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# The effect of component placement on knee kinetics after arthroplasty with an unconstrained prosthesis

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**Abstract**

The mechanical success of a total knee replacement demands stable patellar tracking without subluxation and, stable tracking, in turn, can depend largely on the medial–lateral forces restraining the patella. Patellar button medialization has been advocated as a means of reducing subluxation, and experimental evidence has shown femoral component rotation also affects medial–lateral forces. Surgeons have choices in femoral component rotation and patellar button medialization and must frequently make intra-operative decisions concerning component placement because of anatomical variations among patients. Thus, in seeking to minimize medial–lateral patellar force, we examined the effects of patellar button medialization and external femoral component rotation. The study used an unconstrained total knee system implanted in nine cadaveric specimens tested on a knee simulator operating through flexion angles up to 100°. Tests included all combinations of external femoral component rotation of 0°, 2.5°, and 5° and patellar placement at the geometric center and at 3.75 mm medial to the geometric center. A video-based motion analysis system tracked patellar and tibial kinematics while a six-component load cell measured patellofemoral loads. Repeated measures analysis of variance revealed a statistically significant decrease in the average medial–lateral force with button medialization but no significant change with femoral component rotation. Neither femoral component rotation nor patellar button medialization had an effect on the normal component of the patellar reaction force. External femoral component rotation did cause significant increases in lateral patellar tilt, in tibial varus angle, and in external tibial rotation. Button medialization caused significant increases in lateral patellar tracking, lateral patellar tilt and external tibial rotation. The results in medial–lateral patellar forces quantify the benefit of patellar button medialization and discount any benefit of femoral rotation. The change in tibial kinematics with patellar button medialization and femoral component rotation cannot be measured in vivo with current technology, and the precise clinical implications are unknown. 

## Introduction

Despite the overwhelming success of total knee arthroplasty (TKA) [6,14,21,22,28], patellofemoral complications continue to be a major cause for revision surgery [8,14,18], with a reported incidence of between 7% and 21% [1,7,13,16,29]. Many of these complications have been associated with component positioning and soft tissue balancing [9,10,22]. While the importance of axial alignment has long been recognized, only recently has the importance of patella position and component rotation been advocated.

Yoshii et al. [30] studied the effect of modified design of the femoral component and medialization of the patellar component on patellar tracking. It was found that deepening of the femoral groove and a higher lateral ridge constrained the patella and resulted in a more medial position of the bony patella. A medialized position of the patellar button was found to allow the bony patella to assume its normal lateral position and tilt throughout the range of motion. Rhoads et al. [23] used biplanar radiographs to study the effect of femoral component position on the patellar tracking. The patellar tracking patterns after external rotation of the

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femoral component came closer to reproducing those of the intact knee than any other femoral component position.

Anouchi et al. [2] tested four knee specimens for stability, patellar tracking, and patellofemoral contact with the femoral component positioned in 5° internal, 5° external, and neutral axial rotational alignment. Internal rotation of the femoral component in the knee with perpendicular resection of the tibia caused undesirable results, while external rotation improved both patellar tracking and knee stability. Patellar tracking was closest to normal in the externally rotated specimens.

Singerman et al. [24,25] and Petersilge et al. [19] studied the patellofemoral force balance using load cells. Singerman et al. [25] found differences in the medial–lateral and resultant forces with internal component rotation, and Petersilge et al. found increasing medial–lateral shear loads with increasing flexion angle. Neither study included patellar component medialization in the research.

The purpose of the current study was to examine the hypotheses that medial–lateral patellofemoral forces would be significantly affected by both external femoral component rotation and patellar button medialization and that a combination of the femoral component rotation and patellar button medialization could be found which would minimize the medial–lateral force while maintaining consistent and stable patellar tracking and the normal component of the patellofemoral force. Previous studies neither examined the effect of smaller deviations from the prescribed implantation position nor included patellar button medialization. By recording the results of the imposition of small changes in alignment on knee mechanics, this work concurrently studied the sensitivity to alignment in a cruciate retaining knee system with a modified dome patellar surface and without conforming tibiofemoral surfaces.

