Original Article
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Load-Dependent Characteristics of Cruciate-Retaining and Posterior-Stabilized Total Knee Arthroplasty: A Biomechanical Study

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Background: Increased load bearing across the patellofemoral and tibiofemoral articulations has been associated with total knee arthroplasty (TKA) complications. Therefore, the purpose of this study was to quantify the biomechanical characteristics of the patellofemoral and tibiofemoral joints and simulate varying weight-bearing demands after posterior cruciate ligament-retaining (CR) and posterior-stabilized (PS) TKAs.

Methods: Eight fresh-frozen cadaveric knees (average age, 68.4 years; range, 40–86 years) were tested using a custom knee system with muscle-loading capabilities. The TKA knees were tested with a CR and then a PS TKA implant and were loaded at 6 different flexion angles from 15° to 90° with progressively increasing loads. The independent variables were the implant types (CR and PS TKA), progressively increased loading, and knee flexion angle (KFA). The dependent variables were the patellofemoral and tibiofemoral kinematics and contact characteristics.

Results: The results showed that at higher KFAs, the position of the femur translated significantly more posterior in CR implants than in PS implants (36.6 ± 5.2 mm and 32.5 ± 5.7 mm, respectively). The patellofemoral contact force and contact area were significantly greater in PS than in CR implants at higher KFAs and loads (102.4 ± 12.5 N and 88.1 ± 10.9 N, respectively). Lastly, the tibiofemoral contact force was significantly greater in the CR than the PS implant at flexion angles of 45°, 60°, 75°, and 90° KFA, the average at these flexion angles for all loads tested being 246.1 ± 42.1 N and 192.8 ± 54.8 N for CR and PS implants, respectively.

Conclusions: In this biomechanical study, CR TKAs showed less patellofemoral contact force, but more tibiofemoral contact force than PS TKAs. For higher loads across the joint and at increased flexion angles, there was significantly more posterior femur translation in the CR design with a preserved posterior cruciate ligament and therefore significantly less patellofemoral contact area and force than in the PS design. The different effects of loading on implants are an important consideration for physicians as patients with higher load demands should consider the significantly greater patellofemoral contact force and area of the PS over the CR design.

Keywords: *Total knee arthroplasty, Cruciate retaining, Posterior stabilized, Contact area, Contact force*

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The patellofemoral and tibiofemoral joints experience great mechanical demands after total knee arthroplasty (TKA). Studies have shown that increased weight bearing changes knee kinematics and contact characteristics.[1-3) Both biomechanical and physiological mechanisms of TKAs are associated with the mechanical stresses that are related to osteoarthritis.[4,5) For patients with severe and advanced osteoarthritis, TKAs are the best treatment option. While TKA is one of the most common and cost-effective procedures for osteoarthritis, studies show 20% or more of patients may not have a positive clinical outcome;[6) therefore, it is important to understand the kinematic and biomechanical effects of TKA designs in patients with various weight-bearing demands.

Two of the current functional TKA designs, posterior cruciate ligament-retaining (CR) and posterior-stabilized (PS), are widely used but both have advantages and disadvantages. Advantages of the CR implant are inherent stability, improved kinematics, bone preservation, and better implant stabilization. The advantages of the PS implant are ligament balancing, better knee flexion, femoral rollback restoration, and lower range of axial rotation and condylar translation.[7-9) Thus, there is more importance on observing other patient factors and surgical indications to make the best decision for a CR or PS, weight-bearing demands being one of these necessary considerations.

Certain kinematic and physiological differences may play a role in potentially reducing patellofemoral and tibiofemoral forces, particularly for high weight-bearing demand populations, which may already see elevated patellofemoral and tibiofemoral pressures. Excess joint loading places higher contact stresses on the implants that could lead to increased wear, loosening, and presumably higher failure rates. The role of high weight-bearing demands on implant kinematics and contact characteristics must be further evaluated.

