



**Fig. 2.** The Alpina prosthesis. This device has a spherical-shaped post, whereas the post-cam mechanism conforms fully.

also determined. The distance from the deepest part of the tibial articular surface to the center of the contact area was defined as the contact location. Then, contact area and contact stress were measured again while lifting both medial and lateral femoral condyles 3 mm from the tibial articular surface at 120° of flexion using the TRAC, Alpina, PFC sigma RPF, and Scorpio NRG components.

The contact areas, mean and peak contact stresses, and contact locations were compared between the prostheses at each flexion angle; contact area and stresses with and without femoral condylar lift-off at 120° of flexion were also compared. Statistical significance was evaluated by analysis of variance and a post hoc Scheffe *f* test. Values of *P* less than .05 were regarded as significant.

## Results

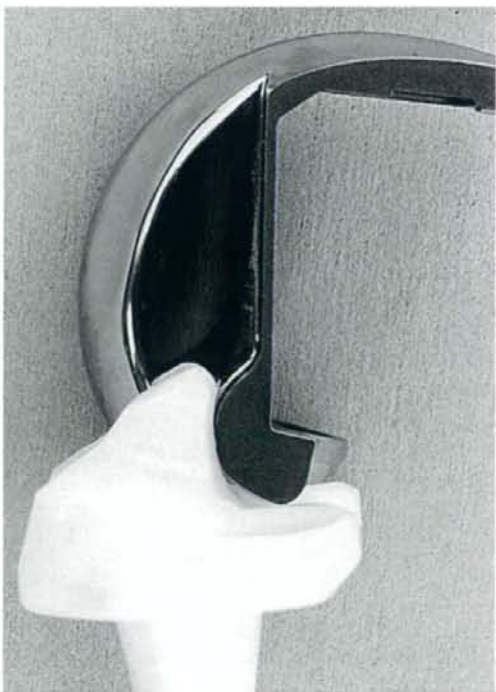
### Contact Area

The TRAC and Alpina components exhibited the largest contact areas at 90° of 67.7 and 73.9 mm<sup>2</sup>, respectively; these values were significantly larger

than those of the other prostheses ( $P < .001$ ) (Fig. 5). The TRAC and Alpina units had decreased contact areas at 120° and 150°, although these components still had larger contact areas than the others at 120° ( $P < .001$ ). The contact areas of the PFC sigma RPF increased with flexion, then the PFC sigma RPF had the largest contact area of all of the prostheses at 150° (81.6 mm<sup>2</sup>) ( $P < .001$ ). The Scorpio NRG had smaller contact area than the other units. The Scorpio NRG showed the smallest contact area of all of the units at 120° (30.0 mm<sup>2</sup>) ( $P < .001$ ), although displayed increased contact area at 150°.

### Mean Contact Stress

Mean contact stress demonstrated an opposite trend as contact area. The TRAC and Alpina units had contact stresses at 90° of 6.7 and 7.3 Mpa, respectively, which were significantly lower than those of the other prostheses ( $P < .001$ ) (Fig. 6). The TRAC and Alpina components exhibited increased contact stresses at 120° and 150°. These units still had lower contact stresses than the other units at



**Fig. 3.** The PFC sigma RPF prosthesis. The cam is in a distally apex shape, with a relatively flat surface from the posterodistal part to the posterior part.



**Fig. 4.** The Scorpio NRG prosthesis. The cam of this unit has oval shape and a relatively large radii of curvature posteriorly.

120° ( $P < .001$ ). The PFC sigma RPF exhibited decreased contact stress with increasing flexion, displaying the lowest contact stress of all of the units (6.1 MPa) at 150° ( $P < .001$ ). The Scorpio NRG demonstrated a similar stress pattern as the PFC sigma RPF, although the contact stresses were higher than the other prostheses at 120° (16.7 MPa) ( $P < .001$ ).

#### Peak Contact Stress

The peak contact stress from 17 to 47 MPa was 2- to 3-fold as high as the mean contact stress at the same angle of flexion (Fig. 7). At 90° and 120°, the TRAC and Alpina components had the lowest contact stresses, measuring less than 18 MPa, which was significantly lower than those of the other units ( $P < .001$ ). At 150°, the PFC sigma RPF exhibited the lowest contact stress of 19.8 MPa, which was significantly lower than those of the other prostheses ( $P < .001$ ). The peak contact stress of the Scorpio NRG increased at 120° to the highest contact stress measured for all of the prostheses ( $P < .001$ ); the stress measurement, however, decreased at 150°.

#### Contact Location

Significant differences were observed in the contact locations of each prosthesis at each angle of flexion (Fig. 8). The TRAC component exhibited the highest contact location at 90° and 150° ( $P < .001$ ). At 90°, the contact location of the TRAC was 13.1 mm, moving to 14.2 and 14.6 mm at 120° and 150°, respectively. The contact location of the Alpina at both 90° and 120° was 9.7 mm, moving to 8.7 mm at 150°. The contact location of the Scorpio NRG peaked at midflexion, with measurements of 12.0, 15.1, and 13.1 mm at 90°, 120°, and 150°, respectively. In contrast, the contact location of the PFC sigma RPF bottomed at the midflexion, and this device possessed the lowest contact locations at all angles of flexion ( $P < .001$ ). The contact location of the PFC sigma RPF was 7.1, 6.4, and 7.3 mm at 90°, 120°, and 150°, respectively.

#### Effect of Femoral Condylar Lift-Off

With a femoral condylar lift-off of 3 mm, the contact area of the TRAC decreased significantly from 61.3 to 44.2 mm<sup>2</sup>, whereas the Alpina decreased from 58.7 to 36.5 mm<sup>2</sup> (Fig. 9). The mean and peak contact stresses of these components with lift-off exhibited significant increases from those of the original measured positions (Figs. 10 and 11). The Scorpio NRG demonstrated slight decrease in contact area and slight increase in the mean contact stress with lift-off (Figs. 9-11). The PFC sigma RPF unit did not show any significant changes in either contact area or stress.

#### Discussion

The post-cam mechanism plays one of the most important roles in knee flexion after PS-TKA, facilitating motions similar to normal knees, such as femoral rollback. Recently, complications of the post-cam mechanism, such as fracture, severe wear of the post, and dislocation of the knee have been reported [4-9]. In a retrieval analysis of 23 components that had been implanted for a mean of 35.6 months, Puloski et al [9] concluded

**Table 1.** Design Feature of the Prostheses

|                    | TRAC  | Alpina | PFC sigma RPF | Scorpio NRG |
|--------------------|-------|--------|---------------|-------------|
| Design of the post | Round | Round  | Curve         | Curve       |
| Design of the cam  | Round | Round  | Curve         | Oval        |
| Conformity         | Full  | Full   | Partial       | Partial     |



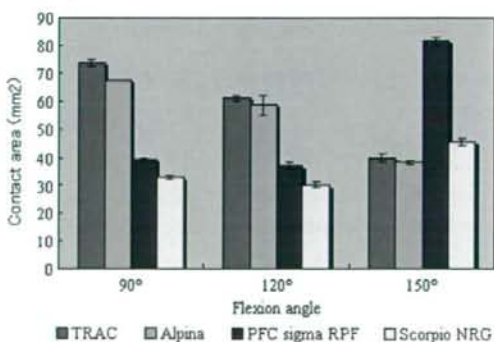
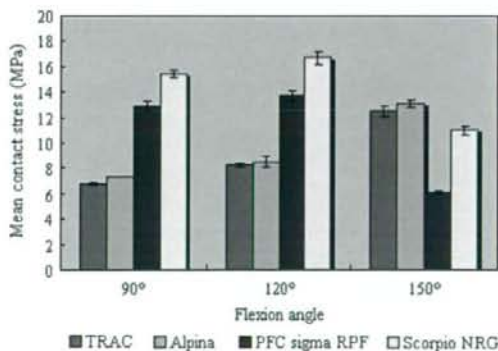
**Table 2.** Size Variation of the Prostheses

| Size | TRAC             |                 | Size | Alpina           |                 | Size | PFC sigma RPF    |                 | Size | Scorpio NRG      |                 |
|------|------------------|-----------------|------|------------------|-----------------|------|------------------|-----------------|------|------------------|-----------------|
|      | Post height (mm) | Post width (mm) |      | Post height (mm) | Post width (mm) |      | Post height (mm) | Post width (mm) |      | Post height (mm) | Post width (mm) |
| 55   | 18.3             | 15.2            | 1    | 15               | 12              | 1    | 15.5             | 14.1            | 3    | 18.8             | 16.3            |
| 60   | 18.3             | 17.3            | 3    | 16               | 13              | 1.5  | 15.8             | 14.1            | 4    | 18.8             | 16.3            |
| 65   | 18.3             | 17.3            | 5    | 17               | 14              | 2    | 17.6             | 14.1            | 5    | 18.8             | 16.3            |
| 70   | 18.3             | 17.3            | 7    | 18               | 14.5            | 2.5  | 17.9             | 14.1            | 6    | 18.8             | 16.8            |
| 75   | 18.3             | 17.3            | 9    | 19               | 15              | 3    | 17.9             | 14.1            | 7    | 18.8             | 16.8            |
|      |                  |                 | 11   | 20               | 16              | 4    | 21.7             | 16.3            | 8    | 18.8             | 16.8            |
|      |                  |                 | 13   | 21               | 17              |      |                  |                 | 9    | 18.8             | 16.8            |

that the post-cam articulation could be an additional source of polyethylene wear debris. The variability in wear patterns observed among the different designs may be due to differences in the post-cam configuration. Mauerhan [6] reported 5 patients with fractures of the post causing subluxation of the femur. Of the 5 patients, 4 exhibited more than 130° flexion after PS-TKA; these patients could perform activities that involved kneeling and squatting. Although anteroposterior force on the tibiofemoral joint is not significantly increased during the gait cycle, biomechanical studies have demonstrated that the anteroposterior shear force at this joint is increased during deep knee flexion [11-16].

