Biomechanical Effects of Kneeling After Total Knee Arthroplasty

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Background: Kneeling following total knee arthroplasty can be a difficult task, impairing the activities of patients to varying degrees. Little is known about the biomechanical effects of kneeling following total knee replacement. The objective of this study was to quantify the effects of kneeling on patellofemoral joint contact areas and pressures, knee joint reaction force, and patellar kinematics.

Methods: Total knee arthroplasties were performed on eight fresh-frozen cadaveric knees, and they were tested with use of a custom knee jig, which permits the simulation of physiologic quadriceps loading as well as the application of an anterior force to simulate kneeling. The knees were tested at flexion angles of 90°, 105°, 120°, and 135° with no anterior force (mimicking a squatting position) and with an anterior force application simulating double-stance kneeling and single-stance kneeling. Patellofemoral joint contact areas and pressures were measured with pressure-sensitive film, and the knee joint reaction force was measured with use of a six-degree-of-freedom load cell. Patellar kinematics were assessed with use of digital photographs tracking fixed markers on the patella.

Results: Compared with the condition without kneeling, both single-stance and double-stance kneeling demonstrated significant increases in patellofemoral contact area ($p < 0.05$) and pressure at all flexion angles ($p < 0.05$), with the exception of double-stance kneeling at 135° of flexion. The resultant knee joint -reaction force increased with kneeling at all flexion angles. The compressive component of this force increased with kneeling for most conditions, while the lateral component of this force decreased significantly ($p < 0.05$) with kneeling only at 90°, and the anterior component of this force increased significantly at all knee flexion angles ($p < 0.05$). Overall, kneeling had minimal changes on patellar tilt, with significant changes in patellar tilt seen only with kneeling at 120° ($p = 0.02$ for double stance, and $p = 0.03$ for single stance).

Conclusions: The findings of this study suggest that kneeling at a higher flexion angle (135°) after total knee arthroplasty has a smaller effect on patellofemoral joint contact area and pressure than kneeling at lower flexion angles (≤120°).

Clinical Relevance: These findings suggest that if greater than 120° of knee range of motion can be achieved following total knee arthroplasty, kneeling may be performed with less risk than was previously believed to be the case.

Kneeling is an important function required for many activities, and the inability to kneel after knee surgery is a frequent cause of dissatisfaction. This is due to the fact that kneeling can be a painful experience for patients with degenerative conditions of the knee both before and after surgery, limiting the ability to perform the activities of daily living[1].

The biomechanical aspects of kneeling following total knee arthroplasty have been investigated recently. In a study of normal knees, Hassaballa et al.[2] measured kneeling at 90° of flexion and at full flexion. They found that the average percentage of body weight transmitted to the anterior aspect of the knee was 97% at 90° of flexion and 51% at full flexion. They suggested that improvements in the range of motion

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after arthroplasty may reduce the forces through the knee while kneeling. Palmer et al.[3] showed differences between the perceived and the actual ability to kneel after total knee arthroplasty, despite the fact that there was no difference with regard to the overall knee score or range of movement. They also documented radiographically two patterns of kneeling: *upright kneeling*, which occurs at 90° of flexion, and *flexed kneeling*, which occurs at >110° of flexion. In upright kneeling, the points of contact are the patella and the tibial tuberosity, while in flexed kneeling, only the tibial tuberosity bears weight. They concluded that kneeling in a flexed position (>110°) reduces the forces across the patellofemoral articulation. However, to our knowledge, there are no studies with quantitative evidence to support this conclusion. Therefore, the objective of the present study was to quantify the effects of kneeling on the patellofemoral joint contact area and pressure, knee joint reaction force, and patellar kinematics. We hypothesized that kneeling at higher flexion angles would result in decreased forces on the total knee arthroplasty components.