The purpose of this study was to quantify the biomechanical and kinematic effects within the patellofemoral and tibiofemoral joints in cadaveric knees with increased loading following TKA with both CR and PS implants. The hypothesis of this study was that with increased weight-bearing demands, the CR implant would demonstrate more posterior translation at higher knee flexion angles (KFAs) than the PS implant due to the preserved posterior cruciate ligament (PCL). A second hypothesis was that both implants would show higher contact pressures, contact area, and contact forces across the patellofemoral and tibiofemoral joints with high weight-bearing conditions.

METHODS

Institutional Review Board approval was waived by Congress Medical Foundation as this was a basic science study.

Specimen Dissection and Preparation

Eight fresh-frozen knees (4 male and 4 female knees) with an average age of 68.4 years (range, 40–86 years) were used. Specimens were screened for any degenerative qualities in bones or soft tissues. Skin and subcutaneous fat were removed, and care was taken to preserve the joint capsule and retinaculum. A medial parapatellar arthrotomy was performed. The TKA system (Lospa; Corentec Co. Ltd.) with a conventional tibia insert, which offers both a CR and a PS implant within a single system, was used according to the manufacturer's guidelines. A 6° distal femoral resection was performed with intramedullary guidance. The femoral component rotation was set by posterior condylar referencing using the epicondylar axis for confirmation. An intramedullary guide was used for the tibial resection, which was perpendicular to the tibial shaft establishing a neutral varus-valgus resection. Tibial resection was then performed with an extramedullary tibial cutting guide. To establish the posterior slope, the shaft of the extramedullary tibial cutting guide was adjusted so that it was parallel to the long axis of the tibia in the anterior/posterior plane. Thus, the posterior slope was set to match the native posterior slope, and the PCL was preserved in all specimens.

Medial and lateral joint gaps were measured, and appropriate soft-tissue releases were performed to balance flexion and extension gaps within 1 mm of each other. Trial components were inserted to verify size and balance. The native patella was measured with calipers and resected to allow for minimal patellar thickness differences with the implant. All final tibial and femoral components were pressed fit. Plaster-of-Paris was used to fix the patellar component. The medial parapatellar arthrotomy was closed. The CR implant specimens were subjected to the testing protocol.

After completing testing of the CR specimen, all soft-tissue constraints, needed to convert to the PS construct, were removed, including the PCL. The femoral cutting guide for the PS implant was used for the femoral box cut and all gaps were balanced within 1mm of each other. The PS femoral component was placed, and the PS tibial polyethylene was inserted. The PS implant specimens were taken through an identical testing protocol.

Specimen Mounting

The femur was potted in a polyvinyl chloride (PVC) pipe with plaster of Paris for mounting onto the custom knee testing system that permitted anatomically based muscle loading and 6 degree-of-freedom to position the knee (Fig. 1).[10-13] The femur was secured within a cylinder with the femoral epicondylar axis aligned with the coronal plane of the testing system and locked into place. An intramedullary rod was inserted into the tibia and guided through a bracket to hold the tibia in 0° of flexion. KFA could be adjusted from the femoral cylinder. Five degree-of-freedom at the tibia remained throughout all testing stages while the femur was fixed.

Muscle Loading

Multi-plane muscle loading of the quadriceps and hamstrings was performed to simulate physiologic loading conditions.[11] The loading of the quadriceps mechanism included the vastus medialis (VM), rectus femoris (RF), vastus intermedius/lateralis (VI/VL), and the hamstrings including the biceps femoris and semimembranosus/semitendinosus (SM/ST). Ratios for loading based on the physiological cross-sectional area of the muscles were applied to create 3 different incremental loading conditions.[14] The total muscle loading was 375 N, 450 N, and 525 N in sum for each condition to simulate high load-bearing conditions across the tibiofemoral and patellofemoral articulations.

The muscle loading for the first condition of 375 N total was VM 65 N, RF 110 N, VL 95 N, BF 40 N, and SM/ST 65; then for 450 N, the muscle ratios were VM 80 N, RF 130 N, VL 115 N, BF 45 N, and SM/ST 80 N; and lastly, for 525 N total, the muscle ratios were VM 90 N, RF 150 N, VL 130 N, BF 60 N, and SM/ST 95 N. The knees were loaded in accordance with the 3 load conditions at 6 different flexion angles of 15° increments (15°, 30°, 45°, 60°, 75°, and 90°). The flexion angles were confirmed with a digital goniometer. The specimens were tested at each of the flexion angles before increasing the load.