In this study, the 4 different post-cam designs exhibited significant differences in the contact area and contact stress at each flexion angle. The TRAC and Alpina had fully conforming surfaces in the sagittal plane (Figs. 1 and 2), displaying the same trend in contact area and contact stress with increasing angles of flexion. Although these devices showed the largest contact area and lowest contact stress from 90° to 120°, the contact area decreased significantly at 150°. The cams of both the TRAC and

Alpina units have round shapes with large radii of curvature; as the flexion angle increases, the cam partially loses contact with the post surface. These findings may suggest that the post-cam shape of the TRAC and Alpina components would exhibit better wear performance in daily activities, such as standing up from a chair or climbing up and down stairs, but are not suitable for squatting and kneeling activities that require greater than 120° of knee flexion. The post-cam mechanism of the PFC sigma RPF unit conformed partially in the sagittal plane (Fig. 3). This device increased contact area and decreased contact stress with increasing knee flexion, exhibiting the largest contact area and lowest contact stress at the 150° angle. The cam of the PFC sigma RPF is in a curved shape, with increasing radii of curvature from the distal part to the posterior part (Fig. 3), which results in almost full conformity between the post and cam at 150°. These findings suggest that the post-cam shape of the PFC sigma RPF components would be suitable for movements requiring 150° of knee flexion, such as squatting. The Scorpio NRG had less of conforming post-cam mechanism in the sagittal plane (Fig. 4). The cam of the Scorpio NRG has oval shapes with a

**Fig. 5.** Contact area (mm<sup>2</sup>) of each component.**Fig. 6.** Mean contact stress (MPa) of each component.

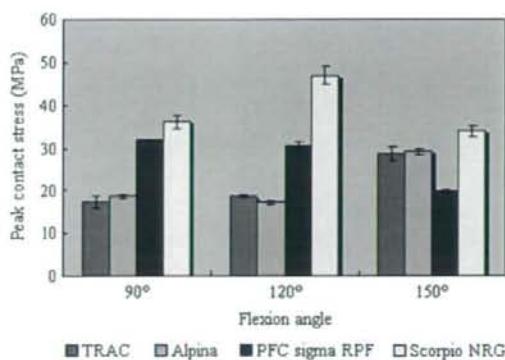


Fig. 7. Peak contact stress (MPa) of each component.

relatively large radii of curvature posteriorly. This design feature is congruent with the high values of contact stress in comparison to other prostheses. However, none of the prostheses examined exhibited constantly low contact stresses, the permissible compressive yield strength of UHMWPE (10 MPa) [30,31], throughout flexion.

Contact location is determined by multiple factors, including the shape of the cam, the position of cam attachment to the femoral component, and the curvature of the posterior femoral condyle. Higher contact point on the post is a risk factor of dislocation of the cam when abnormal motion occurs [23]. The contact location of the PFC sigma RPF remained lower than the other prosthesis throughout flexion, a feature that is beneficial for avoiding excessive stresses at the bone-implant interface and for preventing dislocation of the cam or fracture of the post. The total movements of the contact location during flexion for the TRAC, Alpina, and PFC sigma RPF devices were 1.5, 1.4, and 1.6 mm, respectively.

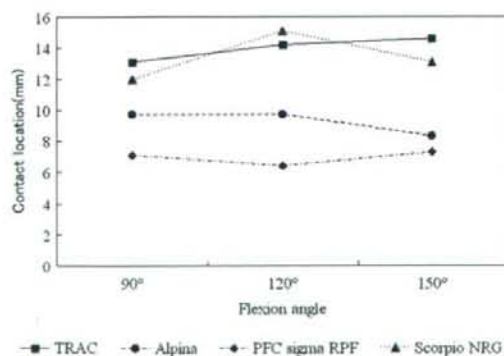


Fig. 8. Distance (mm) from the center of the contact area to the upper surface of the tibia.

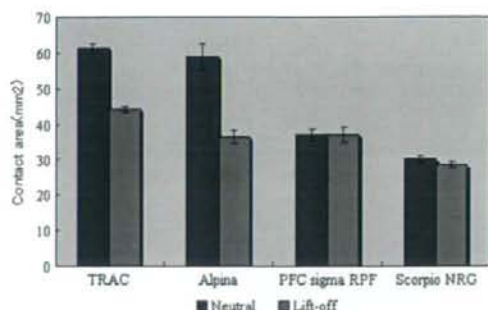


Fig. 9. Contact area ( $\text{mm}^2$ ) of each component with neutral and 3 mm lift-off of the femoral component at 120° of flexion.

which were smaller values than that of the Scorpio NRG (5.1 mm). The reduced movement of the contact location could be advantageous in preventing delamination of the polyethylene and in avoiding excessive wear. The concentration of contact stress in one spot, however, generates creep deformation, jeopardizing the longevity of the polyethylene [32,33]. It is necessary to consider both the sliding velocity and the contact stress when predicting polyethylene wear [34]. Therefore, it will be necessary to investigate the components under dynamic conditions to evaluate the effect of contact location on polyethylene wear.

The effect of femoral condylar lift-off on contact stress is also important in investigating the post-cam mechanism. Numerous kinematic analyses of the knee have demonstrated that the femoral condylar lift-off can occur during both normal gait and deep knee flexion [23-26]. Dennis et al [24] performed a fluoroscopic analysis of 20 patients after PS-TKA (PFC design; Johnson & Johnson, Raynham, Mass), revealing that femoral condylar lift-off was observed at some increment of knee flexion in 80% PS-TKA.

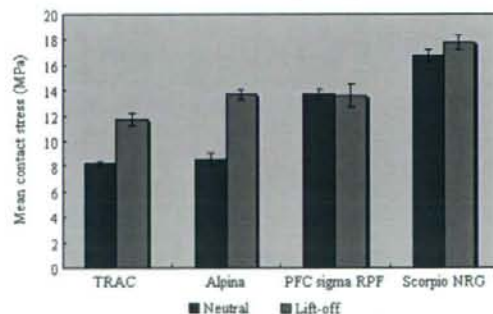


Fig. 10. Mean contact stress (MPa) of each component with neutral and 3 mm lift-off of the femoral component at 120° of flexion.



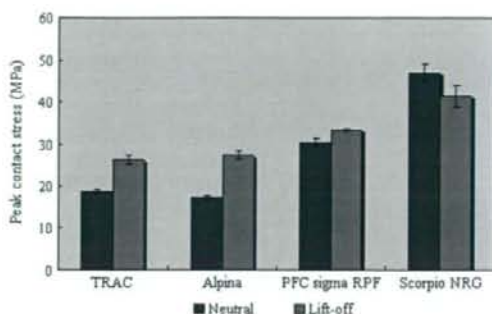


Fig. 11. Peak contact stress (MPa) of each component with neutral and 3 mm lift-off of the femoral component at 120° of flexion.

The maximum value for lift-off was 2.7 mm at 90° flexion. Argenson et al [23] also performed a kinematic analysis of 20 patients who achieved an average weight-bearing range of motion of 125° after PS-TKA (LPS Flex; Zimmer, Warsaw, Ind). They reported 40% of patients experienced femoral lift-off, with a maximum value of 2.5 mm at maximum knee flexion. We therefore examined the contact stress of the post-cam interface under the conditions of 3 mm femoral condylar lift-off at 120° flexion in comparison to neutral setting. Our study demonstrated that the TRAC and Alpina units exhibited increased contact stress and decreased contact area with 3 mm of lift-off. Of all of the prostheses, the TRAC and Alpina displayed the greatest increments of contact stress with lift-off. The configuration of the post-cam in these prostheses conformed to a greater extent than the other prostheses in the sagittal plane (Figs. 1 and 2). Therefore, these results indicated that vertical motion of the femoral component, such as that seen during lift-off, might lose conformity at the post-cam interface, resulting in increased contact stress. In contrast, the PFC sigma RPF and Scorpio NRG prostheses did not exhibit remarkable changes in either contact area or stress with lifting up of the femoral component. The post-cams of these units were less constrained in the proximal-distal direction (Figs. 3 and 4). This design feature avoids excessive stresses between the polyethylene insert and the tibial baseplate during lift-off of the femoral component.

One of the limitations of this study is that the tests were performed under static conditions at only 3 angles. It remains possible that the post-cam designs examined in this study may exhibit their highest contact stresses at angles other than 90°, 120°, or 150° of flexion. Studies examining these units under

dynamic conditions are necessary to assess the force distribution of the post-cam mechanism. Another limitation of this study is that only a relatively low force (500 N) [11,15,16] was used; forces applied to the tibiofemoral articular surface were ignored. This force was determined by previous biomechanical studies. Large force would result in high contact stress, but tendency of the data would not change significantly. The prostheses used in this study has 5 to 7 size variations, but the size of the post does not increase as the tibial size goes up in 3 of 4 prostheses (Table 2). We did not completely match the A-P and M-L size of the components, but the second or third smallest size was used. Therefore, we believe, the size difference would minimally affect the results. Three millimeters of lift-off motion was chosen based on kinematic data of previous studies [23,24] because we currently do not have kinematic data of all the prostheses we tested. However, there is the possibility that each of the prostheses might have different lift-off motions. This is another limitation of this study. Regardless of these shortcomings, we believe that this study reveals important information about the post-cam contact mechanism.

In conclusion, recent modifications of the post-cam designs used in PS-TKA have increased the contact area and conformity, which may contribute to a lower contact stress during deep knee flexion. In this study, however, none of the prostheses examined exhibited consistently low contact stresses throughout flexion. In addition, the contact stresses of the post-cam interface increased significantly with femoral condylar lift-off in a number of the prostheses. Further modification of the post-cam mechanism is necessary to provide lower risk for polyethylene failure in the knee with over flexion.

## References

1. Insall JN, Hood RW, Flawn LB, et al. The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements. *J Bone Joint Surg Am* 1983;65:619.
2. Colizza WA, Insall JN, Scuderi GR. The posterior stabilized total knee prosthesis. Assessment of polyethylene damage and osteolysis after a ten-year-minimum follow-up. *J Bone Joint Surg Am* 1995; 77:1713.
3. Rodriguez JA, Bhende H, Ranawat CS. Total condylar knee replacement: a 20-year followup study. *Clin Orthop* 2001;388:10.
4. Callaghan JJ, O'Rourke MR, Goetz DD, et al. Tibial post impingement in posterior-stabilized total knee arthroplasty. *Clin Orthop* 2002;404:83.

