Biomechanical Consequences of Anterior Femoral Notching in Cruciate-Retaining Versus Posterior-Stabilized Total Knee Arthroplasty

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Abstract

Anterior femoral notching during total knee arthroplasty is a potential risk factor for periprosthetic supracondylar femur fracture.

We conducted a study to determine if the design of the femoral implant changes the risk for periprosthetic supracondylar femur fractures after anterior cortical notching. An anterior cortical defect was created in 12 femoral polyurethane models. Six femora were instrumented with cruciate-retaining implants and 6 with posterior-stabilized implants. Each femur was loaded in external rotation along the anatomical axis. Notch depth and distance from anterior cortical notch to implant were recorded before loading, and fracture pattern was recorded after failure.

There were no statistically significant differences in notch depth, distance from notch to implant, torsional stiffness, torque at failure, final torque, or fracture pattern between cruciate-retaining and posterior-stabilized femoral component designs.

Periprosthetic fracture after anterior femoral notching is independent of the bone removed from the intercondylar notch. After notching, there likely is no significant difference in femoral strength in torsion between cruciate-retaining and posterior-stabilized designs.

Although rare, periprosthetic fractures remain a significant complication after total knee arthroplasty (TKA), occurring in 0.3% to 2.5% of cases.[1-4] Hirsh and colleagues[5] were among the first to suggest that anterior femoral notching during TKA was a potential risk factor for postoperative periprosthetic femoral fracture because notching may weaken the anterior femoral cortex. Anterior femoral notching, a cortex violation occurring during an anterior bone cut, occurs in up to 30% of cases.[6] Using a theoretical biomechanical model, Culp and colleagues[1] found that increasing the depth of the notch defect into the cortex led to reduced torsional strength. In more recent, cadaveric biomechanical studies, notching of the anterior femoral cortex decreased torsional strength by up to 39%.[7,8] Contrary to these biomechanical studies, a retrospective study evaluating 1089 TKAs using 2 implant designs (Anatomic Graduated Component, Biomet and Legacy, Zimmer) demonstrated no significant effect of anterior femoral notching with respect to incidence of supracondylar femur fractures.[6] That study, however, did not address whether implant design is associated with a differential risk for fracture in the presence of anterior notching.

Previous biomechanical studies have primarily investigated cruciate-retaining (CR) femoral components and properties with respect to anterior notching, even though the posterior-stabilized (PS) design is used more often in the United States.[1,7] According to a Mayo Clinic survey, TKAs with a PS design increased from <10% in 1990 to almost 75% by 1997.[9] Today, there is little to no consensus about which implant is better, and use of one or the other depends largely on the surgeon and

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varies widely between countries and regions.[10] PS designs require more bone resection and demonstrate prosthesis-controlled rollback during flexion, whereas CR designs preserve more bone and achieve posterior stabilization via the posterior cruciate ligament.[11] Despite these differences in design and mechanics, a 2013 Cochrane review of TKA design found no clinically significant differences between CR and PS with respect to pain, range of motion, or clinical and radiologic outcomes.[10] The reviewers did not specifically address periprosthetic fractures associated with either femoral notching or TKA design, as they could not quantitatively analyze postoperative complications because of the diversity of reports. Given the limited number of reported cases, a review of radiographic findings pertaining to the characteristics of supracondylar fractures in anterior femoral notching was unsuccessful.[12] As the previous biomechanical studies of anterior notching used primarily CR models or no prostheses at all, a study of biomechanical differences between CR and PS designs in the presence of anterior notching is warranted.[1,7,8] Therefore, we conducted a study to assess the effect of anterior femoral notching on torsional strength and load to failure in CR and PS femoral components.

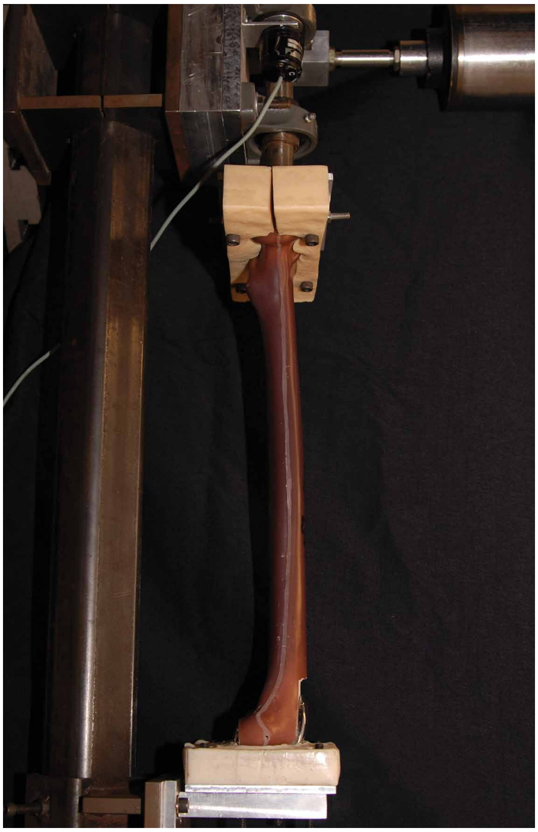

Figure 1. Test specimen loaded on torsional testing apparatus.