Measured Parameters

Patellofemoral and tibiofemoral contact area, contact pressure, peak pressure, and contact force were measured using the K-scan system (Tekscan Inc.). Contact area represents the amount of intra-articular surface area (mm^2) contact between the patella and femur or tibia and femur. The contact force is the force transmitted through the articular contact area. Contact pressures are calculated by contact force divided by contact area (N/mm^2). For this study, the Tekscan pressure sensor system collects contact characteristics, which can be analyzed on the I-Scan software for the contact area, force, and pressure. The K-scan sensor (model 9000) consisted of 1 sensor pad that was inserted through a suprapatellar arthrotomy into the patellofemoral joint to measure contact characteristics. Similarly, for the tibiofemoral contact measurement, 2 pads were inserted posteriorly in the tibiofemoral space under both the lateral and medial condyles between the implant components (Fig. 2). A calibration curve for each sensor was generated for each specimen and validated by a separate material testing machine (Instron Corp.). The calibration was performed for a load of 60 N for the patellofemoral joint and 150 N for the tibiofemoral joint to cover the range of low loads

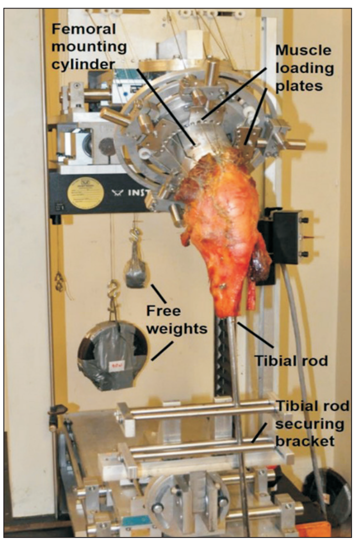

Fig. 1. Photograph of a left knee mounted on the custom knee testing system.

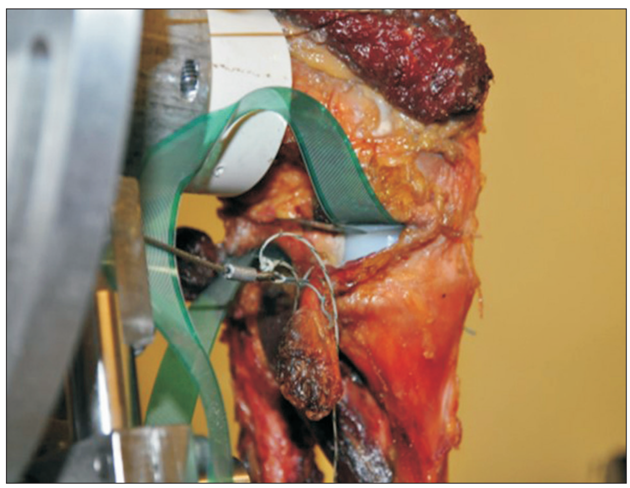

Fig. 2. Photograph of a Tekscan pad inserted posteriorly in the medial tibiofemoral joint space.

to higher muscle loads. Contact areas, contact pressures, peak contact pressures, and contact force were obtained and analyzed for the patellofemoral and tibiofemoral compartments of each specimen.

Femorotibial position was measured with a MicroScribe 3DLX digitizer (Revware Inc.) using 6 separate markers, marked with small screws on the tibial and femoral surfaces. Three points were placed on the distal femur at the lateral femoral epicondyle, the medial epicondylar sulcus, and the posterior femur 6 cm superior to the tibiofemoral joint line. On the proximal tibia, 3 points were placed, 1 on Gerdy's tubercle, 1 3 cm inferior to the medial joint line centered in the anterior-posterior tibial plane, and 1 on the posterior tibia 6 cm inferior to the tibiofemoral joint line. Tibial translation and rotation were measured relative to the femur.