5. Chiu YS, Chen WM, Huang CK, et al. Fracture of the polyethylene tibial post in a NexGen posterior-stabilized knee prosthesis. *J Arthroplasty* 2004;19:1045.
6. Mauerhan DR. Fracture of the polyethylene tibial post in a posterior cruciate-substituting total knee arthroplasty mimicking patellar clunk syndrome: a report of 5 cases. *J Arthroplasty* 2003;18:942.
7. Mestha P, Shenava Y, D'Arcy JC. Fracture of the polyethylene tibial post in posterior stabilized (Insall Burstein II) total knee arthroplasty. *J Arthroplasty* 2000;15:814.
8. Mikulak SA, Mahoney OM, dela Rosa MA, et al. Loosening and osteolysis with the press-fit condylar posterior-cruciate-substituting total knee replacement. *J Bone Joint Surg Am* 2001;83-A:398.
9. Puloski SK, McCalden RW, MacDonald SJ, et al. Tibial post wear in posterior stabilized total knee arthroplasty. An unrecognized source of polyethylene debris. *J Bone Joint Surg Am* 2001;83-A:390.
10. Sultan PG, Most E, Schule S, et al. Optimizing flexion after total knee arthroplasty: advances in prosthetic design. *Clin Orthop* 2003;416:167.
11. Dahlkvist NJ, Mayo P, Seedhom BB. Forces during squatting and rising from a deep squat. *Eng Med* 1982;11:69.
12. Li G, Most E, Otterberg E, et al. Biomechanics of posterior-substituting total knee arthroplasty. An in vitro study. *Clin Orthop* 2002;404:214.
13. Li G, Zayontz S, DeFrate LE, et al. Kinematics of the knee at high flexion angles. An in vitro investigation. *J Orthop Res* 2004;1:90.
14. Li G, Zayontz S, Most E, et al. In situ forces of the anterior and posterior cruciate ligaments in high knee flexion: an in vitro investigation. *J Orthop Res* 2004;2:293.
15. Nagura T, Dyrby CO, Alexander EJ, et al. Mechanical loads at the knee joint during deep flexion. *J Orthop Res* 2002;4:881.
16. Wilk KE, Escamilla RF, Fleisig GS, et al. A comparison of tibiofemoral joint forces and electromyographic activity during open and closed kinetic chain exercises. *Am J Sports Med* 1996;24:518.
17. Nakayama K, Matsuda S, Miura H, et al. Contact stress at the post-cam mechanism in posterior-stabilized total knee arthroplasty. *J Bone Joint Surg Br* 2005;87-B:483.
18. D'Lima DD, Hermida JC, Chen PC, et al. Polyethylene wear and variations in knee kinematics. *Clin Orthop Relat Res* 2001;124030.
19. Currier JH, Duda JL, Aperling DK, et al. In vitro simulation of contact fatigue damage found in ultra-high molecular weight polyethylene components of knee prostheses. *Proc Inst Mech Eng* 1998;212:293.
20. Bartel DL, Rawlinson JJ, Burstein AH, et al. Stresses in polyethylene components of contemporary total knee replacements. *Clin Orthop Relat Res* 1995:76.
21. D'Lima DD, Chen PC, Colwell Jr CW. Polyethylene contact stresses, articular congruity, and knee alignment. *Clin Orthop* 2001;392:232.
22. Ishikawa H, Fujiki H, Yasuda K. Contact analysis of ultrahigh molecular weight polyethylene articular plate in artificial knee joint during gait movement. *J Biomech Eng* 1996;118:377.
23. Argenson JNA, Scuderi GR, Komistek RD, et al. In vivo kinematic evaluation and design considerations related to high flexion in total knee arthroplasty. *J Biomech* 2005;38:277.
24. Denis DA, Komistek RD, Walker SA, et al. Femoral condylar lift-off in vivo in total knee arthroplasty. *J Bone Joint Surg Br* 2001;83-B:32.
25. Nilsson KG, Karrholm J. Increased varus-valgus tilting of screw-fixed knee prostheses: stereoradiographic study of uncemented versus cemented tibial components. *J Arthroplasty* 1993;8:529.
26. Wasielewski RC, Galat DD, Komistek RD. An intraoperative pressure-measuring device used in total knee arthroplasties and its kinematics correlations. *Clin Orthop* 2004;427:171.
27. Matsuda S, White SE, Williams II VG, et al. Contact stress analysis in meniscal bearing total knee arthroplasty. *J Arthroplasty* 1998;13:699.
28. Matsuda S, Williams VG, Whiteside LA, et al. A comparison of pressure sensitive film and digital electronic sensors to measure contact area and contact stress. *Orthop Trans* 1995;19:920.
29. Matsuda S, Whiteside LA, White SE. The effect of varus tilt on contact stresses in total knee arthroplasty: a biomechanical study. *Orthopedics* 1999;22:303.
30. Takeuchi T, Lathi VK, Khan AM, et al. Patellofemoral contact pressures exceed the compressive yield strength of UHMWPE in total knee arthroplasties. *J Arthroplasty* 1995;10:363.
31. Buechel FF, Pappas MJ, Makris G. Evaluation of contact stress in metal-backed patellar replacements. A predictor of survivorship. *Clin Orthop Relat Res* 1991;190.
32. Deng M, Latour RA, Ogale AA, et al. Study of creep behavior of ultra-high-molecular-weight polyethylene systems. *J Biomed Mater Res* 1998;40:214.
33. Murantoglu OK, Perinchieff RS, Bragdon CR, et al. Metrology to quantify wear and creep of polyethylene tibial knee inserts. *Clin Orthop* 2003;410:155.
34. Miura H, Higaki H, Nakanishi Y, et al. Prediction of total knee arthroplasty polyethylene wear using the wear index. *J Arthroplasty* 2002;17:760.



# Dynamic Activity Dependence of In Vivo Normal Knee Kinematics

Taka-aki Moro-oka,<sup>1,2</sup> Satoshi Hamai,<sup>2</sup> Hiromasa Miura,<sup>2</sup> Takeshi Shimoto,<sup>3</sup> Hidehiko Higaki,<sup>3</sup> Benjamin J. Fregly,<sup>1</sup> Yukihide Iwamoto,<sup>2</sup> Scott A. Banks<sup>1,4</sup>

<sup>1</sup>Department of Mechanical and Aerospace Engineering, University of Florida, 318 MAE-A, Mail Stop 116250, Gainesville, Florida 32611

<sup>2</sup>Department of Orthopaedic Surgery, Graduate School of Medical Sciences, Kyushu University, Fukuoka, Japan

<sup>3</sup>Department of Mechanical Engineering, Faculty of Engineering, Kyushu Sangyo University, Fukuoka, Japan

<sup>4</sup>The BioMotion Foundation, Palm Beach, Florida

Received 20 November 2006; accepted 25 January 2007

Published online in Wiley InterScience (www.interscience.wiley.com). DOI 10.1002/jor.20488

**ABSTRACT:** Dynamic knee kinematics were analyzed for normal knees in three activities, including two different types of maximum knee flexion. Continuous X-ray images of kneel, squat, and stair climb motions were taken using a large flat panel detector. CT-derived bone models were used for model registration-based 3D kinematic measurement. Three-dimensional joint kinematics and contact locations were determined using three methods: bone-fixed coordinate systems, interrogation of CT-based bone model surfaces, and interrogation of MR-based articular cartilage model surfaces. The femur exhibited gradual external rotation throughout the flexion range. Tibiofemoral contact exhibited external rotation, with contact locations translating posterior while maintaining 15° to 20° external rotation from 20° to 80° of flexion. From 80° to maximum flexion, contact locations showed a medial pivot pattern. Kinematics based on bone-fixed coordinate systems differed from kinematics based on interrogation of CT and MR surfaces. Knee kinematics varied significantly by activity, especially in deep flexion. No posterior subluxation occurred for either femoral condyle in maximum knee flexion. Normal knees accommodate a range of motions during various activities while maintaining geometric joint congruency. © 2007 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. *J Orthop Res* 26:428–434, 2008

**Keywords:** tibiofemoral joint; kinematics; dynamic activity; flat panel detector; 3D/2D model registration; tibiofemoral contact

## INTRODUCTION

3D-to-2D model registration techniques have been used to measure normal in vivo knee kinematics from radiographic images. These studies have provided clinically insightful information and spurred further complementary investigations. Research topics of current interest include exploration of knee kinematics over the full flexion range and development of techniques for generating and reporting data to provide enhanced physiologic insight.

3D kinematics studies have focused on motion from 0° to 90° of flexion. Asano et al.<sup>1</sup> reported a medial pivot motion pattern during static weight bearing from 0° to 90°. Komistek et al.<sup>2</sup> reported dynamic normal knee kinematics using shape-

matching techniques with computed tomography (CT)-derived bone models. They showed that the lateral condyle experienced significantly more AP translation in activities up to 90° of flexion with no significant differences among activities. More recently, Li et al.<sup>3</sup> reported cartilage contact kinematics from 0° to 90° using bi-plane fluoroscopy and MRI-derived bone models. No study has analyzed dynamic, weight-bearing tibiofemoral motion from extension to maximum knee flexion.

3D knee kinematic studies also have used a variety of coordinate systems and joint contact computations to describe findings, making it sometimes difficult to compare results of one study to another. For example, varus–valgus motion determined from coordinate systems embedded in the bones can be interpreted to indicate separation of the joint surfaces. However, when the different sagittal slopes of the medial and lateral tibial surfaces are considered, varus–valgus motion can result from the condyles moving AP while remaining in contact with these surfaces. A complete

Correspondence to: Scott A. Banks (Telephone: 352-392-6109; Fax: 352-392-7303; E-mail: banks@ufl.edu)

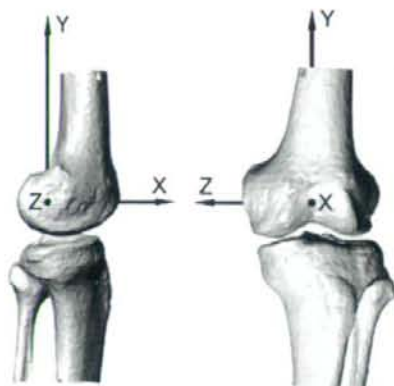
© 2007 Orthopaedic Research Society. Published by Wiley Periodicals, Inc.

mapping of joint surface separations provides a coordinate system-independent measure of articular apposition that avoids misinterpretation. To date, CT-derived bone models (CT model) have been primarily used for model registration-based kinematic measurements.<sup>1,2,4</sup> Determination of tibiofemoral contact conditions from CT models assumes uniform cartilage thickness. MR-based surface models including cartilage are superior for computing surface interactions, but suffer from less accurate bone definition, which can lead to less precise kinematic measurements.<sup>5</sup> The optimal approach, therefore, might be to perform model registration using CT-derived bone models with surface interrogations performed using MR-derived articular surfaces.

The purpose of this study was to analyze healthy knee kinematics in three activities from extension to maximum knee flexion using high-resolution dynamic flat-panel detector images. A secondary purpose was to consider cartilage thickness in tibiofemoral contact analysis and to compare different methods for reporting articular kinematics.

## MATERIALS AND METHODS

Six healthy male subjects, averaging 29 years (28–31), 171 cm (165–177), and 69 kg (55–80) gave informed consent to participate in this study as approved by the institutional review board. Bone models of the femur and tibia/fibula were created from CT (Toshiba, Aquilion, Tochigi, Japan) and MR (Hitachi, Airis II Comfort, Tokyo) scans of one leg. CT scans used a  $512 \times 512$  image matrix, a  $0.35 \times 0.35$  pixel dim, and a 1-mm thickness spanning approximately 150 mm above and below the joint line of the knee, and 2 mm slices through the centers of the hip and ankle. MR scans used a  $512 \times 512$  image matrix, a  $0.39 \times 0.39$  pixel dim, and a 1-mm thickness spanning more than 80 mm above and below the joint line of the knee. Scan time of MR ranged from 11 to 15 min. Exterior cortical bone edges for CT and external edges of both cortical bone and cartilage for MRI were segmented using commercial software (SliceOmatic, Tomovision, Montreal, CA), and these point clouds were converted into polygonal surface models (Geomagic Studio, Raindrop Geomagic, Research Triangle Park, NC). Anatomic coordinate systems were embedded in each bone model following conventions of previous studies.<sup>5</sup> The coordinate systems were first defined for the CT models. The mediolateral (Z) axis of both the femur and tibia/fibula was defined by fitting a cylinder to each posterior condyle of the femur (Fig. 1). The mid point of the cylindrical axis was defined as the coordinate system origin. The proximal/distal (Y) axis for the femur was defined by a line perpendicular to the cylindrical axis in the plane intersecting the femoral head center. The proximal/distal (Y) axis for the shank



**Figure 1.** The mediolateral (Z) axis of both the femur and tibia/fibula was defined by fitting a cylinder to the posterior condyles of the femur. The mid point of the cylindrical axis was defined as the coordinate system origin for both tibia and femur.

was perpendicular to the cylindrical axis in the plane intersecting the ankle center. The anteroposterior (X) axis was formed from the cross product of the first two. Next, the MR model was registered with the corresponding CT model in its initial reference pose in order to align the embedded coordinate systems in each bone model. Proprietary automated alignment software was used to match 3D bone surfaces (Geomagic Studio). Reported accuracy for this technique is less than 0.1 micrometers in length and 0.1 arcseconds in angle compared to a reference standard value.

