing an upper and a lower leg which are mounted on a frame with a hip and ankle assembly. Two actuators control the motion and loads: one moving the hip, and the other pulling the quadriceps. The quadriceps actuator is positioned on the upper leg in a way, that reproduces its anatomical location with respect to the moment arms around the knee. Sensors placed in line with the actuators detect the quadriceps and ankle forces, and the hip height relative to the ankle.

The knee simulator is designed with six degrees of freedom: two degrees of freedom for movement in the hip joint (vertical translation and rotation around the transversal axis) and four degrees of freedom for movement in the ankle joint in order to reproduce normal knee kinematics.[7]

A real-time data acquisition and closed feedback system (Labview, National Instruments, TX, USA) is used to perform a squat with a certain hip velocity (given as a function of time) while simultaneously applying a quadriceps force on the knee to induce a vertical

ankle force. Error feedback from the 6-axis ankle load cell is used to control the quadriceps actuator.

For this research project, the tibia and femur were covered with the components of a posterior-stabilized total knee prosthesis (Genesis II, Smith & Nephew, Memphis, TN, USA). A size 5 femoral component was used with a size 4 tibial component, size 3-4, 9 mm thick insert and a 32 mm resurfacing patella. These intermediate implant size combinations are currently used clinically. A Ø 6.4 mm stainless steel cable (Sanlo, IN, USA) simulated the patellar and quadriceps tendon.

K-Scan 4000/9000 psi sensors (Tekscan, MA, USA) were used to measure contact area, contact force and retropatellar contact pressure. One foot of the sensor was fixed against the patellar articular surface, using custom wire and adhesive fixtures that did not interfere with the patellofemoral interface (Fig. 3). The entire patellar surface was covered by the sensor. The other surface of the sensor was left unattached. A commercial lubricant (Vaseline, Unilever, United Kingdom) was applied at the patellofemoral interface between the sensor and the femoral component to reduce shear forces.

The patellar component was attached to the stainless steel cable at a normal (0,84 BPI) Blackburne-Peel Index.[8] Five different tibial component positions in the antero-posterior plane were tested: +6, +3, 0, -3 and -6 mm translation of the centre of the tibial tray compared to the centre of the antero-posterior axis of the tibial plateau. Load and motion input curves used for this experiment simulated a single knee squat from 30°-120°-30° at a fixed flexion rate over a 20 s time interval and with a constant 267 N (60 lbf) target ankle load, which was close to a physiological loading. This single knee squat was repeated 3 times for each tibial component position.

Pearson correlation coefficients were used to assess the association between maximal patellofemoral contact force, contact area and contact pressure and tibial component

position. Statistical analysis were performed with use of SPSS statistical software (version 16.0; SPSS, Chicago, Illinois) with a significance level set at $p < 0.01$.

Results

Patellofemoral contact force.

For all tibial component positions, the PF contact force increased with increasing knee flexion. It reached a maximum, followed by a decrease, which coincided with the occurrence of quadriceps tendon contact with the femoral trochlear groove (Fig. 4). This tendofemoral contact occurred at lower flexion angles for the more anterior baseplate positions compared to the neutral (zero) position. For the posterior tibial positions the PF contact force did not demonstrate a clear peak and the load shearing effect was less obvious, because the quadriceps tendofemoral contact occured only in flexion angles beyond those that were tested.

Posterior tibial positions were associated with lower patellofemoral contact forces during the entire squat, despite the fact that the same target ankle load was achieved for all tests.

At 90° flexion, the contact force was 1945 N (36%) less for the most posterior position (3396 N) compared to the most anterior position (5341 N).

Plotting the maximal patellofemoral contact forces as function of the tibial component position revealed a linear trend, with an increase in maximal PF contact force with more anterior tibial component positions (Fig. 5). The relative change of the PF contact force was defined as the difference between the PF contact force measured at the different tibial component positions and the PF contact force at the neutral position divided by the PF contact force of the neutral position. As such, the patellofemoral contact force on average decreased 2,2 % for every millimetre posterior translation of the tibial component.

Patellofemoral contact area.