## Materials and methods

Nine cadaveric knee specimens were prepared for use in an Oxford-rig knee simulator [5]. All specimens were anatomically intact and suffered no orthopaedic pathologies. The proximal half of the upper leg and the distal half of the lower leg were removed; all other soft tissues were retained. All subsequent surgical procedures were conducted by an experienced orthopaedic surgeon. Through a medial incision extending from the proximal end of the specimen to the proximal end of the resected tibia, the retinaculum was carefully marked at 10 mm increments with cross-wise ink lines and opened to expose the patella. The thickness of the patella was measured and recorded and, after transection of the patella in the coronal plane and identification of the geometric center by bisection of the lines connecting the apexes, four symmetrically placed, 5 mm diameter clearance holes were drilled in the anterior portion of the patella. An aluminum base plate was fixed with bone screws to this anterior portion with the quadriceps and patellar tendons intact. The holes in the patella were aligned with matching holes in the base plate.

A modified dome patellar prosthesis was fixed with short bone screws to a circular plate of a diameter smaller than the button cross-section. Four, 3 mm diameter rods attached perpendicularly to the surface of the circular plate were inserted through the holes in the bony patella. A six component load cell was placed over the four projecting rods and fixed to the aluminum base plate. The cylindrical load cell [R.W. Denton, Rochester Hills, MI], a 5 cm high by 3 cm diameter unit, had a 665 N normal load capacity full scale and 360 N capacity in the two shear directions (accuracy in all forces: <0.06% full scale of each channel, and in all moments: <0.85% full scale of each channel). Cross talk was evaluated at six loadings for each channel (maximum uncertainties for each channel in cross talk: 0.2% maximum medial–lateral error, 0.6% maximum inferior–superior error, 0.6% maximum normal load error; 2% error in-plane moments, and 3% error in torsional moment). Four clamps built into the load cell fixed the rods from the circular plate to the load cell. In the assembled configuration, the load cell was located over the geometric patellar center, and the rods, attached to the load cell, supported the small circular plate and patellar surface, with the rods adjusted to restore the native patellar thickness.

With the load cell and patellar button detached from the base plate, the specimen underwent a total knee replacement using a posterior cruciate retaining system (MGII, Zimmer, Warsaw, IN). Femoral cuts were made in the standard fashion, and the tibia was resurfaced using a standard cutting guide. All three components were placed with a standard technique. The femoral component was placed in neutral rotation relative to the posterior condylar surfaces, and the center of the tibial component was positioned in the geometric center of the tibial plateau. The tibial component was rotated in the standard fashion to the medial third of the tibial tubercle which was chosen for all specimens as 18° internal rotation with respect to the tip of the tubercle [3,4,20]. The modified dome patellar button attached to the circular plate described above was reinserted. The specimen was mounted in the Oxford-rig knee simulator, and the load cell was reattached to the mounting plate. The retinaculum was then carefully sutured along the incision so that the ink markings aligned as originally drawn.

For measurement of the motions of the tibia, femur and patella, a video-based motion analysis system was employed. Rigid rods drilled through the cortical bone attached arrays of five reflective markers to the tibia and femur, and a special mounting plate attached a third marker array to the top of the load cell. Direct linear transformation of the coordinates was applied to calculate the three-dimensional locations of all markers. Using a standard algorithm to determine the instantaneous positions of each bone with respect to the global reference frames [26], the $X$-, $Y$-, and $Z$-coordinates of the markers were established. Tests performed to determine the accuracy of the system, using rigid blocks and instruments of accuracies 0.0005 mm and 0.001°, found translational accuracy to be ±0.3 mm and rotational accuracy to be ±0.1°. Kinematic measurements were taken in a manner identical to standard techniques in gait analysis except that fixation of the markers was essentially rigid.