Continuous sagittal X-ray images of kneel, squat, and stair-climb activities for each subject were taken using a flat panel detector (Hitachi, Clavis, 3 frames/s, image area  $397 \times 298$  mm,  $0.20 \times 0.20$  mm/pixel resolution) (Fig. 2). For the kneel activity, subjects placed their leg on a box, with the foot and ankle hanging freely, and applied their body weight to achieve maximum knee flexion. For the squat activity, the subject stood on a wheeled cart, moved rearward at  $>2.5$  cm/s, so their knee position could be maintained in front of the flat-panel detector as they flexed from full extension to full flexion. For the stair activity, subjects ascended a two-step staircase in a reciprocal manner with knee motions recorded on the first step. A total of 293 images were used: 90 images for kneel, 130 for squat, and 73 for stair.

The 3D position and orientation of the tibia/fibula and femur were determined using previously reported shape matching techniques<sup>4–6</sup> (Fig. 3). A region of the flat-panel X-ray image was extracted and scaled to  $512 \times 512$  square pixels for 3D shape registration. The CT models were projected onto the X-ray image and manually aligned with the bone projections. An automated matching algorithm using nonlinear least squares optimization, and an image edge-to-model edge distance criteria was used to refine registration of the bone models. RMS errors for this method were 0.53 mm for in-plane





**Figure 2.** Subjects performed kneeling (left), lunge (middle) and stair (right) activities while their knee motion was observed using a large flat panel x-ray image detector. The images show a left knee being studied. Kneeling was performed from  $90^\circ$  to maximum comfortable flexion. Squatting was performed from a standing position to maximum comfortable flexion. The stair activity captured the ascent phase from approximately  $70^\circ$  to  $10^\circ$  flexion.

translation, 1.6 mm for out-of-plane translation, and  $0.54^\circ$  for rotations in a previous study.<sup>5</sup> Propagation of these uncertainties to the articular surfaces results in a 95% confidence interval  $>2$  mm for declaring separation of the femoral condyles from the tibial surface.

The kinematics of the joint were determined from the 3D position of each bone model using Cardan angles as described by Tupling and Pierrynowski.<sup>7</sup> Knee rotations were analyzed as a function of flexion angle. Spline interpolation with  $5^\circ$  flexion increments was used to create average kinematics for the group. The stair climb proceeded from flexion to extension; the kneel and squat proceeded from extension to flexion. Each subject began and completed each activity at slightly different flexion angles, so averages are reported for flexion angles from three to six knees. One-way ANOVA and post hoc tests (Bonferroni/Dunn) ( $p < 0.05$ ) were used to examine differences among overall results.

The surface of each CT-derived tibial model was divided into medial and lateral compartments and represented as a cloud of points. For every image, a surface separation map was created by computing the minimum distance between each point on the tibial surface and all points on the femoral surface (Fig. 4). Medial and lateral condylar contact locations were computed as the geometric centroid of the region having  $<6$  mm separation, which acknowledges uncertainty about cartilage thickness,<sup>8,9</sup> cartilage deformation, and measurement errors. The choice of separation threshold had a negligible effect on centroid location for thresholds above 3 mm. The angle between the kinematics of the femur with respect to the tibia.

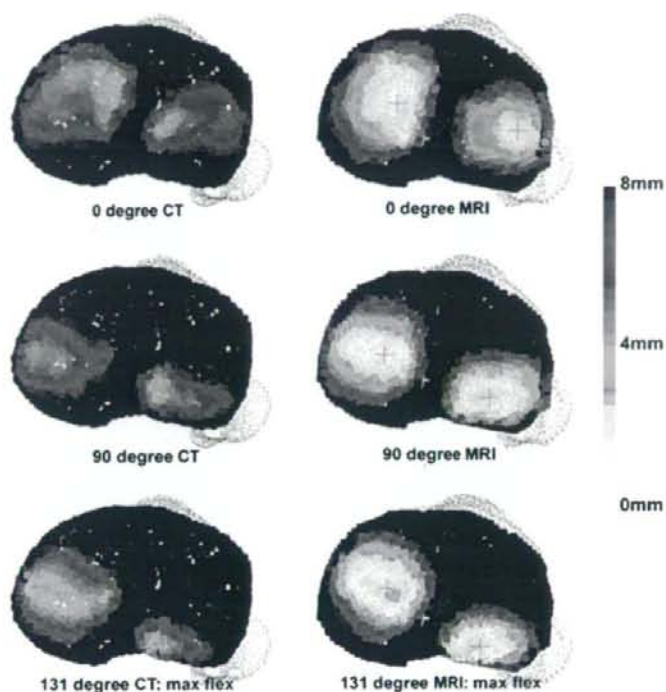
Interactions of cartilage surfaces were determined from the MR-derived models. The bony part of the MR models was registered with the corresponding CT models using automated alignment software (Geomagic Studio). Knee kinematics determined from CT model registration were then used to orient the cartilage surfaces. Contact points were defined in the same manner for the MR models, except a 2-mm maximum separation threshold was used to define the region of contact, contact centroids, and rotational kinematics (Fig. 4).

## RESULTS

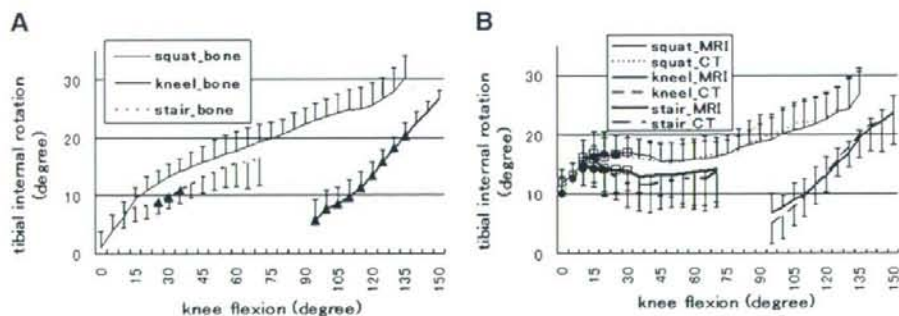
Bony kinematics varied among activities. Femoral rotation during squatting was significantly different from kneeling at all flexion angles and differed from stair climbing from  $25^\circ$  to  $35^\circ$  flexion (Fig. 5A). However, the femur rotated externally with flexion in all three activities. Sharp external rotation was seen from  $0^\circ$  to  $15^\circ$  in squatting,



**Figure 3.** 3D-to-2D CT model-to-flat panel image registration was used to determine the kinematics of healthy knees during three weight bearing activities.



**Figure 4.** Separation distance mapping for squat at 0°, 90°, and maximum flexion (131°) in one subject. Pictures calculated from the MRI model and CT model are shown at the same flexion angle. The centroids of the contact areas are shown as a red cross to indicate the estimated contact location. The picture shows irregular estimated contact regions based on the CT derived models, while the MR derived models of articular cartilage show uniformly rounded contact region estimates



**Figure 5.** Femoral external rotation increased with knee flexion for stair, squat and kneel activities. A: Femoral external rotations based on bone-embedded coordinate systems. Solid triangles on the stair and kneel represent statistically significant differences from the corresponding rotation during the squat activity ( $p < 0.05$ ). B: Femoral external rotations based on contact locations determined from MR and CT derived cartilage and bone models, respectively. There were no statistically significant differences between rotations measured by these two methods. Solid squares represent significant differences between rotations determined by the MR model based contact locations and the rigid body bone motions in graph A ( $p < 0.05$ ). Solid circles represent significant differences between rotations determined by the CT model based contact locations and the rigid body bone motions in graph A ( $p < 0.05$ ).



similar to the "screw home" movement reported for passive knee motions.<sup>10</sup>

The regions used to estimate contact location were irregularly shaped for the CT models and consistently circular or elliptical for the MR models (Fig. 4). Rotations determined from the CT and MR model contact points were not significantly different (Figs. 4 and 5B). There were significant differences between rotations determined from rigid body bone kinematics and contact points for stair and squat activities at flexion angles  $<30^\circ$  (Fig. 5A and B). There were no statistically significant differences between rotations measured by MR derived model and CT derived model.

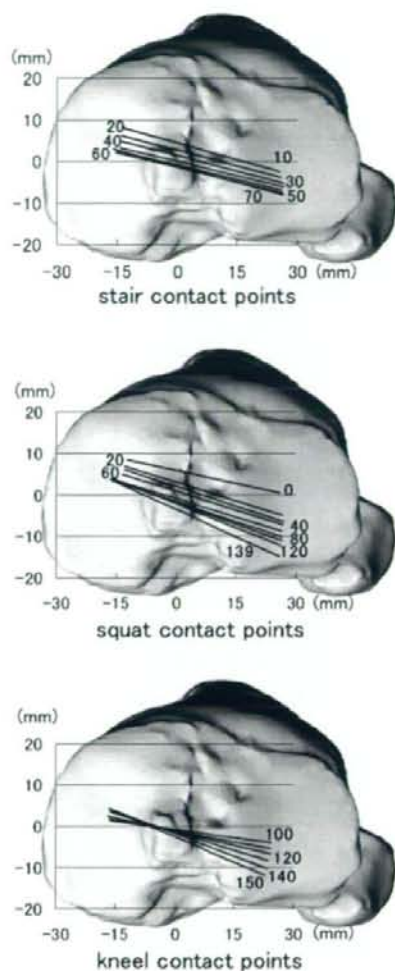
AP translation of the condylar contact regions showed three phases over the range of motion (Fig. 6). From extension to  $20^\circ$  flexion, the contact points rotated externally with posterior translation. From  $20^\circ$  to  $80^\circ$ , they translated posteriorly with little additional rotation. For flexion  $>80^\circ$  (squat and kneel), additional external rotation resulted predominantly from posterior lateral translation, essentially pivoting about the medial compartment. There were significant differences in lateral contact positions between squat and kneel from  $100^\circ$ – $139^\circ$  flexion ( $<0.01$  for  $100^\circ$  to  $135^\circ$ , and  $p=0.015$  at  $139^\circ$ ). There were no differences in contact positions for the medial condyle.

Condylar separation was not observed at flexion  $<150^\circ$ . In two subjects the minimum medial surface separation exceeded 2 mm (MR cartilage model) in terminal kneeling at  $>150^\circ$  flexion.

## DISCUSSION

This study examined dynamic 3D normal knee kinematics in three activities, including two types of maximum knee flexion, using a high-resolution flat panel X-ray detector. Different values of knee axial rotation were observed with different activities, with the greatest femoral external rotation observed during the squat activity. No subluxation of the lateral femoral condyle was observed in maximum knee flexion, and condylar separation from the tibial surface was not commonly observed. Rigid body kinematics measured from bone-embedded coordinate systems produced different joint axial rotations than were measured from the orientation of the medial and lateral condylar contact locations for flexion  $<30^\circ$ .