For all tibial component positions the patellofemoral contact area gradually increased with increasing knee flexion, reaching a maximum at a flexion angle ranging from 70 to 95° (Fig. 6). The maximal contact area occurred at the flexion angle where there was a transition of the patella from the femoral trochlear groove onto the two femoral condyles. For the neutral tibial component position the contact area reached its maximum at 90° of knee flexion.

Anterior tibial component positions were associated with larger contact areas. The mean difference between the maximum contact area of the most anterior (191 mm^2) and the most posterior (134 mm^2) position of the baseplate was 57 mm^2 (30%).

Patellofemoral contact pressure.

During the entire squat, the average patellofemoral contact pressure was higher for tibial components with a more anterior positions (Fig. 7). The relationship between increasing contact pressure and a more anterior tibial component position was however not linear. The most posterior position (= 6mm posterior) had a higher average contact pressure (= 26,2 MPa) in comparison to the neutral and the 3mm posterior position. The lowest average contact pressure over the entire squat (= 24,6 MPa) was noted for the 3 mm posterior position.

In initial flexion (30°- 80°) the contact pressure of all tibial component positions fluctuated between 19 and 29 MPa. In deeper flexion (90 – 120°), at the stage where the patella leaves the patellofemoral groove and reaches the femoral condyles, a decrease in contact areas and increase in contact pressure was noted for all tibial component positions. The average contact pressures in deeper flexion varied between 28 and 34 MPa, thereby

exceeding the commonly accepted threshold for polyethylene. The maximal contact pressure for the neutral and posterior tibial component positions was always less than 30 MPa, but reached 34 MPa for the most anterior position.

Patellofemoral contact forces during the extension phase of the squat were approximately 10% higher than during the flexion phase, but qualitatively, the effect of the tibial component position on the PF contact force, contact area and contact pressure as function of the flexion angle was the same for the flexion and extension cycle.

Discussion

Our results demonstrate that posterior positioning of the tibial component leads to lower patellofemoral contact forces when performing a knee squat. These lower PF contact forces can biomechanically be explained by the more posterior position of the rotation centre of the knee, which leads to an increased moment arm of the patellar tendon, and consequently to a more efficiently use of the extensor mechanism (fig.1).[9,10]

Anterior displacement of the tibial tubercle, as used in the Maquet and Fulkerson procedures, is based on the same principle. The measurements of Shirazi-Adl and Mesfar[5] demonstrate that at full extension elevation of the tubercle by 1.25 cm results in a 1 cm posterior tibial shift, which diminishes the patellofemoral joint reaction force, an effect that is less obvious with increasing knee flexion.

Maquet measured that two centimetres of anterior displacement was required to reduce the patellofemoral contact force by 50% during a gait cycle.[11] Our data demonstrated a 2,2% reduction of average patellofemoral contact force with each mm of posterior displacement, which is similar to Maquet's findings. This is also broadly consistent with

the data of Singerman et al, who noted a 17% reduction of patellofemoral contact force for a 1 cm anteriorisation of the tibial tubercle.[12]

Our data therefore suggest that a more posterior position of the tibial component is beneficial since this unloads the patellofemoral joint.

Surgeons determine the position of the tibial component usually based upon the best fit they can achieve onto the tibial resection surface. Obtaining cortical support in order to provide solid component seating, while at the same time avoiding overhang, are usually the main criteria to choose the tibial component size as well as the seating position.

Bloebaum et al showed that the tibial cortical bone is significantly thicker in the posterior and medial areas of the tibial plateau.[13,14] Based on their data the most posteromedial position of the tibial tray therefore provides the best bone support for fixation.

In cases where the surgeon has the option between a more anterior or posterior position of the tibial component, the more posterior is therefore advisable because of stronger bone support as well as a reduction in the patellofemoral forces.

In cases where an optimal tibial component fit is available, shifting the component is of course impossible without creating overhang. When the tibial dimensions however fall in between two component sizes, the surgeon has the option of choosing the smaller size baseplate, and may shift the position somewhat away from the anterior tibial bone margin, thereby establishing a more posterior position.

Our study demonstrated lower patellofemoral contact forces for more posterior tibial component positions. Despite the fact that patellofemoral contact area was smaller for posterior tibial positions, the resulting patellofemoral stress during the squat was less.