Statistical Analysis

A sample size calculation for comparing paired differences was performed on the first 2 specimens tested with an average difference in translation of 8 ± 6 mm. Testing 8 specimens would achieve a power of 80% and a level of significance of 5%. For all measurements, 2 reproducible trials were performed for each measurement and reproducibility was confirmed. The average of the 2 trials was used for analysis. Statistical analysis was performed using a repeated measures analysis of variance with a Tukey post hoc test to compare data across the 3 loading conditions. Comparisons between the CR and PS groups were made with a paired *t*-test. All data are reported as means with standard deviations. The significance level was set at 0.05.

RESULTS

Femorotibial Kinematics

At 75° and 90° of knee flexion, the position of the femur in the CR knees was significantly more posterior relative to the tibia compared to the PS knees, averaging 36.6 ± 5.2 mm and 32.5 ± 5.7 mm, respectively (Fig. 3). The femoral rollback describes how at higher flexion angles the femur articulates with the tibia, approximately 4 mm more posterior in the CR than in the PS implant (Table 1): at 75° flexion and for 375 N, 450 N, and 525 N, $p = 0.010$ for all; at 90° flexion and for 375 N, $p = 0.049$; and at 90° flexion and for 450 N and 525 N, $p = 0.040$.

There was a positive correlation between femur posterior position and load in both knee implants; greater loads resulted in more femur posterior translation. Only at 75° of KFA was there a significant difference between the CR and PS implants for the change in posterior position from 375N to 525 N (Fig. 4). The PS implant was more sensitive to loading increases as it experienced a greater change in posterior position between loads than the CR implant. Lastly, the tibial rotation was unaffected by an

Table 1. Average Difference Between Posterior Stabilized and Cruciate Retaining Total Knee Arthroplasty

Knee flexion angle	375 N muscle loading (mm)	450 N muscle loading (mm)	525 N muscle loading (mm)
75°	4.5 ± 3.7	3.9 ± 3.2	3.5 ± 3.0
90°	4.7 ± 5.6	4.3 ± 4.9	3.8 ± 4.4

Values are presented as mean ± standard deviation.

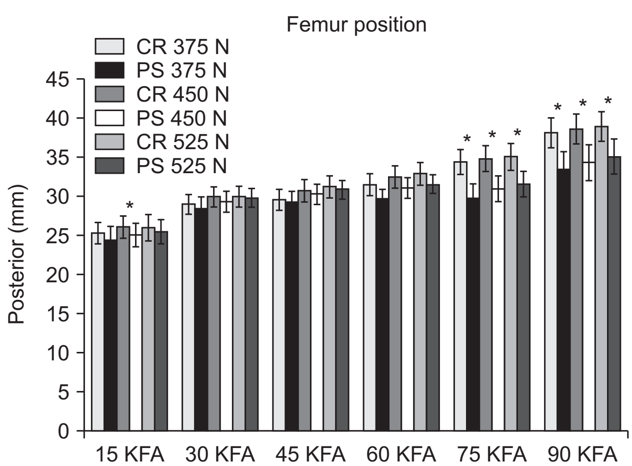

Fig. 3. Femur position relative to the tibia for each implant, cruciate-retaining (CR) or posterior-stabilized (PS), and muscle loading condition simulating increased loading. Values for each knee flexion angle (KFA) are shown. *$p < 0.05$, CR vs. PS.

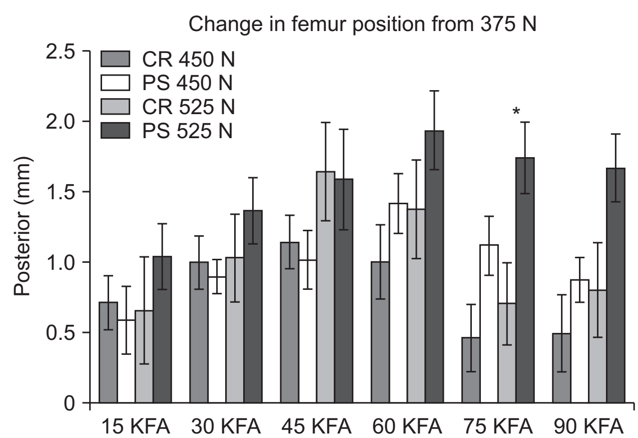

Fig. 4. Change in femur position relative to the tibia from 375 N loading condition for each implant, cruciate- retaining (CR) or posterior-stabilized (PS). Values for each knee flexion angle (KFA) are shown. *$p < 0.05$, CR vs. PS.