Knee axial rotations differed according to the method of calculation. Rotations defined from bone-embedded coordinate systems showed gradual femoral external rotation across the entire flexion arc (Fig. 5A). Rotations measured from condylar



**Figure 6.** Average contact locations superimposed on an MR model of the tibial surface. For all activities, the contact points in the lateral compartment moved from anterior to posterior with knee flexion. Contact locations are shown for  $10^\circ$  flexion increments for stair and kneel activities,  $20^\circ$  flexion increments for the squat activity. There were no significant differences between contact locations for the squat and stair activities.

contact locations exhibited three phases: sharp external rotation from  $0^\circ$  to  $15^\circ$  flexion, relatively constant rotation from  $20^\circ$  to  $80^\circ$  flexion, and increasing rotation from  $80^\circ$  to  $150^\circ$  flexion (Fig. 5B). Significant differences in early flexion ( $0^\circ$  to  $30^\circ$ ) are due to femoral geometry. Iwaki et al.<sup>11</sup> showed that the sagittal profile of the medial femoral condyle is composed of a larger distal and smaller posterior circular arc, while the lateral condyle is well described by a single circular facet.

From extension to mid flexion, the medial tibial contact area had a more ovoid shape and moved posteriorly (Figs. 5 and 6). This relative internal rotation of the contact locations was offset by relative external femoral rotation, resulting in less rotation measured at the contact locations than was measured from the bone-embedded coordinate systems over this range. From 20° to 80° flexion the contact locations moved posterior, inclined 15° to 20° with respect to the tibia, maintaining the femur approximately in line with the foot as previously reported.<sup>11</sup> However, unlike previous reports, contact-derived rotation continued from 80° to full flexion in squat and kneel. From 80° to maximum flexion, the contact-derived and rigid body bone rotations showed similar curves, which is reasonable considering the circular geometry of the posterior femoral condyles.<sup>11</sup>

Banks et al.<sup>12</sup> showed kinematics varied with implant design and activity in total knee arthroplasty patients. In this study, kinematics varied with activity in healthy knees, particularly with squat and kneel. The greatest total rotation and the most posterior lateral contact was observed with squatting, but the maximum flexion angle and the rate of tibial rotation from 100° to maximum flexion was greater with kneeling. Medial contact was in the same area for both activities. The lateral femoral condyle translated to the posterior edge of the tibial surface during squat, although there was no subluxation. Subjects were encouraged to achieve comfortable maximum flexion, thus results may have differed if the squat was forced to greater knee flexion.

Comparison of stair climb and squat activities revealed small kinematic differences in the middle range of flexion, which were not statistically significant. The femur showed greater external rotation for all flexion angles during squatting. Results from cadaver studies,<sup>11</sup> MRIs of living knees,<sup>13</sup> and living knees with RSA<sup>14</sup> have shown that a posture with tibial external rotation can suppress tibial internal rotation over the flexion arc. The wider stance width used during the squat activity probably biased the knee toward greater femoral external rotation in the mid range of flexion. Standard deviations for the squat and stair activities were 20% greater than for the kneel activity, suggesting kneeling is more geometrically constrained or stereotypic.

Surface distance calculations revealed only two cases showing evidence of condylar separation, both occurring at terminal flexion in kneeling beyond 150°. In the static MRI study of maximally flexed knees, Nakagawa et al.<sup>15</sup> reported the

medial condyle remained positioned over the tibial articular surface but lifted away from it, mechanically similar to the present finding. Two differences may account for infrequent observation of condylar separation in this study. First, a conservative threshold (95% confidence interval) was used for declaring separation, acknowledging uncertainty accrued from measurement uncertainty, surface modeling errors, and possible cartilage deformation. Second, Nakagawa et al.<sup>15</sup> studied a different activity. In this study, subjects knelt to maximum passive flexion with partial body weight and an upright trunk posture (hip angle approximately 90°). The subjects reported in Nakagawa et al.<sup>15</sup> were supine, with the hip extended in the MRI scanner, similar to an athlete's quadriceps stretching posture. Hip extension might have caused greater quadriceps tension and resulted in a more exaggerated joint posture.

Nakagawa et al.<sup>15</sup> also concluded that the lateral femoral condyle subluxed posteriorly from the tibia at full passive flexion, a phenomenon not seen in the present study. Nakagawa et al.<sup>15</sup> tracked a cylindrical axis embedded in the posterior condyles and determined that subluxation occurred when the lateral axis moved posterior to the tibial surface at 162° flexion. However, subluxation is an articular phenomenon where the surfaces of a joint no longer face each other exactly but remain partially aligned. MR and cryosection pictures in the report of Nakagawa et al.<sup>15</sup> show the lateral femoral condyle has adequate contact with the edge of the tibial articular surface even when the cylindrical axis has exceeded the tibial periphery. These pictures are consistent with the observations of this study.

Findings of articular separation and subluxation in healthy joints sometimes can result from methodological issues and sometimes from articular reality. In the latter case, it these phenomena might be better appreciated as the boundaries of the passive characteristics of the normal healthy joint than negative attributes. For example, the healthy knee has been observed to have articular gaps in 90° of flexion in the medial compartment and in the lateral compartment depending on the posture.<sup>16</sup> The knee joint has long been known to possess an envelope of possible motions that only now are being explored by advanced measurement techniques.

The combination of dynamic imaging with flat-panel detectors and CT/MR bone models provide some novel measurement opportunities. However, refinements and enhancements to the method will expand the range of possible studies. First,



the flat-panel detector used for this study provided high spatial resolution but only three images/s. Solid-state detectors now commercially available with 30 images/s capabilities will permit study of highly dynamic activities. Second, single-plane imaging provides tremendous experimental flexibility and relatively large viewing volumes, but suffers from the main limitation of monocular vision—poor sensitivity for motions perpendicular to the image plane, which was approximately 2 mm in our preliminary study using a lower resolution image source.<sup>5</sup> Bi-plane imaging techniques can provide measurements with enhanced precision and more uniform uncertainties when required, but restrict the viewing volume, increase X-ray exposure and at least double the cost. Third, geometric and coordinate alignment errors in the CT and MR models affect the measurement accuracy, and it is well known that MR images/models suffer from geometric distortion.<sup>5</sup> For this reason, CT-derived models were used for 3D-to-2D model-to-image registration, only using the MR-derived models for articular surface analysis once the 3D pose of the bones was determined. In addition, the MR scans were performed nonweight bearing, so the compliant cartilage surfaces presumably were unloaded. Additional measurement uncertainty results when interrogating these cartilage surfaces based on weight-bearing conditions,<sup>17</sup> and the contact locations must be defined conservatively. Fourth, the current approach to analyzing joint contact ignores the menisci, which are invisible on X-ray but obviously affect joint contact and load distribution. There is not currently an X-ray-based technique that will overcome this limitation, but dynamic MR studies can provide this capability.<sup>18</sup> Finally, simple experimental issues limited the range of the current observations. Subjects were instructed to perform the three activities in a natural and comfortable manner, and some subjects extended and/or flexed their knees more than others. The data reported here are limited to the common flexion ranges for the subjects. In summary, the study provides observations of healthy knee kinematics in three different activities over the flexion range from extension to 150° using dynamic imaging with CT/MR-derived models. Future experimental and technical improvements might further expand the range and quality of measurements that can be obtained.

## REFERENCES

- Asano T, Akagi M, Tanaka K, et al. 2001. In vivo three-dimensional knee kinematics using a biplanar image-matching technique. *Clin Orthop Relat Res* 388:157–166.
- Komistek RD, Dennis DA, Mahfouz M. 2003. In vivo fluoroscopic analysis of the normal human knee. *Clin Orthop Relat Res* 410:69–81.
- Li G, DeFrate LE, Park SE, et al. 2005. In vivo articular cartilage contact kinematics of the knee: an investigation using dual-orthogonal fluoroscopy and magnetic resonance image based computer models. *Am J Sports Med* 33:102–107.
- Fregly BJ, Rahman HA, Banks SA. 2005. Theoretical accuracy of model-based shape matching for measuring natural knee kinematics with single-plane fluoroscopy. *J Biomech Eng* 127:692–699.
- Moro-oka T, Hamai S, Miura H, et al. 2007. Can MR derived bone models be used for accurate motion measurement with single-plane 3D shape registration? *J Orthop Res* 7:867–872.
- Banks SA, Hodge WA. 1996. Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy. *IEEE Trans Biomed Eng* 43:638–649.
- Tupling SJ, Pierrynowski MR. 1987. Use of cardan angles to locate rigid bodies in three-dimensional space. *Med Biol Eng Comput* 25:527–532.
- Adam C, Eckstein F, Milz S, et al. 1998. The distribution of cartilage thickness in the knee-joints of old-aged individuals—measurement by A-mode ultrasound. *Clin Biomech (Bristol, Avon)* 13:1–10.
- Koo S, Gold GE, Andriacchi TP. 2005. Considerations in measuring cartilage thickness using MRI: factors influencing reproducibility and accuracy. *Osteoarthritis Cartilage* 13:782–789.
- Moglo KE, Shirazi-Adl A. 2005. Cruciate coupling and screw-home mechanism in passive knee joint during extension–flexion. *J Biomech* 38:1075–1083.
- Iwaki H, Pinskerova V, Freeman MA. 2000. Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee. *J Bone Joint Surg Br* 82:1189–1195.
- Banks SA, Hodge WA. 2004. 2003 Hap Paul Award Paper of the International Society for Technology in Arthroplasty. Design and activity dependence of kinematics in fixed and mobile-bearing knee arthroplasties. *J Arthroplasty* 19:809–816.
- Hill PF, Vedi V, Williams A, Iwaki H, et al. 2000. Tibiofemoral movement 2: the loaded and unloaded living knee studied by MRI. *J Bone Joint Surg Br* 82:1196–1198.
- Karrholm J, Brandsson S, Freeman MA. 2000. Tibiofemoral movement 4: changes of axial tibial rotation caused by forced rotation at the weight-bearing knee studied by RSA. *J Bone Joint Surg Br* 82:1201–1203.
- Nakagawa S, Kadoya Y, Todo S, et al. 2000. Tibiofemoral movement 3: full flexion in the living knee studied by MRI. *J Bone Joint Surg Br* 82:1199–1200.
- Tokuhara Y, Kadoya Y, Nakagawa S, et al. 2004. The flexion gap in normal knees. An MRI study. *J Bone Joint Surg Br* 86:1133–1136.
- Forster H, Fisher J. 1996. The influence of loading time and lubricant on the friction of articular cartilage. *Proc Inst Mech Eng [H]* 210:109–119.
- Sheehan FT, Zajac FE, Drape JE. 1999. In vivo tracking of the human patella using cine phase contrast magnetic resonance imaging. *J Biomech Eng* 121:650–656.