A strong, 25 mm wide nylon strap was sutured to the proximal end of the quadriceps tendon. The strap from the quadriceps wound through a buckle on one end of a uniaxial load cell, and a power screw of the knee simulator pulled on the other side of this load cell. When mounted in the knee simulator, the strap attachment and power screw actuator were aligned with the long axis of the femur. Threaded rods were bored into the intramedullary canals along the anatomical axes of the tibia and femur and were used to extend the length of each segment to the recorded physiologic length so that the simulator could receive the ends of the rods. The femur was attached to the simulator so that the femoral condylar axis was in the coronal plane at full extension and so that both femoral condyles contacted the tibial plateau with minimal strain in the collateral ligaments. The simulator allowed the knee to assume its physiologic valgus orientation and modeled the anatomic situation for squatting. The quadriceps actuator stayed in a fixed orientation with respect to the femur and acted as the vastus intermedius. A counterweighting system of the knee simulator left a total weight of 44.5 N vertical load applied at the simulated hip. The true weight of half the supported body was not applied because the large loads applied through the quadriceps actuator would cause fracture of the patella along lines connecting the load cell mounting holes.

A joint coordinate system [11] for each body was established using four bony landmarks. A digitizing pointer with five reflective markers on a rigid rod was in view of the motion analysis system and was touched to each landmark. The femoral anatomical axis and the

Table 1
Results of repeated measures ANOVA tests

| *P*-values | Medial–lateral force | Patellar tilt | Patellar shift | Tibial varus–valgus | Tibial external rotation | Normal force |
|---|---|---|---|---|---|---|
| Femoral rotation | 0.82 | <0.01* | 0.90 | <0.01* | 0.03* | 0.89 |
| Patellar medialization | <0.01* | <0.01* | <0.01* | 0.88 | 0.03* | 0.82 |
| Knee flexion | <0.01* | <0.01* | 0.46 | <0.01* | 0.68 | <0.01* |

* Statistically significant.

epicondylar axis established the femoral system with the origin chosen to be at the intersection of the femoral anatomical axis with the intercondylar notch. The tibial anatomical axis and the lateral and medial tibial prominences established the tibial system with the origin along the tibial anatomical axis at the level of the native tibial plateau. The patellar system was established using the resected cross-section of the patella. That is, resection had been performed at the level of the patellar rim, and the load cell base plate had been mounted so that inferior and superior dimples on the base plate aligned with the axis connecting the inferior and superior apexes of the patella. Lateral and medial dimples formed a line perpendicular to the inferior–superior axis. The origin of the patella was the intersection of lines connecting the dimples. The tilt of the patella was most relevant with respect to the femoral component and was essentially the difference between the amount of component rotation and the tilt with respect to the femur. Although kinematic theory shows that one cannot directly compute the difference between the patellar tilt and the amount of component rotation as a simple subtraction of two numbers, because inverse trigonometric functions must be applied to rotation matrices in the computation of sequential rotations, the maximum theoretical difference caused by this imprecise calculation is only about 0.1°.

To study the effects of femoral component rotation and patellar component medial–lateral placement, specimens were tested with six combinations of patellar medial–lateral placement and femoral component rotation. A pair of plates with sets of matching holes was accurately machined to achieve the external rotation of the femoral component. One plate (with three sets of holes corresponding to 0°, 2.5°, and 5° rotations) was attached to the femur and the second plate (with three sets of parallel holes) was attached to the first plate and then to the femoral component. By aligning different sets of holes on the two plates, exact amounts of femoral component rotation were achieved while maintaining firm fixation of the component to the femur. Patellar component positioning was achieved by changing the attachment of the patellar button with respect to the patellar load cell mounting plate. The mounting plate had holes placed corresponding to the exact neutral position and to 3.75 mm medialization.

The amount of external femoral rotation and button medialization were chosen to reflect typical choices a surgeon would make. The 2.5° variations bracketed the placements based on assessment of the epicondylar axis and the plane of the posterior condyles and also included the possibility that one may intentionally rotate the component externally to improve patellar tracking. The patellar placement mimicked positioning that would replicate the average location of the patellar ridge or would offer a more balanced coverage of the resected surface when a circular patellar button is used.

Using the quadriceps mechanism to actuate movement, each specimen went through three flexion-extension cycles of 0°–100° for each choice of component placement. The movement was very slow, requiring 40 s for each cycle. A signal from the motion analysis control unit synchronized kinematic and load data collection.