Materials and Methods

Eight fresh-frozen left cadaveric knees without gross deformity, from five male and three female donors who had ranged from seventy-five to ninety-two years old at the time of death, were used. A total knee arthroplasty was performed with use of the Foundation knee system (Encore Medical, Austin, Texas) following the recommended surgical protocol. The total knee arthroplasty was performed through a medial parapatellar arthrotomy. Intramedullary alignment devices were used to align the femoral and tibial surfaces. After the initial femoral and tibial bone resections were performed, flexion and extension gaps were measured and standard methods were used to correct any differences in the gaps[4]. The gap measurements were obtained with use of tongue depressors, which had a uniform thickness of 1.5 mm; when stacked medially and laterally, they provided an accurate means to measure both gaps separately. The number of tongue depressors needed to fill the gap was multiplied by the thickness to obtain the gap measurements. The tibial tray was sized to obtain maximal coverage of the cortical bone rim, and it was placed so that the center of the anterior ridge of the tray was aligned above the medial one-third of the tibial tuberosity. The patella was then resected with the cutting guide, with care taken to restore the thickness to within 1 mm. A dome-shaped patellar component was centered on the resected surface and was then positioned an additional 2 mm medially to improve patellar tracking and reduce the need for lateral retinacular release. The posterior cruciate ligament was resected, and a standard ultracongruent tibial insert, which incorporated a deep-dish design with a 12-mm-high anterior lip to prevent tibial subluxation, was used. After the arthroplasty components were implanted, the knees were taken through a full range of motion to assess soft-tissue balance. In three specimens, soft-tissue releases were performed to improve mediolateral stability in accordance with accepted clinical practices[4]. The arthrotomy was carefully closed to avoid overtensioning of the anteromedial retinaculum. The soft tissues were closed in anatomic layers.

Skin, subcutaneous tissue, muscles of the posterior aspect of the thigh, and muscles of the lower leg were removed, preserving the extensor muscles. The individual quadriceps muscles (vastus lateralis, vastus intermedius, rectus femoris, and vastus medialis) were identified and isolated along fascial planes to each tendinous insertion into the extensor mechanism. All of the muscle insertions were left intact and trimmed so that clamps of varying width could be used to apply forces uniformly across the entire width of the individual quadriceps muscles. The fibula was fixed with a single screw and was resected distal to the proximal tibiofibular joint, leaving the lateral collateral ligament insertion intact. The femur and the tibia were potted in polyvinyl chloride pipe with use of plaster of Paris.

The knees were mounted on a custom knee jig (Fig. 1) that permitted the simulation of physiologic quadriceps loading as well as the application of an anterior force to simulate kneeling. The polyvinyl chloride-mounted femur and tibia were secured in cylinders with use of eight fixation pins and were rigidly fixed with use of two bicortical, diaphyseal cross-bolts in each.

Each knee was then positioned at the starting point of 90° of flexion, and a custom guide was used to place a Kirschner wire through the epicondylar axis as a reference for the patellar tilt measurement. Two additional Kirschner wires were then placed parallel across the patella from lateral to medial to allow assessment of patellar kinematics.

The tendons of the isolated quadriceps muscles were trimmed to accommodate the width of the loading clamps and were wrapped with one layer of gauze to prevent slippage. During muscle-trimming, care was taken to select a portion of the muscle that was representative of the resultant force direction of all muscle fibers. For the vastus medialis and vastus lateralis, this was typically the distal portion of the muscle as it inserted into the patellar tendon and retinaculum. The vastus intermedius was trimmed symmetrically to simulate a resultant force vector along the anterior axis of the femur. Clamping of the muscles was made as close to their respective tendinous insertions as possible so that tendinous fibers could be incorporated within the clamp. This assisted in preventing the muscle tissue from slipping or pulling out from the clamp during loading. The vastus intermedius and rectus femoris were clamped together since the directions of the resultant force vectors of these muscles with respect to the patella are similar[5]. Clamping was performed such that the muscle fibers were perpendicular to the clamp itself. This ensured uniform loading of all muscle fibers.

The muscles were loaded under anatomically based conditions as described by Powers et al.[5]. Therefore, the vastus lateralis and vastus medialis each had a posteriorly oriented component of their loads, and the Q angle was maintained by loading the rectus femoris along the femoral axis. The direction of force for the vastus medialis was 40° medially in the frontal plane and 55° posteriorly in the sagittal plane (with