# Kinematic Analysis of Kneeling in Cruciate-Retaining and Posterior-Stabilized Total Knee Arthroplasties

Satoshi Hamai,<sup>1</sup> Hiromasa Miura,<sup>1</sup> Hidehiko Higaki,<sup>2</sup> Shuichi Matsuda,<sup>1</sup> Takeshi Shimoto,<sup>2</sup> Kousuke Sasaki,<sup>1</sup> Masaaki Yoshizumi,<sup>2</sup> Ken Okazaki,<sup>1</sup> Nobuaki Tsukamoto,<sup>1</sup> Yukihide Iwamoto<sup>1</sup>

<sup>1</sup>Department of Orthopaedic Surgery, Graduate School of Medical Sciences, Kyushu University, 3-1-1 Maidashi, Higashi-ku, Fukuoka 812-8582, Japan

<sup>2</sup>Department of Mechanical Engineering, Faculty of Engineering, Kyushu Sangyo University, 2-3-1 Matsugadai, Higashi-ku, Fukuoka 813-8583, Japan

Received 17 August 2006; accepted 26 July 2007

Published online in Wiley InterScience (www.interscience.wiley.com). DOI 10.1002/jor.20512

**ABSTRACT:** Kneeling is an important function of the knee for many activities of daily living. In this study, we evaluated the in vivo kinematics of kneeling after total knee arthroplasty (TKA) using radiographic based image-matching techniques. Kneeling from 90 to 120° of knee flexion produced a posterior femoral rollback after both cruciate-retaining and posterior-stabilized TKA. It could be assumed that the posterior cruciate ligament and the post-cam mechanism were functioning. The posterior-stabilized TKA design had contact regions located far posterior on the tibial insert in comparison to the cruciate-retaining TKA. Specifically, the lateral femoral condyle in posterior-stabilized TKA translated to the posterior edge of the tibial surface, although there was no finding of subluxation. After posterior-stabilized TKA, the contact position of the post-cam translated to the posterior medial corner of the post with external rotation of the femoral component. Because edge loading can induce accelerated polyethylene wear, the configuration of the post-cam mechanism should be designed to provide a larger contact area when the femoral component rotates. © 2007 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. *J Orthop Res* 26:435–442, 2008

**Keywords:** kinematics; kneeling; posterior cruciate ligament; post-cam mechanism; total knee arthroplasty

## INTRODUCTION

Kneeling is an important function for many activities of daily living. More than 50% of all patients undergoing total knee arthroplasty (TKA) consider kneeling important and participate in kneeling.<sup>1</sup> Inability to kneel after knee surgery is a frequent cause of dissatisfaction. Kneeling for different functions requires different degrees of knee flexion. Kneeling in 90° of knee flexion has been called "upright kneeling" and kneeling in more than 110° of knee flexion "flexed kneeling."<sup>2</sup> Flexed kneeling is important especially in Middle and Far Eastern patients as they sit on and rise from the floor. The patella and the tibial tuberosity contact the ground in upright kneeling, while only the tibial tuberosity bears weight in flexed kneeling.<sup>3</sup>

Kneeling ability depends on many factors. Scar position and skin hypoesthesia can limit kneeling ability. A lateral incision produces less dysesthesia,

fewer neuromas, and less discomfort.<sup>4</sup> A patient's ability to kneel may be greater than their own perception of their ability to do so.<sup>2,5</sup> Many patients undergoing TKA avoid kneeling for fear of harming the prosthesis.

Weight-bearing load may be exerted on the patellofemoral articulation and the retained posterior cruciate ligament (PCL) after cruciate-retaining (CR) TKA or the post-cam mechanism after posterior-stabilized (PS) TKA during activities requiring deep flexion.<sup>6–8</sup> However, very little is known regarding knee kinematics during upright and flexed kneeling, and even less is known about how TKA design affects knee biomechanics at this limit of knee flexion. The purpose of this study was to clarify in vivo knee kinematics during kneeling. In addition, we compared kinematics with the joint in a hyperflexed position between CR TKA and PS TKA.

The in vivo 3D position and orientation of the components of CR and PS TKA during both upright and flexed kneeling were evaluated using an image-matching method. In a previous study, Komistek et al.<sup>9</sup> analyzed femorotibial contact positions in static upright kneeling using an iterative computer model-fitting technique. However,

Correspondence to: Hiromasa Miura (Telephone: +81-92-642-5488; Fax: +81-92-642-5507; E-mail: miura@ortho.med.kyushu-u.ac.jp)

© 2007 Orthopaedic Research Society. Published by Wiley Periodicals, Inc.



little information is available on the in vivo dynamic kinematics of kneeling from 90° to 120° of flexion after TKA.<sup>10</sup> In addition, this study analyzed subjects having a PS TKA to determine the contact position at the post-cam mechanism. Nakayama et al.<sup>11</sup> measured in vitro contact areas and stresses at the post-cam mechanism using a digital electric stress sensor. However, in vivo post-cam contact positions after PS TKA have not been reported.

We hypothesized that knee flexion angles and type of TKA design would affect the tibiofemoral contact positions during kneeling, and that the post-cam contact position after PS TKA would not be in the center of the post at flexed positions.

## MATERIALS AND METHODS

Twenty subjects who had undergone clinically successful TKA and were willing to participate enrolled in the study. Inclusion criteria were the ability to perform both upright and flexed kneeling. All subjects gave informed consent, and an institutional review board approved the study. Ten knees in eight subjects received CR TKA (Foundation knee, Encore Medical Co., Austin, TX) and 10 knees in seven subjects received PS TKA (Nexgen LPS, Zimmer, Inc., Warsaw, IN). The articulating surface geometry of the polyethylene tibial insert is conforming in PS TKA, but relatively flat in CR TKA, allowing femoral rollback produced by tensioning of the PCL during flexion.

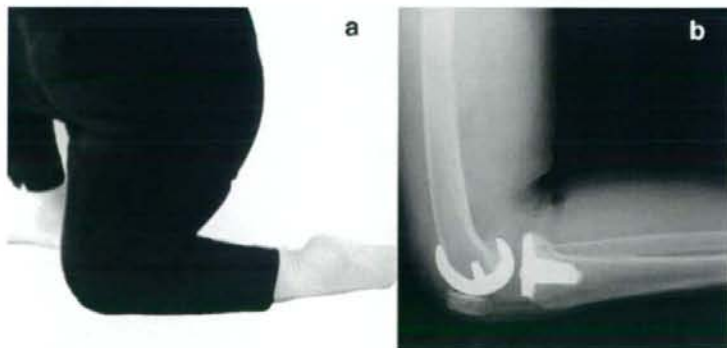
The 20 subjects included 18 women and 2 men with an average follow-up of 17.3 months. The preoperative diagnoses were osteoarthritis in 15 knees and rheumatoid arthritis in 5, and the average age at surgery was 71.6 years. Average postoperative knee extension/flexion angle was -2.8/128.4°. The Knee Society score (knee score/function score) was 91.4/80.6. Three experienced surgeons (H.M., S.M., and K.O.) performed all TKAs using a similar standard technique. The components were aligned so that the mechanical axis passed through

the center of the knee prosthesis. Rotational alignment was adjusted to bony landmarks, such as the trans-epicondylar axis and tibial tuberosity. The PCL was retained when it appeared normal in tension and appearance at surgery. Soft tissues were balanced to achieve varus and valgus stability.

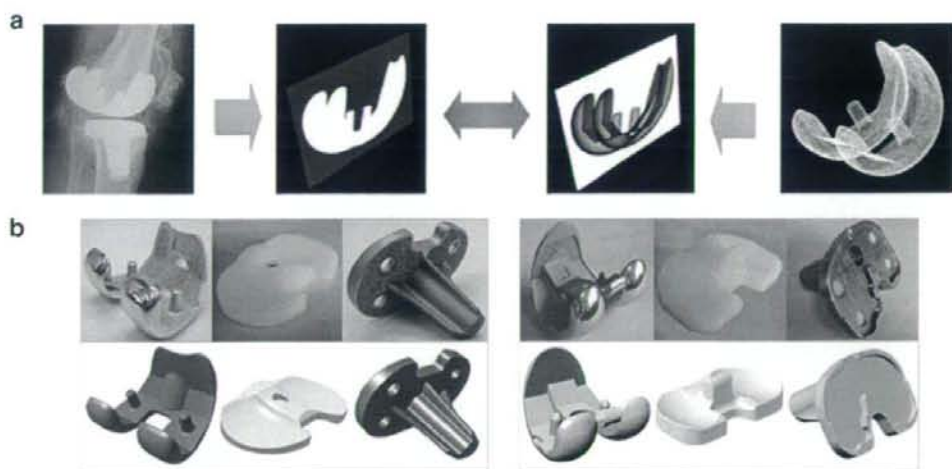
Each subject was asked to perform upright and flexed kneeling in maximum knee flexion under radiological surveillance (Fig. 1). Knee flexion in each subject reached more than 120° of knee flexion passively. During kneeling, subjects placed their leg on a box with the ankle joint extended. Continuous X-ray images were taken using a flat panel detector (Clavis, Hitachi, Co., Ltd., Tokyo). The detector provided higher resolution dynamic images (Distortion-free image, 0.20 × 0.20 mm/pixel resolution, three frames/s, with an image area size 397 (H) × 298 (V) mm). The high contrast resolution of X-ray images provided the basis for the required accuracy.<sup>12,13</sup> Images of kneeling at 90, 100, 110, and 120° of flexion were selected from the continuous images and analyzed using a previously published image-matching technique.<sup>14,15</sup>

The projected 3D computer-aided design (CAD) models of the femoral and tibial components were superimposed on the 2D radiographic images (Fig. 2a). The 3D CAD models of the components and the tibial insert were reconstructed after 3D shape measurements of the total knee prostheses by a profilometer (Mitsutoyo, Co., Ltd., Kawasaki, Japan) (Fig. 2b). The position of the radiation source relative to the X-ray detector was determined from the projection image of a calibration cage made from an acrylic resin board and metallic balls. Projecting all surface points of the 3D model produced the component silhouette. A model silhouette was matched with the actual silhouette by translating and rotating the 3D CAD model to minimize the number of unmatched pixels between the silhouettes. After the 3D pose of the components was estimated, the 6 degrees of freedom of the femoral component relative to the tibial component were determined by transforming the coordinate systems into one common system.

The accuracy and precision of the measurement technique was determined by in vitro investigations.<sup>15,16</sup>



**Figure 1.** Each subject was asked to perform kneeling from 90° to maximum knee flexion (a). Continuous X-ray images of kneeling were taken using a flat panel detector (b).



**Figure 2.** A model silhouette was matched with the actual object silhouette by translating and rotating the 3D CAD models (a). The models of the components (lower stand) were reconstructed at CR TKA (left) and PS TKA (right) after 3D shape measurements of the total knee prostheses (upper stand) by a profilometer (b).

The components mounted on a stage were rotated and translated to known values. The image-matching process was performed for the radiographic images at each position to determine the relative pose of the components. The root-mean-square (RMS) accuracy errors for this process for the femoral component were 0.29 mm for in-plane translation, 0.37 mm for out-of-plane translation, and  $0.27^\circ$  for rotation, and the accuracies for the tibial component were 0.23 mm for in-plane translation, 0.30 mm for out-of-plane translation, and  $0.25^\circ$  for rotation. The maximum errors associated with tracking the position of the tibia relative to the femur were the sum of the RMS errors, which were 0.52, 0.67 mm, and  $0.52^\circ$  respectively, for in-plane and out-of-plane translations and rotation.