To account for variations due to any viscoelasticity with longer time scales, only the third of the three flexion-extension cycles was processed. In order to minimize the noise in the measured results, a sixth-order Butterworth filter with a cutoff frequency of 2.5 Hz was used. Forward and backward passes were used to eliminate a phase shift. The weight of the load cell (~400 g) introduced an additional load which the patella and quadriceps mechanism had to equilibrate. The results for smaller flexion angles were more affected by the weight because the ratio of the quadriceps and patellar reaction loads to the weight was small. The results reported herein do not include any adjustment for the effect of this weight, but this work includes results only at flexion angles greater than 10°. Beyond 10°, the effect of the load cell was very small in comparison to that of the quadriceps and resultant normal patellar loads.

To study the interaction among femoral component rotation, patellar component medialization, and knee flexion angles, repeated measures ANOVA tests were performed. One specimen, which exhibited poor tracking in the usual component alignment, would not have been acceptable in the operating room and appropriate corrective steps such as a lateral release would have been performed. Because no corrective steps were taken, the statistical analysis was, therefore, conducted both with and without this specimen included. There were no differences in the statistical results between inclusion or omission of that specimen; it was not included in the presentation of results.

## Results

Statistically significant effects in the repeated measures analysis were found ($P < 0.05$) for several dependent variables (Table 1). No results showed statistical significance at individual flexion angles, however.

Patellofemoral medial–lateral forces were significantly affected by patellar button medialization but not

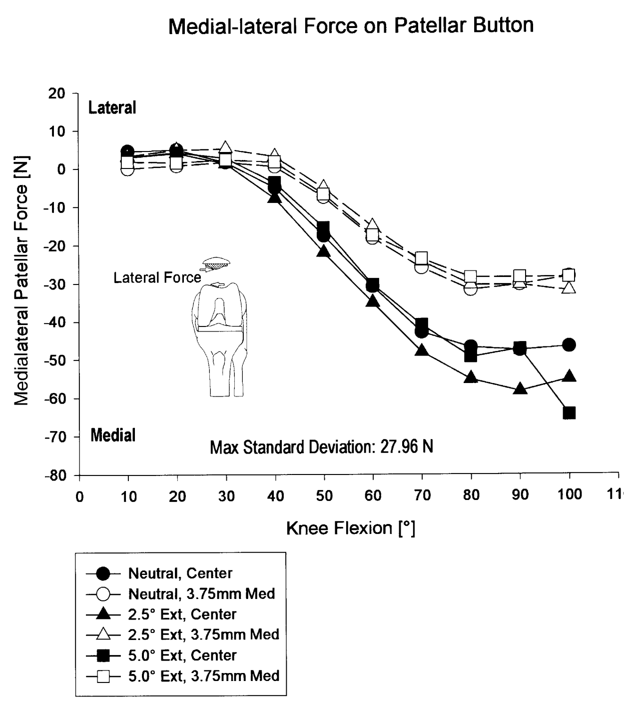


Fig. 1. Mediolateral force on the patellar button as measured in a patellar reference frame ($n = 8$ except at 100°, where $n = 7$; the maximum S.D. = 28 N: each curve can be considered to lie in a band of less than this value).

by femoral component rotation ($P < 0.01$) (Fig. 1). The average normal patellofemoral forces and quadriceps actuation forces for all six combinations of component placement increased with flexion angle as expected, and no effect due to changes in femoral or patellar alignment was evident.

On average, as the knee flexed, the patella tilted either less medially or more laterally with respect to the femoral component with either button medialization or femoral component rotation (Fig. 2). The general trend to more laterally directed tilt was statistically significant ($P < 0.01$) for both femoral component rotation and patellar button medialization.

Medialization of the patellar button caused the bony patella to shift laterally with respect to the origin of the femur as the knee flexed (Fig. 3). Rotation of the femoral component had little effect on patellar shift, but the effect of patellar button medialization was statistically significant ($P < 0.01$).

Increasing femoral component rotation had an increasing and significant ($P < 0.01$) effect on tibial varus–valgus alignment (Fig. 4). That is, more external femoral component caused a more varus tibial alignment, and the amount of varus increased with flexion. External tibial rotation tended to increase due both to external femoral component rotation and patellar button medialization (Fig. 5). The effect of femoral component rotation became smaller as flexion angle increased, until there was little difference in tibial rotation at 90° flexion.