The patellofemoral contact area for all tibial component positions reached a maximum around 90° of flexion and there after decreased. This corresponded with the transition from one central contact area to two separate smaller contact area's as the patella

progressed from the femoral trochlea onto the two femoral condyles and the intercondylar notch, causing a reduction in total patellofemoral contact area.

The patellofemoral contact pressure (P=F/A) gradually increased with flexion for all tibial component positions (fig. 7). In deeper flexion the contact pressure ranged between 28 MPa and 34 MPa, which exceeds the tensile yield strength of ultra high molecular weight polyethylene, which has been reported to be 25.6 MPa (sd 3.3). Contact between the metal and the polyethylene components in a total joint replacement results in complex stress distributions on the surface and within the polyethylene. In itself, the compressive stress that we measured is not sufficient to explain damage to the polyethylene, as observed for example in retrieved components. These observations must be explained by the range of cyclic stresses that the poly-ethylene experiences during a load cycle. It has been shown that if this range exceeds 23 MPa, UHMWPE fails in fewer than 10^6 cycles.[16,17] Patellofemoral contact pressures exceeding this yield strength will cause polyethylene damage and will jeopardize the longevity of the implant. For the more anterior positions, the patellofemoral contact pressure was continuously greater than 30 MPa as soon as flexion exceeded 90°.

The anteroposterior positioning of the tibial component influences patellofemoral contact mechanics in TKR, with a more anterior position leading to increased stresses on the patellofemoral joint. When the surgeon has the option to position the tibial component more posteriorly, this would be beneficial to the joint.

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Fig. 1:

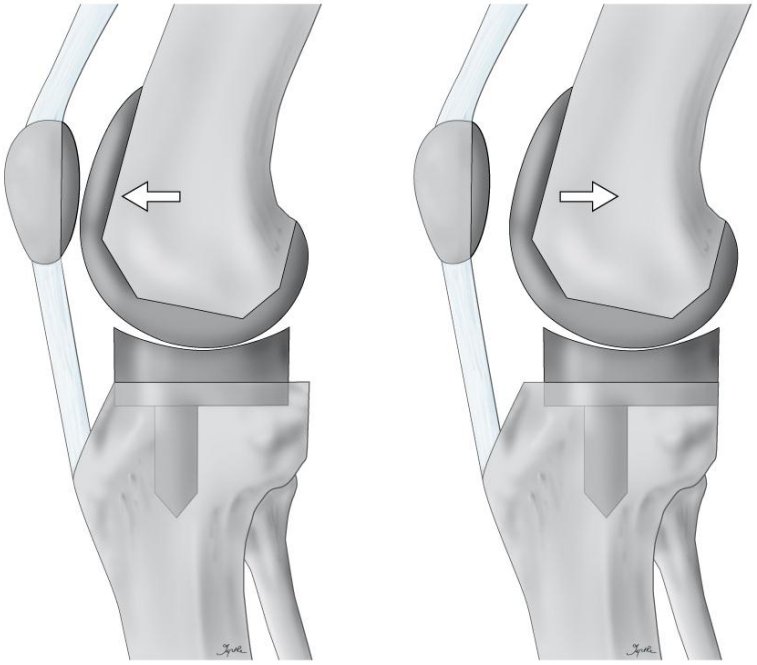

Fig. 1: Diagram demonstrating the effect of changing the anteroposterior position of the tibial component onto the patellofemoral joint.

Fig. 2:

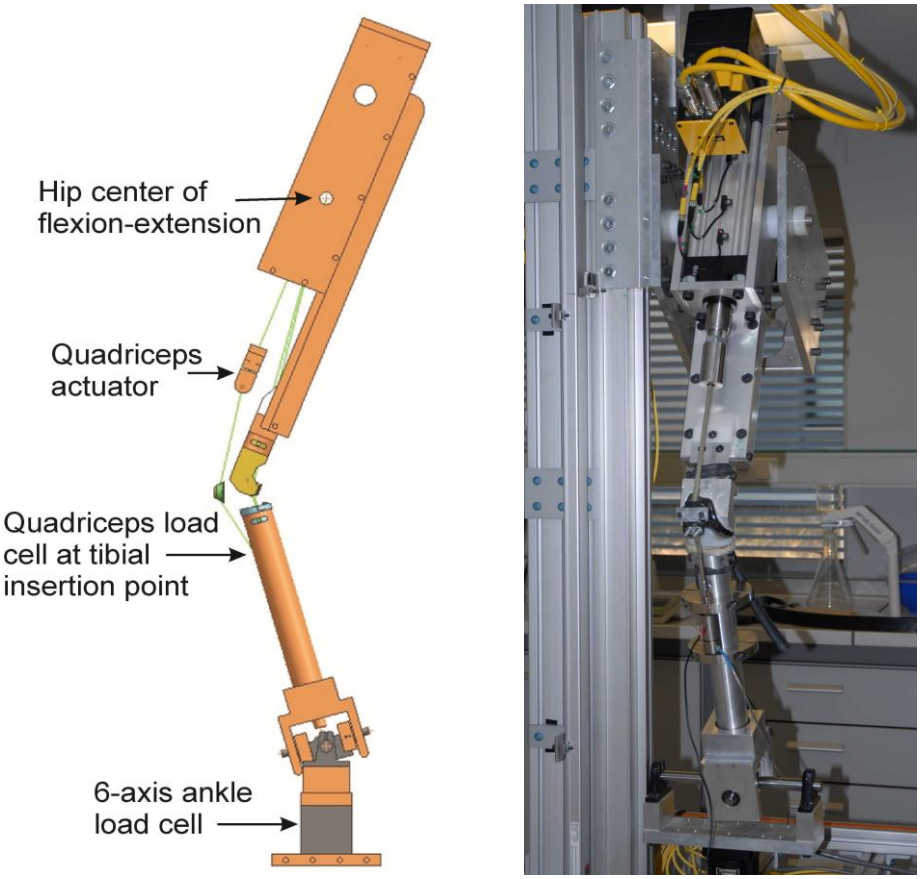

Fig. 2: Schematic drawing and photograph of the knee simulator used for the experiments.

Fig. 3:



Fig. 3: Retropatellair fixation of the K-scan sensor to measure the patellofemoral contact area, contact force and contact pressure.

Fig. 4:

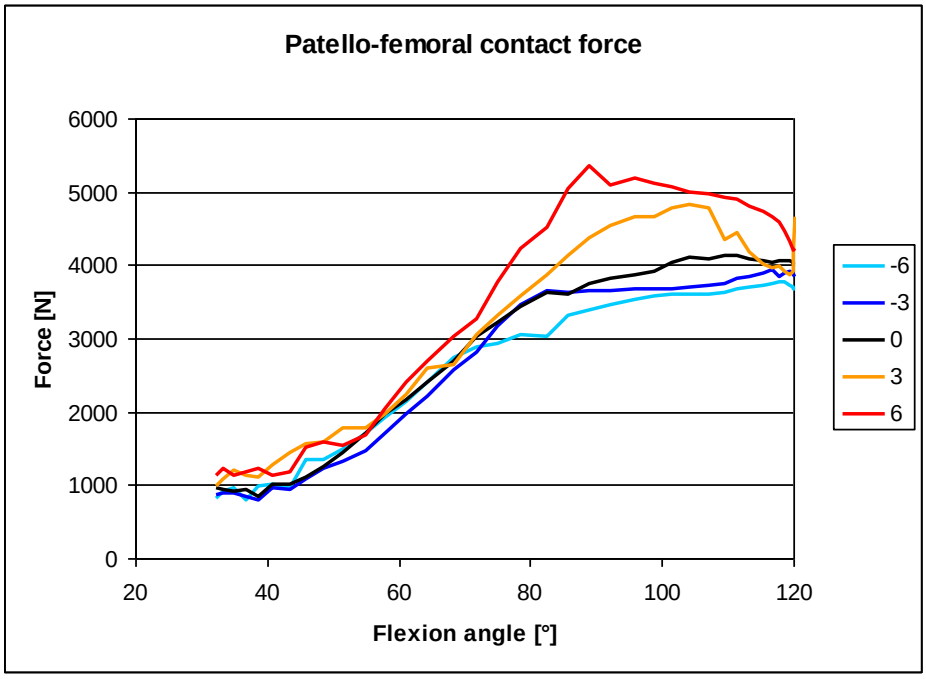

Fig. 4: Experimentally determined patellofemoral contact force as a function of flexion angle for the different positions of the tibial component.

