INTRODUCTION:
Kneeling is an activity that is important in certain occupations and recreational activities. Culturally, kneeling plays a large role in Middle Eastern and Asian countries, where activities of daily living require the ability to kneel and achieve deep knee flexion.

It has been shown that kneeling at higher knee flexion angles after total knee arthroplasty (TKA) has a smaller effect on patellofemoral joint contact area and pressure than kneeling at lower flexion angles. However, despite the importance of the posterior cruciate ligament (PCL) for resisting posterior translation, the effects of kneeling in posterior cruciate ligament retaining (CR) and substituting (PS) TKA have not been determined. Therefore, the objective of this study was to quantify the tibiofemoral contact areas, pressures, and kinematics with kneeling in CR and PS TKA.

METHODS:
Five cadaveric knees (3 Male, 2 Female) average age 84 years old (range 78-93) were dissected of all skin and subcutaneous tissue, leaving the extensor mechanism, hamstrings, joint capsule, and ligaments intact. The femur was potted in PVC pipe with the femoral epicondylar axis aligned perpendicular to the tibial shaft in the coronal plane. Posterior slope was determined by the femoral epicondylar axis and the PCL was initially preserved in all specimens.

CR and PS TKA were performed using the Encore Foundation System (Encore Medical Corp, Austin, TX) according to the manufacturer’s protocol. The knee was first exposed with a medial parapatellar arthrotomy. The distal femur cut was made with intramedullary guidance. Posterior condylar referencing was used to set the rotation of the femoral component, utilizing an intramedullary axis for additional reference. Using intramedullary guidance the tibia was cut perpendicular to the tibial shaft in the coronal plane. Posterior slope was set to 7° and the PCL was initially preserved in all specimens.

The medial and lateral joint gaps were measured in flexion and extension and appropriate soft tissue releases were performed to balance all gaps to within 1mm of each other. Trial components were inserted to ensure proper balancing and component size. The final CR TKA components were then implanted and cemented in place with plaster of Paris. The CR TKA specimens were then subjected to the testing protocol. Thereafter, the femoral component was carefully removed and the femoral cutting guide for the PS TKA system was utilized in order to cut the femoral box. The PS component was then cemented in place, and a PS tibial polyethylene insert was inserted. The PS TKA specimens were then subjected to the testing protocol.

After TKA was performed, the knees were securely mounted on a custom knee testing system that allowed physiologic muscle loading and the application of a posteriorly directed force to simulate kneeling (Fig. 1). The femoral epicondylar axis was aligned parallel to the coronal plane of the jig and locked in place. The tibia remained with 5 degrees of freedom during the experiment.

Anatomically based multi-plane loading of the quadriceps mechanism (v. medialis 51N, rectus femoris v. intermedius 87N, v. lateralis 77N) and hamstrings (biceps femoris 31N, semimembranosus/semitendinosus 54N) was used. The knees were tested under no anterior loading, simulated double stance kneeling (339N anterior force, corresponding to 50% mean body weight (MBW) of a 70kg person), and simulated single stance kneeling (678N) at a flexion angle of 90°, 105°, 120°, and 135°. The anterior load was applied to the tibial axis with a load plate attached to a uniaxial force transducer S load cell (Omega Inc).

The anterior load was applied to the tibial axis with a load plate attached to a uniaxial force transducer S load cell (Omega Inc). Tibiofemoral joint contact areas and pressures were measured using K-Scan (Tekscan Inc, South Boston, MA) sensors inserted through the posterior capsule (Fig. 2). Repeated measures ANOVA was used for statistical analysis (p<0.05).

RESULTS:
Double and single stance kneeling significantly increased contact areas in both designs (p<0.05). Double and single stance kneeling increased pressures compared to no kneeling with variable significance in both groups (p<0.05) (Figure 3 and 4). Moving from double to single stance kneeling tended to increase pressures in the CR group, but decreased pressures in the PS group. The CR group had significantly larger contact areas than the PS group, although no significant difference in pressures were observed (p>0.05).

Both designs exhibited tibial external rotation with kneeling, though the CR group exhibited more variability between specimens. With the initial 339N of force, the PS design exhibited significantly more posterior translation than with the second 339N (p<0.05). At 90° the PS knees exhibited significantly less posterior translation than at higher flexion angles (p<0.05). Both of these trends were observed in the CR group but were not significant.

DISCUSSION:
Increased contact areas and pressures with kneeling indicate a potential for increased polyethylene wear. The smaller contact areas in the PS group likely result from contact being partly assumed by the cam-post articulation, causing similar contact pressures at the articulation of the condyles and tibial polyethylene surface between the groups. In the PS design, the higher stresses of single stance kneeling are likely placed on the cam-post articulation.

Kneeling produces more consistent tibial external rotation with a PS design likely due to more variable biomechanics of the native PCL in the CR group. Both designs limit posterior translation with kneeling most effectively at 90° than at the higher flexion angles, possibly due to the biomechanics of the PCL at this flexion angle in the CR group and improved cam-post interaction in the PS group.

REFERENCES:
1. Wilkens et al. JBJS 2007

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