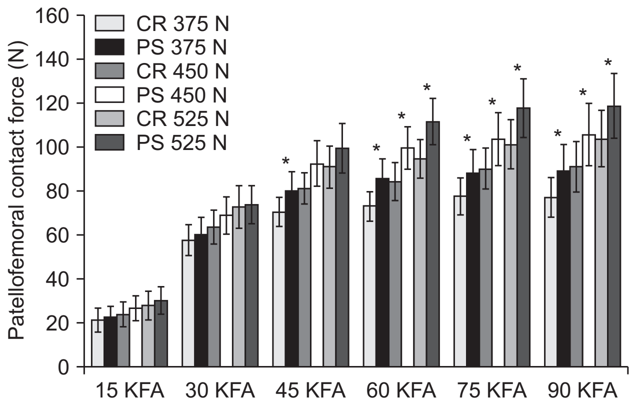

Fig. 5. Patellofemoral contact force for each implant, cruciate-retaining (CR) or posterior-stabilized (PS), and muscle loading condition simulating increased loading. Values for each knee flexion angle (KFA) are shown. *$p < 0.05$, CR vs. PS.

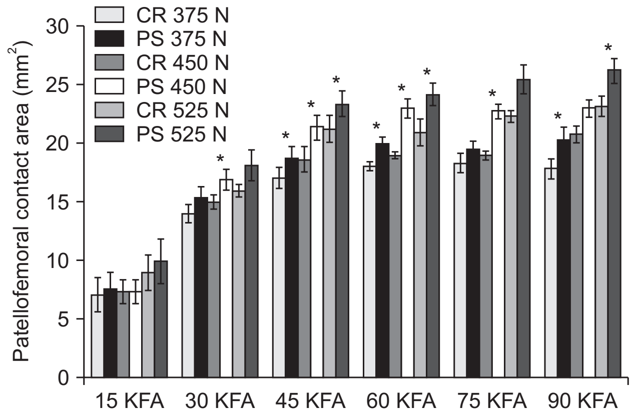

Fig. 6. Patellofemoral contact area for each implant, cruciate-retaining (CR) or posterior-stabilized (PS), and muscle loading condition simulating increased loading. Values for each knee flexion angle (KFA) are shown. *$p < 0.05$, CR vs. PS.

Table 2. Difference in Contact Force in the Patellofemoral Joint

Knee flexion angle	375 N muscle loading (N)	450 N muscle loading (N)	525 N muscle loading (N)
60°	13.1 ± 10.6	15.3 ± 8.9	17.1 ± 8.2
75°	10.9 ± 9.0	13.5 ± 10.9	16.8 ± 9.3
90°	12.6 ± 10.6	14.5 ± 8.9	15.2 ± 7.8

Values are presented as mean ± standard deviation.

increase in loading through all flexion angles, regardless of knee implant type.

Patellofemoral Contact Characteristics

At higher KFAs (60°, 75°, and 90° KFA), there was a significantly greater patellofemoral contact force for PS knees than CR knees at each loading condition tested ($p < 0.05$), averaging 102.4 ± 12.5 N and 88.1 ± 10.9 N, respectively (Fig. 5). The differences in contact force between the CR and PS implants are described in Table 2. There was also a significantly greater patellofemoral contact area for PS than CR knees for multiple positions, including all loading conditions at 45° and 60° KFA, and several other loading conditions at 30°, 75°, and 90° KFA (Fig. 6). There were no significant differences in mean and peak patellofemoral contact pressure comparing CR and PS knees at each loading condition tested.