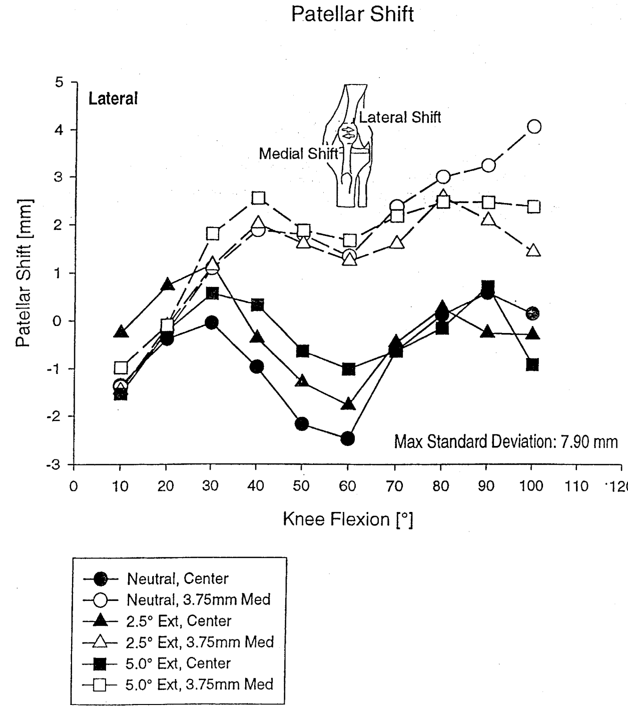


Fig. 3. Mediolateral patellar shift with respect to the femur. A lateral shift is positive on the vertical axis ($n = 8$; the maximum S.D. = 7.9 mm: each curve can be considered to lie in a band of less than this value).

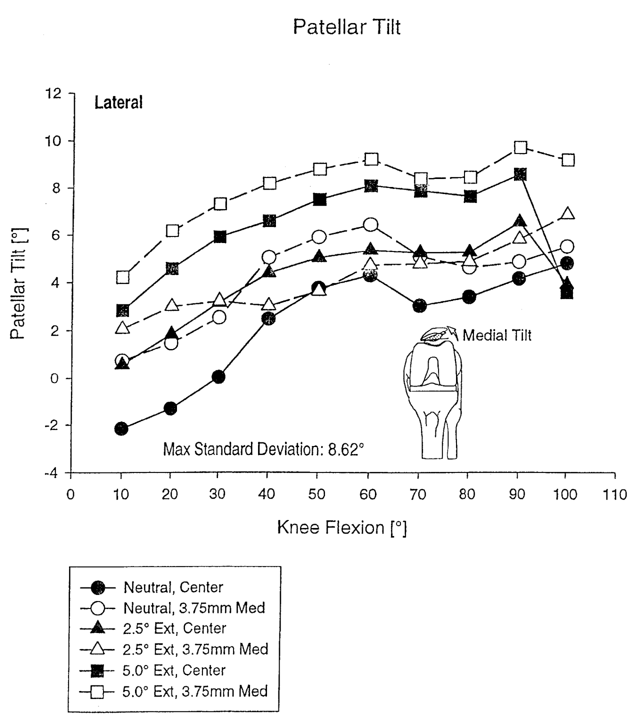


Fig. 2. Patellar tilt with respect to the femur. Lateral tilt is positive on the vertical axis ($n = 8$; the maximum S.D. = 8.6°: each curve can be considered to lie in a band of less than this value).