Fig. 5:

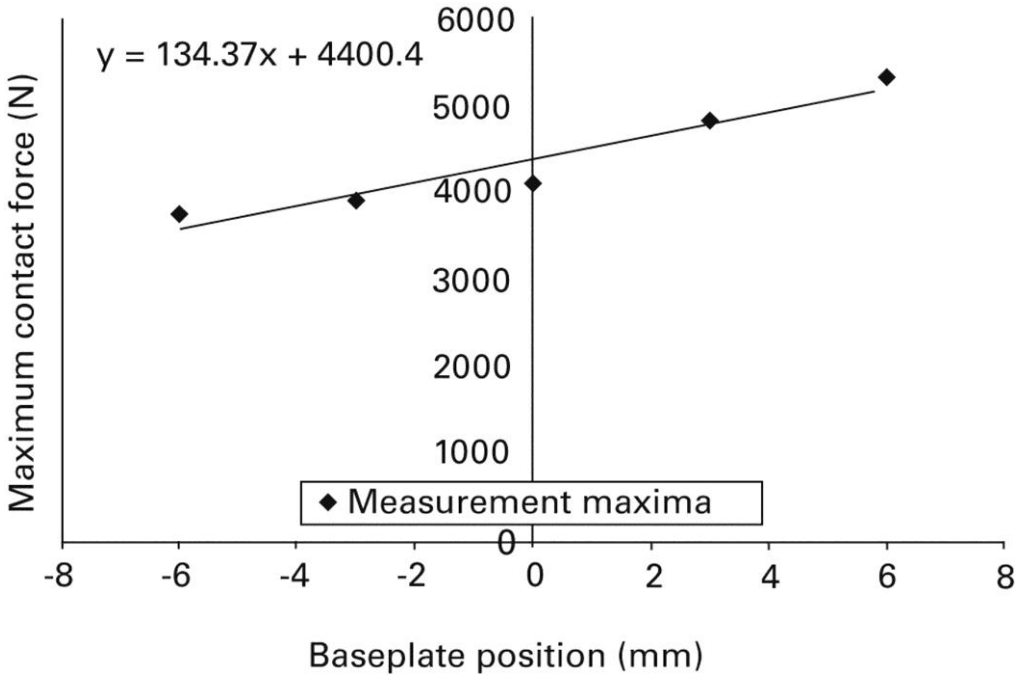

Fig 5: Measured maximum contact forces for the different tibial component positions.

Fig. 6:

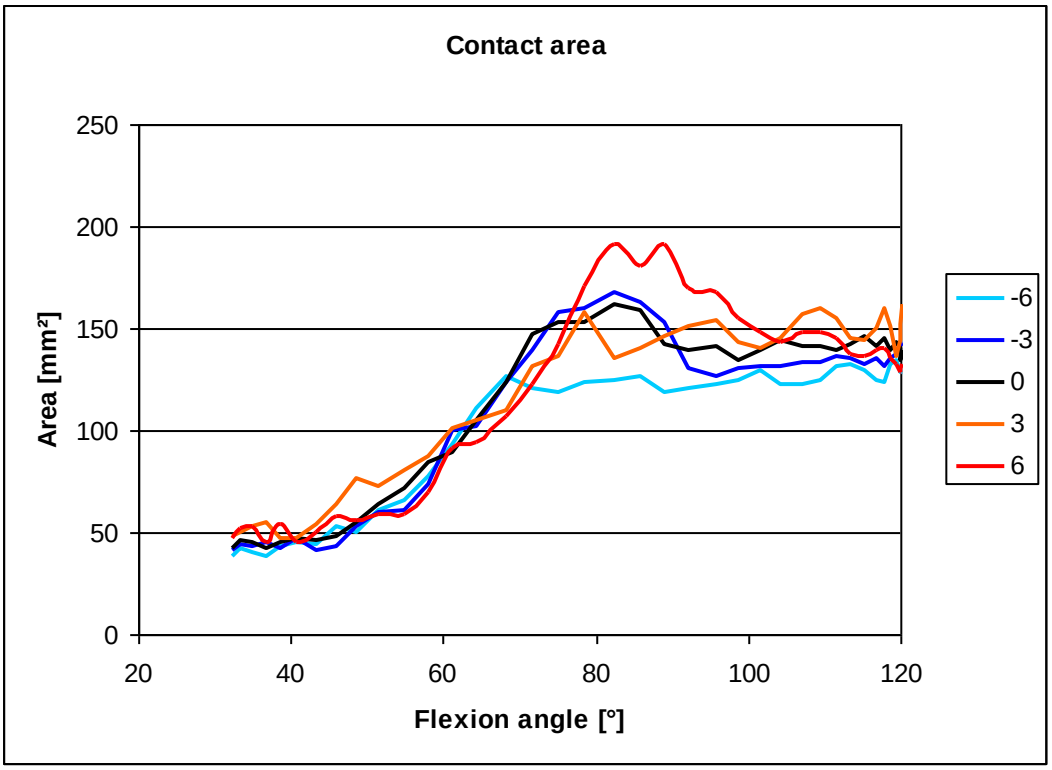

Fig. 6: Experimentally determined patellofemoral contact area for the different tibial component positions as a function of flexion angle.

Fig. 7:

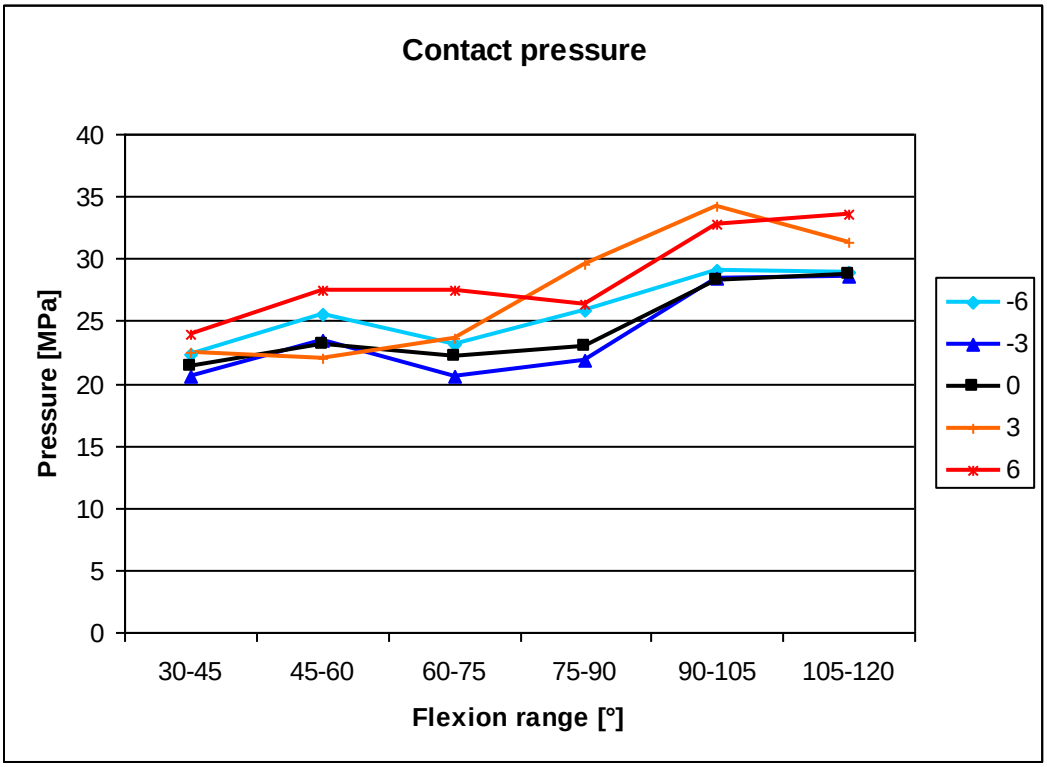

Fig. 7: Average contact pressure for the different tibial component positions as a function of flexion.