The contact positions between the femoral component and tibial insert and tibiofemoral axial rotation were evaluated after both CR and PS TKA (Fig. 3). The contact positions were evaluated by measuring the centroid of contact area between the surfaces of the femoral condyles and the insert using a CAD software program (SolidWorks® 2001Plus SP3.0, SolidWorks Corporation, Concord, MA) (Fig. 3a). If surface separation was recognized, the location of closest proximity between the surfaces of the femoral condyles and the insert was measured in 3D space. The positive or negative value of AP tibiofemoral contact position was defined as anterior or posterior to the midline of the polyethylene insert. The tibiofemoral rotation was evaluated to measure the rotation angle of the femoral component relative to the tibial component (Fig. 3b). The positive or negative value of tibiofemoral rotation was defined as internal or external rotation of the femur relative to tibia. After PS TKA, the medial-lateral and upper-lower translation of contact position

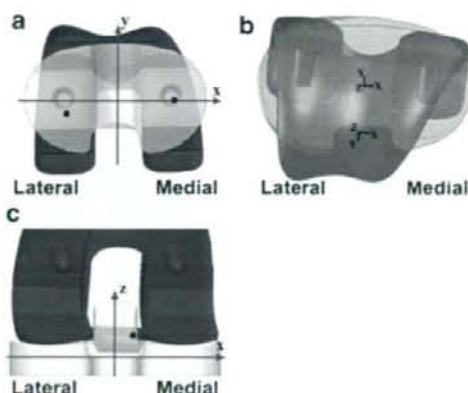
between the femoral cam and tibial post was evaluated (Fig. 3c). The positive or negative value of ML post-cam contact position was defined as medial or lateral to the center of the post. The positive or negative value of proximal-distal post-cam contact position was defined as proximal or distal to the most inferior point of the articular surface.

Statistical analyses were done using Stat View 5.0 (Abacus Concepts, Inc., Berkeley, CA). All values were expressed as the mean  $\pm$  standard deviation. The unpaired *t*-test was used to analyze postoperative knee extension/flexion angles and the Knee Society scores (knee score/function score) comparing CR with PS TKA. Two-way, repeated measures ANOVA were used to analyze the tibiofemoral contact positions and tibiofemoral rotations when investigating the effects of knee flexion angle and TKA design and a comparison of the interaction between these two values. Post hoc tests (Bonferroni/Dunn) were used to identify the source of any differences for relevant comparisons. The paired *t*-test was used to analyze the ML post-cam contact position when comparing  $90^\circ$  with  $120^\circ$  of flexion. A value of  $p < 0.05$  was considered significant.

## RESULTS

The average postoperative extension/flexion angle was  $-3.8 \pm 2.5^\circ/129.3 \pm 5.4^\circ$  after CR TKA and  $-2.0 \pm 3.3^\circ/127.7 \pm 8.6^\circ$  after PS TKA. No significant differences were found in postoperative knee extension ( $p = 0.24$ ) and flexion ( $p = 0.66$ ) between CR TKA and PS TKA. The Knee Society score (knee score/function score) was  $90 \pm 6.3/78 \pm 11.6$





**Figure 3.** (a) Locations of closest proximity or contact between the femoral component and the tibial insert on the medial and lateral sides. The positive or negative value of the AP tibiofemoral contact position was defined as anterior or posterior to the midline of the tibial insert. (b) The rotation angle of the femoral component relative to the tibial component. The positive or negative value of tibiofemoral rotation was defined as internal or external rotation of the femur relative to tibia. (c) Location of contact between the femoral cam and tibial post after PS TKA. The positive or negative value of medial-lateral post-cam contact position was defined as medial or lateral to the center of the post. The positive or negative value of proximal-distal post-cam contact was defined as proximal or distal to the most inferior point of the articular surface.

for CR and  $93 \pm 5.7/82 \pm 13.8$  for PS TKA. No significant differences (knee score:  $p = 0.27$ ; function score:  $p = 0.59$ ) were found between CR TKA and PS TKA.

The average AP tibiofemoral contact positions in  $90^\circ$  of flexion after PS TKA were significantly ( $p < 0.0001$ ) more posterior than after CR TKA with both condyles (Table 1 and Fig. 4). Kneeling from  $90$  to  $120^\circ$  of flexion produced a posterior femoral rollback for both designs. Anterior femoral translation was not found for any subject from  $90$  to  $120^\circ$  of flexion. A significant interaction was found between flexion position and TKA design (ANOVA,  $p < 0.01$ ), indicating that the pattern of changes during kneeling differed between TKA designs. The total rollback after PS TKA (medial =  $-5.2 \pm 2.4$  mm, range  $-1.1$  to  $-8.5$  mm; lateral =  $-4.8 \pm 3.2$  mm, range  $-0.3$  to  $-10.1$  mm) was more than after CR TKA (medial =  $-4.3 \pm 3.7$  mm, range  $-0.1$  to  $-11.4$  mm; lateral =  $-2.4 \pm 1.4$  mm, range  $-0.3$  to  $-4.4$  mm) with both condyles. The average AP tibiofemoral contact positions in  $120^\circ$  of flexion after PS TKA were significantly ( $p < 0.0001$ ) more posterior than after CR TKA with both condyles (Table 1).

Changes in the rotation angle from  $90$  to  $120^\circ$  of flexion were  $0.4 \pm 2.5^\circ$  (range  $-2.8$  to  $4.4^\circ$ ) after CR TKA and  $-1.3 \pm 0.8^\circ$  (range  $-2.3$  to  $-0.1^\circ$ ) after PS TKA. After PS TKA, the change in the external rotation was consistently recognized in each  $10^\circ$  of flexion. The rotation angle in  $120^\circ$  of flexion was  $-7.5 \pm 5.4^\circ$  (range  $-15.7$  to  $-0.5^\circ$ ) after CR TKA and  $-6.9 \pm 4.1^\circ$  (range  $-12.9$  to  $-2.1^\circ$ ) at PS TKA. The femoral components for both designs continued to show external rotation from  $90$  to  $120^\circ$  of flexion. More than  $10^\circ$  of the external rotation was recognized during kneeling for both designs (CR, 4; PS, 3).

The average ML post-cam contact position in  $90^\circ$  of flexion was  $4.0 \pm 2.2$  mm (range  $0.1$  to  $6.1$  mm) medial from the center of the post with external rotation of the femoral component (Table 1 and Fig. 4c).  $1.2 \pm 1.2$  mm (range  $0.1$  to  $3.4$  mm) medial translation of the post-cam contact position occurred from  $90$  to  $120^\circ$  of flexion because the femoral component rotated externally. The average ML post-cam contact position in  $120^\circ$  of flexion was significantly more medial than in  $90^\circ$  ( $p < 0.05$ ). The average proximal-distal post-cam contact position in  $90^\circ$  of flexion was proximal from the most inferior point of the articular surface (Table 1 and Fig. 4c).  $1.4 \pm 0.9$  mm (range  $0.0$  to  $2.4$  mm) distal translation of the post-cam contact position occurred from  $90$  to  $120^\circ$  of flexion. No proximal translation was observed from  $90$  to  $120^\circ$  of knee flexion.

## DISCUSSION

Kneeling in more than  $110^\circ$  of knee flexion tends to produce posterior translation of the tibia relative to the femur because the tibial tuberosity bears weight.<sup>2</sup> The ground reaction force on the tibial tuberosity causes a posterior directed shear force on the tibia during kneeling. However, posterior translation is prevented by the retained PCL after CR TKA and the post-cam mechanism after PS TKA. Without PCL retention or the post-cam mechanism, the femoral component would not rollback posteriorly, instead translating anteriorly. If the retained PCL and the post-cam mechanism did not provide posterior rollback, an excessive patellofemoral contact force may occur. In this study, kneeling from  $90$  to  $120^\circ$  of flexion produced posterior rollback in both designs (Fig. 4a and b). Previous studies reported that the retained PCL after CR TKA reproduces posterior rollback.<sup>17-20</sup> In addition, after PS TKA, the post-cam mechanism was designed to function over  $75^\circ$  of knee flexion, so we can assume

**Table 1.** AP Tibiofemoral Contact Positions [+ Anterior, - Posterior] and Tibiofemoral Rotation [+ IR, - ER] after Both CR and PS TKA, and the ML [+ Medial, - Lateral] and Proximal-Distal [+ Proximal from the Most Inferior Point of the Articular Surface] Post-cam Contact Position after PS TKA while Kneeling from 90° to 120° of Knee Flexion

| AP Tibiofemoral Contact Positions (mm) |   |              |
|--|---|--------------|
| Cruciate-retaining TKA                 | Medial Condyle/Lateral Condyle                    | Rotation (°) |
| 90°                                    | 4.2 ± 3.6 <sup>a</sup> /-4.8 ± 3.3 <sup>a</sup>   | -7.8 ± 4.8   |
| 100°                                   | 2.6 ± 2.5 <sup>a</sup> /-5.4 ± 3.4 <sup>a</sup>   | -7.6 ± 4.7   |
| 110°                                   | 1.1 ± 2.8 <sup>a</sup> /-6.1 ± 3.4 <sup>a</sup>   | -7.5 ± 5.3   |
| 120°                                   | -0.1 ± 3.3 <sup>a</sup> /-7.2 ± 3.9 <sup>a</sup>  | -7.5 ± 5.4   |
| ML Tibiofemoral Contact Positions (mm) |   |              |
| Posterior-stabilized TKA               | Medial Condyle/Lateral Condyle                    | Rotation (°) |
| 90°                                    | -3.7 ± 3.2 <sup>a</sup> /-10.5 ± 3.3 <sup>a</sup> | -5.6 ± 4.2   |
| 100°                                   | -5.8 ± 4.1 <sup>a</sup> /-12.2 ± 1.9 <sup>a</sup> | -5.9 ± 4.2   |
| 110°                                   | -7.6 ± 4.1 <sup>a</sup> /-14.3 ± 1.2 <sup>a</sup> | -6.4 ± 4.1   |
| 120°                                   | -8.9 ± 4.6 <sup>a</sup> /-15.3 ± 2.1 <sup>a</sup> | -6.9 ± 4.1   |
| Post-cam Contact Position (mm)         |   |              |
| Posterior-stabilized TKA               | Medial-lateral/Proximal-distal                    |              |
| 90°                                    | 4.0 ± 2.2/7.6 ± 0.7                               |              |
| 100°                                   | 4.3 ± 2.3/6.9 ± 0.5                               |              |
| 110°                                   | 4.7 ± 1.5/6.3 ± 0.7                               |              |
| 120°                                   | 5.1 ± 1.2/6.2 ± 0.8                               |              |

Values are mean ± standard deviation.

<sup>a</sup>Significantly different between CR TKA and PS TKA ( $p < 0.05$ ).