Tibiofemoral Contact Characteristics

The medial and lateral tibiofemoral contact area, contact pressure, peak pressure, and contact force increased linearly with each increase in loading regardless of implant type. The CR knees had significantly greater tibiofemoral contact force compared to the PS knees at all loading conditions for 75° and 90° KFA and at 2 loading conditions at 45° and 60° KFA ($p < 0.05$), averaging 246.1 ± 42.1 N and 192.8 ± 54.8 N for CR and PS implants, respectively (Fig. 7). There was no significant difference in the change in tibiofemoral contact force from loading between CR and PS knees, indicating that both implants proportionately experienced more contact force with incremental loading.

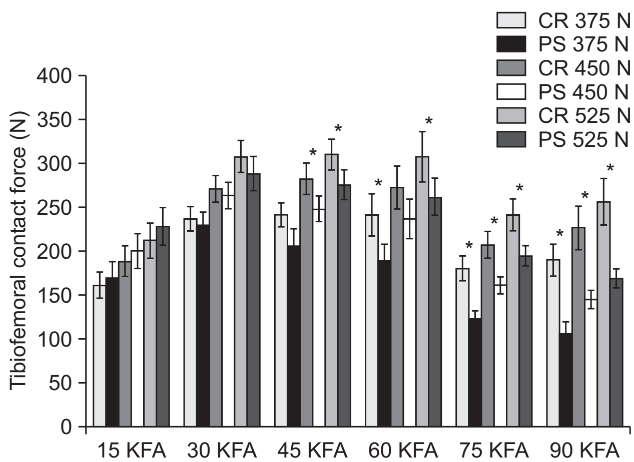

Fig. 7. Tibiofemoral contact force for each implant, cruciate-retaining (CR) or posterior-stabilized (PS), and muscle loading condition simulating increased loading. Values for each knee flexion angle (KFA) are shown. *$p < 0.05$, CR vs. PS.

DISCUSSION

This study quantified the effects of increased loading on tibiofemoral kinematics and tibiofemoral and patellofemoral contact characteristics following PS and CR implants. The most important finding of this study was that for greater loads across the joint and at increased flexion angles, there was greater femoral rollback and lower patellofemoral contact force and area in the CR implant compared to the PS implant. Additionally, there was a positive but not statistically significant relationship between increased joint loading across the patellofemoral and tibiofemoral joints and the contact force and contact area of the CR and PS implants. This study demonstrated that at higher flexion angles, there was greater femoral rollback in the CR implant compared to the PS design. At 90° KFA, the femur in the CR implant rested approximately 4.7 mm further posterior than in the PS implant. A functional PCL, which limits the posterior sag of the tibia immediately following implantation, contributes to the position of the tibia with the CR design. The PCL engages to translate the femur posteriorly, creating posterior femoral rollback throughout flexion.[15)]

Biomechanical and clinical studies on normal knee kinematics have shown that the femur typically translates posteriorly between 20 mm to 25 mm at maximal knee flexion and approximately 15 mm at 90° KFA.[16,17)] Therefore, this study's results, femoral posterior translation between 30 mm and 40 mm for KFAs over 60°, demonstrate how TKA implants alter knee kinematics and femoral rollback. The altered physical components by the PS and CR knee implants change the effects of weight-bearing across the joint. Li et al.[18)] reported that the CR implant type exhibits approximately 5.6 mm of posterior translation at 90° of KFA. Additionally, Khasian et al.[19)] showed that the PS implant type exhibited 5.4 mm posterior translation. Compared to the literature, this study described more posterior translation at 90° KFA; this could be due to the high weight demand modeled in the current study. Interestingly, in a biomechanical study, Victor et al.[20)] reported that increased loading reduced femoral translation in the native knee; however, the current study's results exhibit how femoral translation under high loading conditions is not predictable across various implant types compared to normal knee kinematics. Therefore, it is important to study the kinematic behaviors of TKA implant types under different weight-bearing conditions. This study's findings of greater femoral rollback at higher flexion angles for the CR over the PS design reflect the intact functionality of the PCL.