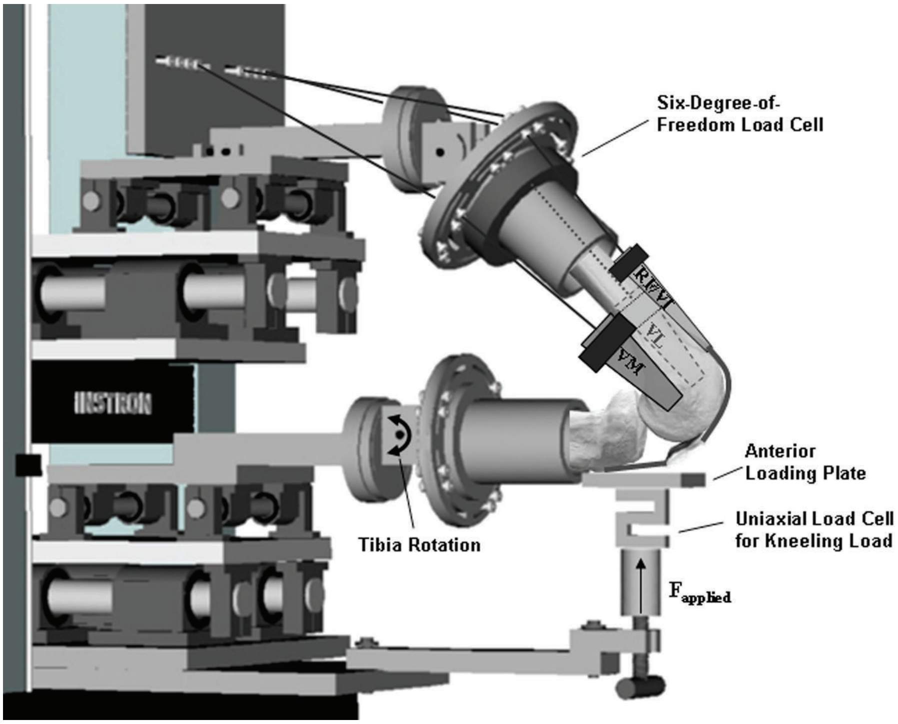

Fig. 1

Schematic of the knee testing system, including muscle-loading vectors for the vastus medialis (VM), vastus lateralis (VL), and rectus femoris-vastus intermedius (RF/VI), and the anterior loading device simulating kneeling at 135° of flexion.

respect to the long axis of the femur), while the force direction of the vastus lateralis was 35° laterally in the frontal plane and 55° posteriorly in the sagittal plane[5].

A total load of 276 N was used to load the isolated tendons by way of a cable and pulley system connected to the muscle clamps. The distribution of forces, which was based on the ratio of the physiological cross-sectional areas of each muscle as reported by Wickiewicz et al.[6], was 67 N for the vastus medialis, 98 N for the vastus lateralis, and 111 N for the vastus intermedius and rectus femoris.

Each knee was tested in four flexion angles: 90°, 105°, 120°, and 135°. A custom-designed kneeling plate connected to a uniaxial force transducer was used to apply an anterior load to the fixed knee specimen to simulate kneeling. With the knee at 90° of flexion, the plate contacted the patella and the tibial tubercle. As the knee flexed, the forces were transmitted primarily through the tibial tubercle. An average body weight of approximately 67 kg or 660 N was used to determine the force to apply to the knee. Each knee flexion angle had three anterior loading conditions: no anterior load (mimicking a squatting position), 330 N to simulate double-stance kneeling, and 660 N to simulate single-stance kneeling. Two trials of each condition were performed to ensure reproducibility. At each condition, patellofemoral contact area and pressure, knee joint reaction force, and patellar kinematics were measured.

The intra-articular contact areas and pressures were measured with use of pressure-sensitive film (Fuji Photo Film, Tokyo, Japan) sealed within thin waterproof sheets. The total thickness of the film and polyethylene sheets was approximately 250 μm. This has been reported to have a negligible effect on patellofemoral contact area and pressure[7]. Contact pressures for this study were within the range of the medium pressure-grade film (range, 9.8 to 49.0 MPa).

The tendons of the quadriceps musculature were tensioned to a cumulative constant load of 276 N for two minutes to obtain the patellofemoral joint contact pressure patterns. The Fuji film images were analyzed with use of NIH Image software (version 1.6; National Institutes of Health, Bethesda, Maryland). This program converts the Fuji film image into a scaled image with 256 levels of gray. The accuracy of Fuji film for area and pressure measurements is within 10% and 2%, respectively[8]. The scanned images were analyzed for total contact area (mm^2), peak contact pressure (MPa), and mean contact pressure (MPa). The mean contact pressure was defined as the average contact pressure of the entire contact area. The peak contact pressure was defined as the average of the top 10% of the contact area with the highest contact pressure.