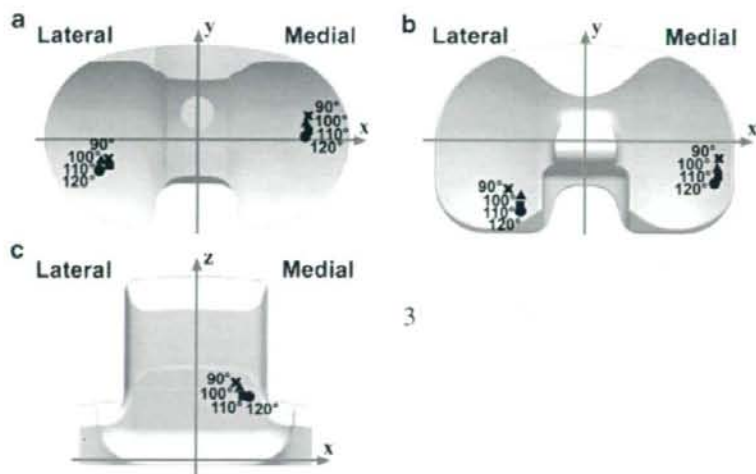
that both the retained PCL and the post-cam mechanism were functioning.

Posterior femoral rollback was preferable to decrease patellofemoral contact force during kneeling. Increased femoral rollback in flexion correlates with reduced patellofemoral contact load in TKA.<sup>21</sup> This reduction in contact force is attributed to the increased extensor moment arm. Browne et al.<sup>22</sup> reported that the design with the longer extensor moment arm required less quadriceps force and reduced patellofemoral compressive force. In PS TKA, contact positions at 90° of flexion were more posterior, and rollback distance with both condyles was longer compared to CR TKA. PS TKA may be preferable to CR TKA to reduce patellofemoral forces during kneeling. Komistek et al.<sup>9</sup> found similar results using in vivo fluoroscopy to evaluate femorotibial contact positions in upright kneeling, reporting that the PCL has less resistance to the

anterior thrust of the femur relative to the tibia in PS TKA.

The PS design had contact regions located more posterior on the tibial insert in comparison to the CR design. Specifically, the lateral femoral condyle in the PS translated to the posterior edge of the tibial surface during kneeling, although subluxation did not occur. This finding has important implications regarding the location and severity of the polyethylene wear. The posteromedial corner of the lateral compartment of the tibial insert suffered edge loading by combining posterior translation and external rotation of the femoral component during kneeling. Wasielewski et al.<sup>23</sup> and Harman et al.<sup>24</sup> showed wear patterns on retrieved polyethylene insert consistent with these articular contact patterns. The contact locations between the femoral component and polyethylene insert during dynamic activities are thus considered to





**Figure 4.** The average tibiofemoral contact positions after CR TKA (a) and PS TKA (b) demonstrated posterior roll back during kneeling from 90 to 120° of knee flexion. The average post-cam contact position after PS TKA translated medially and downward during kneeling (c).

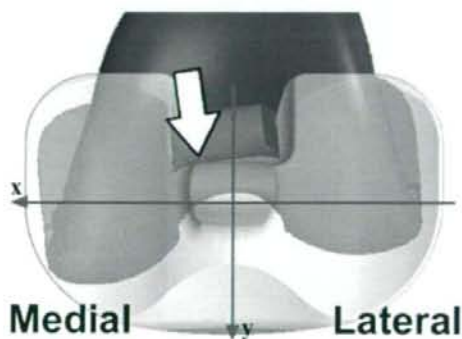
correspond to the wear pattern on the polyethylene articular surface.

No *in vivo* studies have yet focused on the post-cam contact positions in PS TKA. The post-cam contact position during kneeling translated medially and distally along the post (Fig. 4c). The effect of femoral rotation is important for the ML post-cam contact position. External rotation of the femoral component from 90 to 120° of flexion created a medial translation of the post-cam contact position (Fig. 5). Furthermore, 3 of 10 knees after PS TKA had more than 10° of femoral external rotation during kneeling with the risk that the posterior medial corner of the flat-surfaced post could suffer excessive stress because of edge loading. Nakayama et al.<sup>11</sup> reported that when the tibial component was rotated 10°, post-cam contact stress increased more than twofold in comparison to neutral rotation. Edge loading may cause chronic wear and fracture of the post, affecting the long-term prognosis following TKA. The post-cam design should be modified to provide a larger contact area and prevent increased contact stress when the femoral component rotates.

Distal translation of the post-cam contact position was found during kneeling. Condylar liftoff is an important issue for proximal–distal post-cam contact position. However, the proximal–distal post-cam contact position demonstrated no proximal translation. The distal translation is more beneficial in preventing posterior dislocation of the tibial component and in decreasing the risk of

fracture of the post. If flexion instability exists after PS TKA, posterior dislocation of the tibial component may occur.<sup>25–28</sup> However, none of the subjects in this study had evidence of ligamentous laxity or pain on physical examination.

We are concerned about the effect of the load that is exerted on the post-cam mechanism during kneeling. Fracture or polyethylene wear of the post and dislocation of the knee have been reported after PS TKA.<sup>25–33</sup> In addition, the post-cam mechanism could transmit anteroposterior shear and tensile stresses to the bone–implant interfaces and cause component loosening. After CR TKA, the PCL would play a role in dissipating shear and tensile stresses on the bone–implant interfaces.<sup>34</sup> However, further



**Figure 5.** The contact position of the post-cam translated to the posterior medial corner of the post with the external rotation of the femoral component (arrow).

studies are needed to examine changes in PCL laxity due to degeneration or attenuation with time. Flexion instability due to late PCL insufficiency could be a cause of persistent pain and functional impairment after CR TKA leading to revision arthroplasty.<sup>35-37</sup> Frequent kneeling may cause a dysfunction of the retained PCL after CR TKA and the post-cam mechanism after PS TKA over time because the retained PCL and the post-cam mechanism receive significant load.

Our study has limitations. Kinematic information concerning deep knee flexion could not be included in the analysis because most of the subjects could not kneel beyond 130°. Both CR and PS designs used in this study are not intended to flex beyond 130°. More research is needed to assess the dynamic in vivo function of the PCL and the post-cam mechanism in deep flexion.

In conclusion, our results suggest that the PCL after CR TKA and the post-cam mechanism after PS TKA function during kneeling from 90 to 120° of knee flexion. Based on our results, PS TKA may be preferable to CR TKA to reduce forces across the patellofemoral articulation. The PS design has contact regions located far posterior on the tibial insert in comparison to the CR TKA. Specifically, the lateral femoral condyle in PS TKA translated to the posterior edge of the tibial surface. Furthermore, this study suggests the hypothesis that increased edge loading may be associated with an abnormally high degree of stress in the polyethylene, and that this may be associated with an increased risk of failure. Only a clinical study with long term follow-up can determine if this relationship exists, but further attention must be given to the configuration of the post-cam when designing posterior-stabilized total knee prostheses.

## ACKNOWLEDGMENTS

No benefits in any form were received from a commercial party related directly or indirectly to the subject of this article.

## REFERENCES

- Weiss JM, Noble PC, Conditt MA, et al. 2002. What functional activities are important to patients with knee replacements? *Clin Orthop* 404:172-188.
- Palmer SH, Servant CT, Maguire J, et al. 2002. Ability to kneel after total knee replacement. *J Bone Joint Surg Br* 84:220-222.
- Hassaballa M, Vale T, Weeg N, et al. 2002. Kneeling requirements and arthroplasty surgery. *Knee* 9:317-319.
- Berg P, Mjoberg B. 1991. A lateral skin incision reduces peripatellar dysaesthesia after knee surgery. *J Bone Joint Surg Br* 73:374-376.
- Schai PA, Gibbon AJ, Scott RD. 1999. Kneeling ability after total knee arthroplasty. Perception and reality. *Clin Orthop* 367:195-200.
- Dahlkvist NJ, Mayo P, Seedhom BB. 1982. Forces during squatting and rising from a deep squat. *Eng Med* 11:69-76.
- Toutoungi DE, Lu TW, Leardini A, et al. 2000. Cruciate ligament forces in the human knee during rehabilitation exercises. *Clin Biomech* 15:176-177.
- Nagura T, Dyrby CO, Alexander EJ, et al. 2002. Mechanical loads at the knee joint during deep flexion. *J Orthop Res* 20:881-886.
- Komistek RD, Scott RD, Dennis DA, et al. 2002. In vivo comparison of femorotibial contact positions for press-fit posterior stabilized and posterior cruciate-retaining total knee arthroplasties. *J Arthroplasty* 17:209-216.
- Incavo SJ, Mullins ER, Coughlin KM, et al. 2004. Tibiofemoral kinematic analysis of kneeling after total knee arthroplasty. *J Arthroplasty* 19:906-910.
- Nakayama K, Matsuda S, Miura H, et al. 2005. Contact stress at the post-cam mechanism in posterior-stabilized total knee arthroplasty. *J Bone Joint Surg Br* 87:483-488.
- Garling EH, Kaptein BL, Geleijns K, et al. 2005. Marker configuration model-based roentgen fluoroscopic analysis. *J Biomech* 38:893-901.
- Fukuoka Y, Hoshino A, Ishida A. 1999. A simple radiographic measurement method for polyethylene wear in total knee arthroplasty. *IEEE Trans Rehabil Eng* 7:228-233.
- Banks SA, Hodge WA. 1996. Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy. *IEEE Trans Biomed Eng* 43:638-649.
- Higaki H, Shimoto T, Miura H, et al. 2002. Application of pattern matching method in motion analysis for TKA. *Trans Jpn Soc Mech Eng* 68:3061-3068.
- Shimoto T, Higaki H, Hukunaga T, et al. 2003. Study of the motion analysis for total knee arthroplasty using flat panel detector. Proceedings of the IASTED International Conference on Biomechanics. Calgary, Canada: ACTA Press; p 28-33.
- Hazaki S, Yokoyama Y, Inoue H. 2001. A radiographic analysis of anterior-posterior translation in total knee arthroplasty. *J Orthop Sci* 6:390-396.
- Bertin KC, Komistek RD, Dennis DA, et al. 2002. In vivo determination of posterior femoral rollback for subjects having a NexGen posterior cruciate-retaining total knee arthroplasty. *J Arthroplasty* 17:1040-1048.
- Nozaki H, Banks SA, Suguro T, et al. 2002. Observations of femoral rollback in cruciate-retaining knee arthroplasty. *Clin Orthop* 404:308-314.
- Most E, Zayontz S, Li G, et al. 2003. Femoral rollback after cruciate-retaining and stabilizing total knee arthroplasty. *Clin Orthop* 410:101-113.
- Churchill DL, Incavo SJ, Johnson CC, et al. 2001. The influence of femoral rollback on patellofemoral contact loads in total knee arthroplasty. *J Arthroplasty* 16:909-918.
- Browne C, Hermida JC, Bergula A, et al. 2005. Patellofemoral forces after total knee arthroplasty: effect of extensor moment arm. *Knee* 12:81-88.
- Wasielewski RC, Galante JO, Leighty RM, et al. 1994. Wear patterns on retrieved polyethylene tibial inserts and