This study's findings revealed greater patellofemoral contact forces at all loading conditions and at various flexion angles for the PS implant compared to the CR implant. With a competent PCL, the femur in the CR implant showed greater femoral rollback than the PS implant in higher flexion, resulting in less force and contact area in the patellofemoral joint. There was a linear increase in force and pressure across the patellofemoral joint as weight-bearing loads increased. These results are consistent with other studies that evaluated the patellofemoral biomechanics after TKA, which found increases in peak pressure or shifting patella pressure distribution.[21,22)] Therefore, as overall weight-bearing demand increases, the concern for component failure increases. Patellofemoral complications have been an area of concern following TKA. Everyday activities and exercises involving larger knee flexion expose the patellofemoral joint to higher joint reaction forces than walking.[23)] As the average patient weight or weight-bearing demand increases, there is a concern that the pressures across the patellofemoral joint may reach threshold values for pain as well as possible patella component fracture. Other clinical studies have shown that there are no significant differences in functional or patient satisfaction scores between CR and PS total knee implants,[24-26)] yet this study demonstrated that extreme weight-bearing across the patellofemoral joint illuminates certain implant differences. This is a clinically relevant finding as we believe the PS implant design may not be best suited for individuals with high weight-bearing demands due to the heightened contact area and the contact force in the patellofemoral joint.

Tibiofemoral contact forces were lower in the PS than the CR implants for all loading conditions in 75° and 90° KFAs. Similar findings were recorded for kneeling, in which there was higher tibiofemoral contact force and area in the CR design than in the PS design.[16)] Although pressure measurements between the tibial post and femoral component were not obtained, the PS post is thought to mitigate some of the total load in the PS design.[27)] In the PS implant design, the femoral component is fashioned to engage the post on the tibial polyethylene liner as the knee flexes. Therefore, the tibiofemoral contact force of higher load demands is more concerning at higher flexion angles when the post and cam mechanism is engaged. Several studies have documented failure of the post mechanism leading to accelerated polyethylene wear, osteolysis, loosening, and polyethylene fracture.[27)] One should be cognizant of these potential risks of the PS design when used in high load demand populations as greater loads are translated to the tibial post, especially at mid and terminal KFAs.

There were several limitations in this biomechanical study. Although the tested knee specimens were within the typical age group of average TKA patients, they did not necessarily suffer knee osteoarthritis.[28] While it was a consideration that these cadavers do not clinically simulate the typical progression of soft-tissue compromises that characterize those in need of TKAs, it was to this study's advantage that these specimens had more consistent properties for a reproducible study. Another limitation of note was the inability to randomize the order of testing of the CR and PS implants and not further alter the bones and ligaments between different implant testing. However, due to the importance of using the same knee for testing both CR and PS, to directly compare the 2 implants, the CR was always tested first followed by the PS. While this may pose minor differences in our findings, the authors do not believe that there is a significant effect on the data as all testing performed was nondestructive. Only 1 manufacturer's prosthesis, the Corentec Lospa knee system, was included in the study, and therefore, the results may not extend to other available implants on the market. Certain manufacturers now offer highly conforming patellar components, which may counteract the increased forces. Lastly, the muscle loading forces used in this study were also less than those experienced in vivo, and therefore, the quantitative data, while simulated increased weight-bearing across the knee, may not be directly clinically applicable; however, the differences in the biomechanical and kinematic properties between implant choices could be extrapolated to higher loading conditions.

In this biomechanical study, CR TKAs showed less patellofemoral contact force, but more tibiofemoral contact force than PS TKAs. For higher loads across the joint and at increased flexion angles, there is significantly more posterior femur translation in the CR design with a preserved PCL and therefore significantly less patellofemoral contact area and force than in the PS design. The different effects of loading on implants are an important consideration for physicians as patients with higher load demands should consider the significantly greater patellofemoral contact force and area of the PS over the CR design.

CONFLICT OF INTEREST

No potential conflict of interest relevant to this article was reported.

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