A six-degree-of-freedom load cell connected to the femoral shaft was used to measure the knee joint reaction force (Fig. 1). The components of the force were recorded in the anterior-posterior, medial-lateral, and compression-distraction directions at each loading condition. Since the amount of total force applied to the knee changes with each kneeling

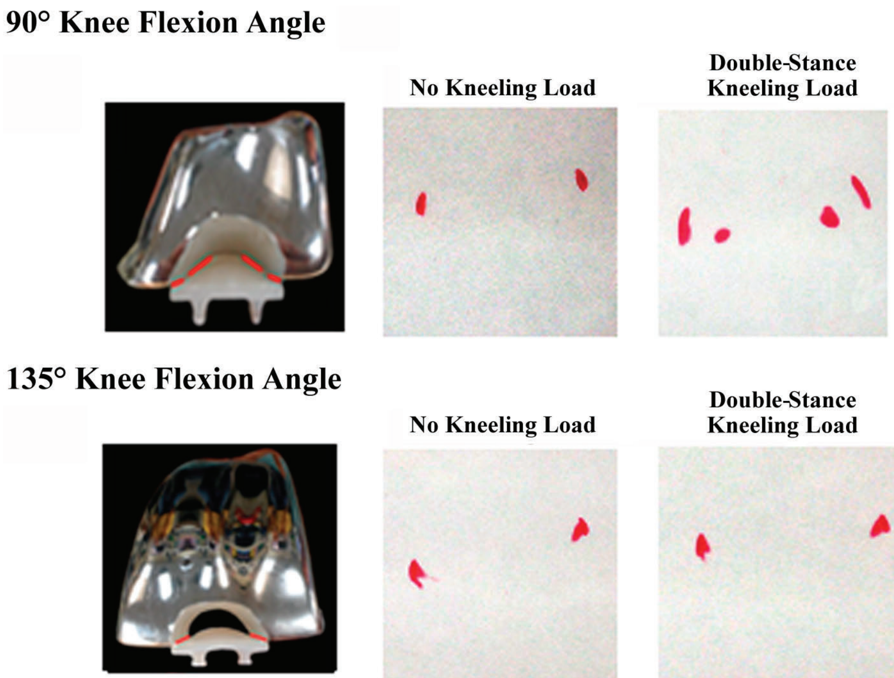

Fig. 2

Patellofemoral contact at 90° and 135° of knee flexion and the typical contact patterns on pressure-sensitive film made without kneeling and with double-stance kneeling.

condition, the forces were normalized by the total resultant force for comparison between kneeling conditions.

The patellofemoral, patellotibial, and patellar tilt angles were measured at each kneeling condition and each knee flexion angle. The angles were measured by means of digital photographs taken by two cameras mounted orthogonally to each other. One camera mounted on the front of the jig recorded photographs in the frontal plane to calculate patellar tilt, while another, mounted laterally, recorded photographs in the sagittal plane to calculate patellofemoral and patellotibial angles. The cameras were rigidly fixed to provide reproducible photographs. The digital images were analyzed with use of

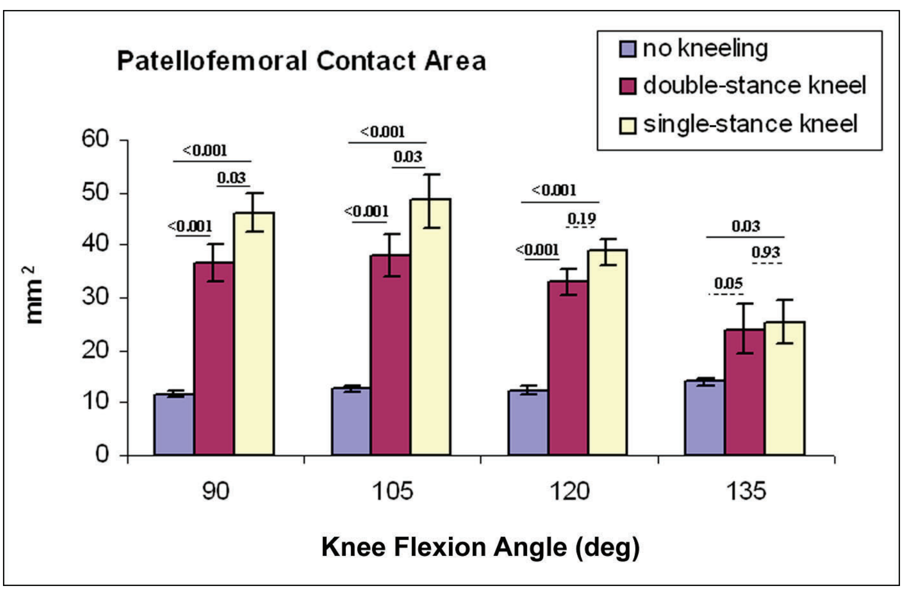

Fig. 3

Mean patellofemoral contact area (mm^2) (and standard error) for each knee flexion angle and loading condition with the p values for each comparison.