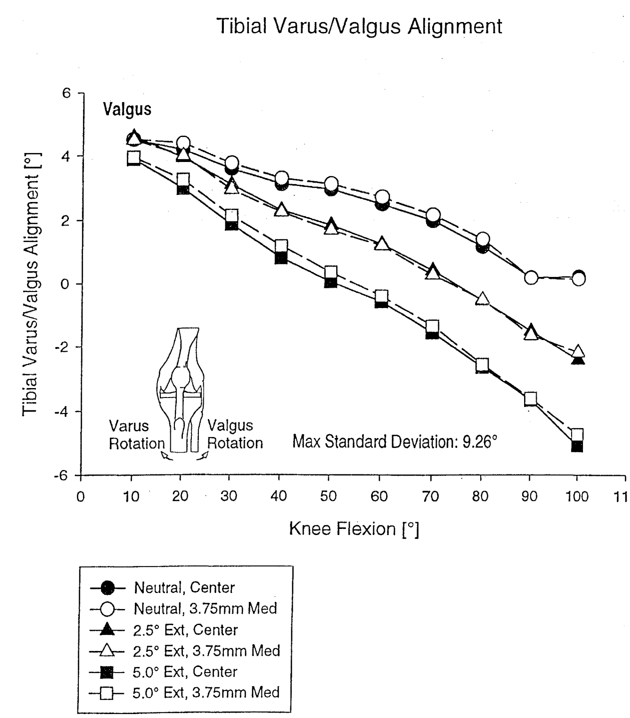


Fig. 4. Tibial varus/valgus orientation. Valgus alignment is positive on the vertical axis ($n = 8$; the maximum S.D. = 9.3°: each curve can be considered to lie in a band of less than this value).

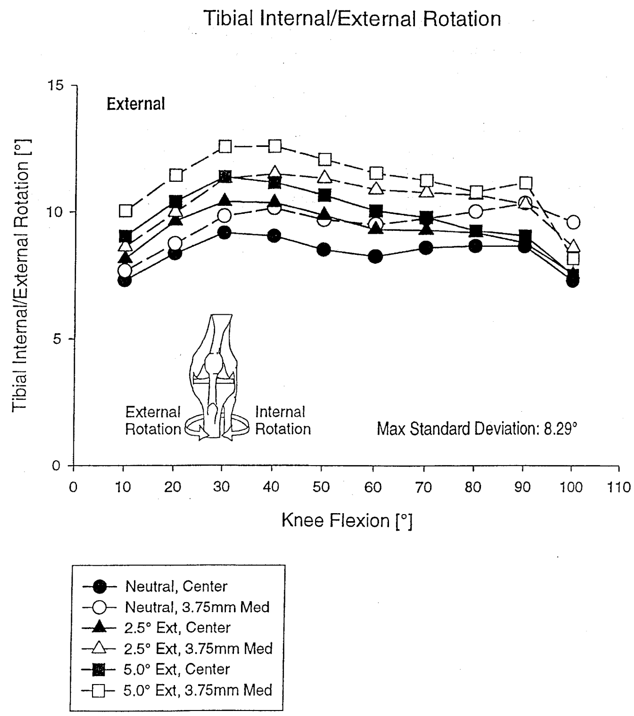


Fig. 5. Tibial internal/external orientation. External rotation is positive on the vertical axis ($n = 8$; the maximum S.D. $= 8.3°$: each curve can be considered to lie in a band of less than this value).

The overall effect was significant ($P < 0.03$). Patellar button medialization also produced a statistically significant amount of external tibial rotation ($P = 0.03$).

## Discussion

The reduction in medial–lateral forces showed quantifiable evidence of an advantage of patellar button medialization. While clinical evidence [12] already indicated the benefit of this procedure, the current results indicate that the effect of patellar placement for a TKA system might be determined preclinically using available in vitro techniques. There was no optimal placement of the components for minimization of medial–lateral patellar force because femoral component rotation had no effect on this force.

The changes in loading were a consequence of the kinematic changes. The reduction in medial–lateral force due to patellar button medialization should be due largely due to a resulting change in the $Q$-angle. The $Q$-angle changed because the bony patella moved laterally relative to the contact between the patella button and the femur. The movement of the bony patella relative to the femur and the medial direction of the medial–lateral force on the patella (Fig. 3) indicate that the patella tracked on the lateral femoral rim in all cases. Although femoral component rotation alone could alter the $Q$-angle, the change in femoral component rotation instead altered tibial rotation (Fig. 4) and not medial–lateral patellar force (Fig. 1). As the tibia rotated with the femoral component, the $Q$-angle would have remained essentially constant.