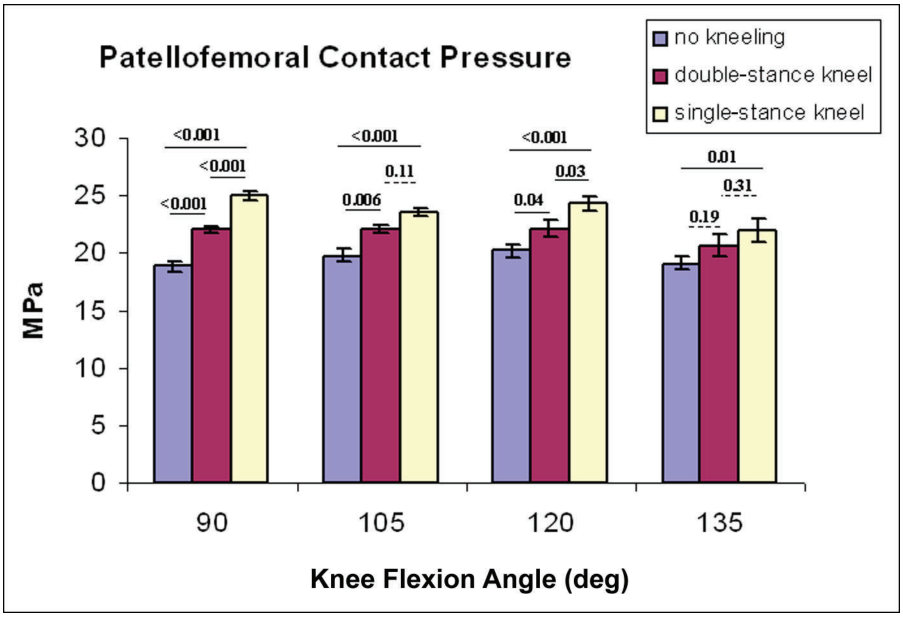

Fig. 4

Mean patellofemoral contact pressure (MPa) (and standard error) for each knee flexion angle and loading condition with the p value for each comparison.

Photoshop software (Adobe Systems, San Jose, California) by measuring the angles between lines drawn on the image, with the Kirschner wires placed along the epicondylar axis and with the patella used as a reference. This method was determined to be accurate within 0.2° and repeatable within 0.1°.

Significance was evaluated with use of a repeated-measures analysis of variance with post hoc comparisons with use of the Tukey honestly significant difference test with a confidence level of 0.05. Each specimen acted as its own control, with the position without kneeling functioning as the starting point. All data are reported as the mean and the standard error. The smallest p value reported is 0.001.

Results

Patellofemoral contact area showed significant increases with kneeling at 90°, 105°, and 120° of flexion ($p < 0.001$ for all angles). At 135° of flexion, only kneeling in the single-stance position resulted in a significant increase in patellofemoral contact area from the condition without kneeling ($p = 0.03$). Double-stance kneeling at 135° of knee flexion did not significantly increase patellofemoral joint contact area ($p = 0.05$) (Figs. 2 and 3).

Mean patellofemoral contact pressure showed significant increases with kneeling at flexion angles of 90°, 105°, and 120° ($p < 0.001$ and $p < 0.001$; $p = 0.006$, and $p < 0.001$; $p = 0.04$ and $p < 0.001$, respectively). At 135° of flexion, only kneeling in the single-stance position showed a significant increase in patellofemoral contact pressure from the condition without kneeling ($p = 0.01$). Double-stance kneeling at 135° of flexion did not significantly increase patellofemoral joint contact pressure ($p = 0.19$) (Fig. 4).

The resultant knee joint reaction force increased significantly at all flexion angles for each kneeling condition ($p < 0.05$ for all angles) (Table I). The normalized lateral component of the knee joint force decreased significantly at 90° with double-stance and single-stance kneeling compared with the condition without kneeling ($p < 0.05$) (Table II). There was a significant decrease in normalized posterior force or a significant increase in normalized anterior force at 90° of flexion with single-stance kneeling and at 105°, 120°, and 135° of flexion for both double-stance and single-stance kneeling ($p < 0.05$) (Table II). The normalized compressive component of the knee joint-reaction force increased significantly at 90° and 105°, and decreased significantly at 135° of flexion for each kneeling condition ($p < 0.05$) (Table II).

TABLE I Resultant Knee Joint Reaction Force at Each Flexion Angle and Kneeling Condition

	Resultant Force* *(N)*		
Knee Flexion	Without Kneeling	Double-Stance Kneeling	Single-Stance Kneeling
90°	169.6 (7.4)	377.9 (13.3)†	584.9 (24.5)†
105°	170.2 (4.5)	403.9 (20.1)†	608.4 (29.5)†
120°	176.9 (2.2)	412.2 (18.5)†	616.7 (12.4)†
135°	180.9 (4.4)	383.8 (10.8)†	565.5 (15.7)†

*The values are given as the mean with the standard error in parentheses. †Compared with the condition without kneeling, the difference is significant ($p < 0.05$).

TABLE II Knee Joint Reaction Force Components as a Percentage of the Resultant Force for Each Flexion Angle and Kneeling Condition

	Lateral Force as Percentage of Resultant Force*			Posterior Force as Percentage of Resultant Force*			Compression as Percentage of Resultant Force*		
Knee Flexion	Without Kneeling	Double-Stance Kneeling	Single-Stance Kneeling	Without Kneeling	Double-Stance Kneeling	Single-Stance Kneeling	Without Kneeling	Double-Stance Kneeling	Single-Stance Kneeling
90°	7.2 (2.1)	2.1 (1.2)†	1.2 (1.0)†	17.6 (1.2)	16.2 (1.0)	12.3 (1.1)†	98.0 (0.2)	98.6 (0.2)†	99.2 (0.1)†
105°	6.6 (2.1)	4.0 (0.7)	3.1 (0.7)	18.9 (1.0)	3.1 (2.7)†	−1.4 (2.6)†	97.8 (0.3)	99.6 (0.1)†	99.7 (0.08)†
120°	6.5 (1.8)	4.8 (1.0)	4.5 (0.8)	18.9 (1.6)	−9.7 (3.1)†	−17.0 (2.3)†	97.8 (0.4)	99.0 (0.3)	98.2 (0.4)
135°	5.0 (1.7)	6.0 (0.8)	6.3 (0.7)	14.2 (2.0)	−22.3 (2.0)†	−29.4 (1.9)†	98.6 (0.3)	97.1 (0.5)†	95.2 (0.6)†

*The values are given as the mean with the standard error in parentheses. †Compared with the condition without kneeling, the difference is significant ($p < 0.05$).

Significant changes in patellar tilt were found with kneeling at 120° of flexion, where the patella tilted laterally a mean of 1.1° ± 0.4° ($p = 0.02$) and 1.0° ± 0.4° ($p = 0.03$) with double-stance and single-stance kneeling, respectively. The patellofemoral angle increased significantly at all flexion angles between the condition without kneeling and the double-stance and single-stance kneeling conditions ($p < 0.05$). There was also a significant increase in the patellofemoral angle between single-stance and double-stance kneeling at flexion angles of 90° ($p = 0.001$) and 105° ($p = 0.01$). The changes in patellotibial angle were reciprocally related to changes in patellofemoral angle (Fig. 5).

Discussion

Our hypothesis that kneeling at higher flexion angles would result in decreased forces on the total knee arthroplasty components was proven since the present study showed that kneeling at a higher flexion angle (135°) after total knee arthroplasty has a smaller effect on patellofemoral joint contact area and pressure than kneeling at lower flexion angles (≤120°). The patellofemoral contact patterns, areas, and pressures in our study are consistent with those in previous studies[9-12]. With the knee at 90° of flexion, patellofemoral contact separated into two regions close to the medial and lateral margins[10,12]. A remarkable finding in this study was that double-stance kneeling at a high flexion angle (135°) did not significantly increase patellofemoral contact area or pressure. This lack of a significant increase in patellofemoral contact area and pressure is most likely due to the fact that, when kneeling in high degrees of flexion, the loading point is primarily on the tibial tubercle and not the patella[3]. However, it is important to recognize that the peak patellofemoral joint contact pressures observed in this study for high knee flexion angles and kneeling exceeded the yield strength of ultra-high molecular weight polyethylene,