Although femoral component rotation and patellar button medialization both significantly affected patellar tilt, the implications of the relationships are profoundly different. As the femoral component rotated, the patella merely rotated with it. This fact is supported by the lack of substantial change in patellar shift with rotation. Thus, one may describe the patellar tracking to be insensitive to femoral component rotation. The increase in patellar tilt with button medialization indicated that patellofemoral contact was probably altered. The nonconformal patellofemoral surfaces did not prescribe a single tracking path. As the button was medialized, the bony patella shifted laterally (Fig. 3), but the button remained in approximately the same position in the condylar groove, riding along the lateral femoral ridge. Given that tilt changed with medialization, the contact regions of the patella and condyles would have changed slightly with medialization thus causing the tilt when the quadriceps and patellar tendon loads were equilibrated by the patellofemoral reaction loads.

The statistically significant change in tibial varus–valgus and tibial internal–external rotation were associated with the femoral component rotation, but these changes have no known consequences. The MGII design used in this study has an anterior lip and low central ridge separating the lateral and medial compartments. This ridge would cause the tibial internal–external rotation to follow the femoral component alignment because the tibial tray would be pressed against the lateral and medial condyles. At 90° of knee flexion, however, the tibia contacted the posterior condylar surface instead of the inferior condylar surface. The posterior condyles necessarily rotated with the component and are oriented at a 90° angle to the inferior condylar surface. Thus, the tibial varus–valgus alignment at 90° of knee flexion was altered by an amount equal to the component external rotation. What was an internal–external constraint at full extension became a varus–valgus constraint at 90° of knee flexion.

In studying larger femoral component rotations, Singerman [25] found that 10° internal component rotation caused higher medial–lateral forces up to 95° knee flexion. Although they also studied 10° external component rotation, the results were not specifically reported and apparently did not produce a significant effect. If Singerman found no differences with external rotation, the current work is in general agreement. The results of the current study, however, quantify the effect for the smaller external rotations that are more likely to occur in practice. Petersilge et al. [19] reported increasing medial–lateral shear loads with increasing flexion angle, but did not differentiate between medially and laterally directed loads. The increase in medial–lateral force followed the increase in quadriceps load with flexion angle, as in the current work. In a study of advancement of the tibial tubercle, Singerman et al. [24] found almost no change in the average force on the lateral patellar facet. They reported a typical value of the mediolateral force and also a value for the average change and standard deviations in this force. Their typical result combined with their average result indicates that the mediolateral force could have been either laterally or medially directed depending on the individual specimen. That variation in directionality generally agrees with the mediolateral force results of the current study, although only at small flexion angles.

The alteration of patellar kinematics with button medialization generally fits well with work relating *Q*-angle and patellar kinematics. The report of Merkow et al. [17] successfully treated patellar dislocation with proximal realignment of the extensor mechanism, and Van Kampen and Huiskes [27] imposed tibial rotation with respect to the intact native femur in vitro to alter the patellar shift and tilt. The change in patellar component positioning of the current study also would have altered the *Q*-angle so that the present results in patellar tilt and medial–lateral loading agree in principle with these two studies.

Although no previous in vitro work studied clinically relevant patellar button medialization and femoral component rotation, the relationship of tilt, shift and medialization is in general agreement with the work of Yoshii et al. [30] and Lewonoski et al. [15]. Although Yoshii et al. found an increased lateral tilt with medialization that was directionally the same as that in the current work, the amount of button medialization used was much greater. Lewonoski et al. clinically demonstrated that medialization alters tilt but did not report the direction of the change.

The current results showed that the modified dome patella tilted almost as much as the femoral component rotated. The amount of tilt per degree of femoral component rotation differed only slightly from the result of Rhoads et al. who imposed a 10° femoral rotation and found that a patella resurfaced with a dome tilted less than 10°. The difference could be due to the amount of femoral component rotation, the shape of the patellar surface or nonlinearity in the response of the patella to the component rotation.

No large kinematic differences were noted for the eight specimens in this study, so the limits of patellar stability must lie outside the ranges of patellar and femoral component placement. Small changes in component alignment should cause only small changes in joint kinetics in a well-designed implant, given good ligamentous stability and a well-aligned extensor mechanism. The insensitivity of the knee system to these small variations is a benefit because anthropometric differences and small surgical deviations must be tolerable. The current study did not track relevant anthropometric variables such as the relative positions of ligamentous insertions and the shape of the original articular surfaces. Variables such as these would certainly affect patellar kinetics and clinical outcome. Continued attention to their relevance could provide a more unified understanding of the patellofemoral joint.

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