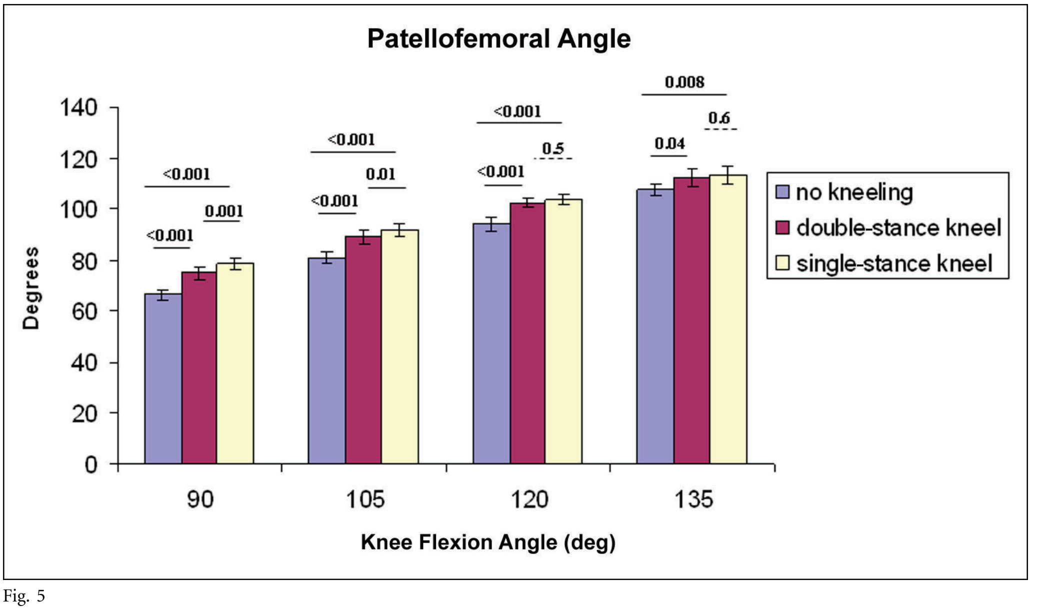

Fig. 5
Mean patellofemoral angle (and standard error) for each knee flexion angle and loading condition with the p value for each comparison.

which has been reported to be 14.4 MPa[13]. This may have implications with respect to increased polyethylene wear with kneeling, which may be a limiting factor in the survival of total knee arthroplasty components[14,15]. Our results for the changes in patellofemoral and patellotibial angle with respect to knee flexion angles are consistent with those in the study by Lee et al.[16], who stated that the primary predictor of patellofemoral and patellotibial angles was the knee flexion angle. Other authors have also shown that the mean patellar angle increases with knee flexion for dome-shaped patellae[17]. With kneeling, a smaller change in patellofemoral angle was observed at 135° of knee flexion. This also supports the finding that there is less anterior loading on the patella with kneeling at high flexion angles.

A limitation of the study was that it was performed with a cruciate-substituting prosthesis from a single manufacturer; other total knee arthroplasty systems may produce different results. Additional limitations include the inability of this experimental model to duplicate the native so-called screw-home mechanism and to account for other muscular forces and gait differences that may be experienced in vivo. Also, although the muscle forces applied to the quadriceps were based on the physiological muscle cross-sectional area ratios, the muscle forces were not adjusted for different flexion angles and/or kneeling condition. Finally, this model did not include hamstring muscles, which would be important in evaluating the tibiofemoral joint parameters; however, doing so would have a constant effect on the patellofemoral joint parameters and would not change the results for comparisons across kneeling conditions.

This study provides new information on the effects of kneeling at high knee flexion angles following total knee arthroplasty. The ability to kneel after total knee arthroplasty remains a problem that compromises the success of the procedure for some patients. With double-stance kneeling, there was no significant increase in patellofemoral joint contact area and pressure at 135° of flexion. These findings suggest that if greater than 120° of knee range of motion can be achieved following total knee arthroplasty, kneeling may be performed with less risk than was previously believed to be the case. ■

NOTE: The implants used in this study were donated by Encore Medical.

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