Materials and Methods

Twelve fourth-generation composite adult left femur synthetic sawbones (Sawbones; Pacific Research Laboratories) were selected for their consistent biomechanical properties, vs those of cadaveric specimens; in addition, low intersample variability made them preferable to cadaveric bones given the small sample used in this study.[13,14] All bones were from the same lot. All were visually inspected for defects and found to be acceptable. In each sample, an anterior cortical defect was created by making an anterior cut with an undersized (size 4) posterior referencing guide. In addition, the distance from the proximal end of the notch to the implant fell within 15 mm, as that is the maximum distance from the implant a notch can be placed using a standard femoral cutting jig.[15] Six femora were instrumented with CR implants and 6 with PS implants (DePuy Synthes). Implants were placed using standardized cuts. Before testing, each implant was inspected for proper fit and found to be securely fastened to the femur. In addition, precision calipers were used to measure notch depth and distance from notch to implant before loading. A custom polymethylmethacrylate torsion jig was used to fix each instrumented femur proximally and distally on the femoral implant (**Figure 1**). Care was taken to ensure the distal jig engaged only the implant, thus isolating the notch as a stress riser. Each femur was loaded in external rotation through the proximal femoral jig along the anatomical axis. Use of external rotation was based on study findings implicating external rotation of the tibia as the most likely mechanism for generating a fracture in the event of a fall.[12] Furthermore, distal femur fractures are predominantly spiral as opposed to butterfly or bending—an indication that torsion is the most likely mechanism of failure.[16] With no axial rotation possible within the prosthesis, increased torsional stress is undoubtedly generated within adjacent bone. Each specimen underwent torsional stiffness testing and then load to failure. Torsional stiffness was measured by slowly loading each femur in external rotation, from 1 to 18 Nm for 3 cycles at a displacement rate of 0.5° per second. Each specimen then underwent torsional load-to-failure testing on an Instron 5800R machine at a rate of 0.5° per second. Failure was defined as the moment of fracture and subsequent decrease in

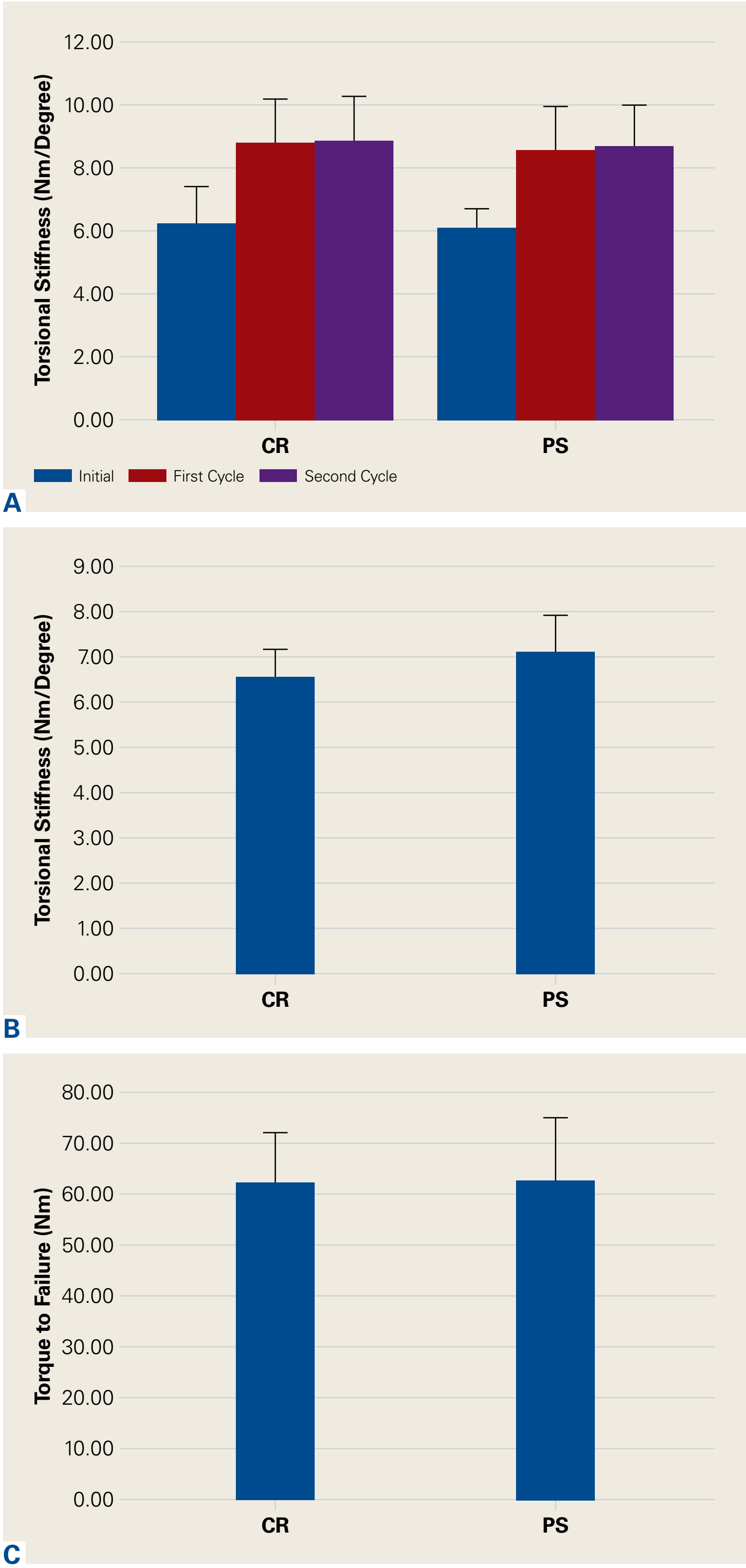

Figure 2. Three comparisons: (A) Torsional stiffness over 3 cycles and until failure for cruciate-retaining (CR) and posterior-stabilized (PS) designs (P = .82, .77, .84). (B) Stiffness at failure for CR and PS designs (P = .24). (C) Torque to failure for CR and PS designs (P = .95).

torsional load—determined graphically by the peak torsional load followed immediately by a sharp decrease in load. Stiffness was determined as the slope of torque to the displacement curve for each cycle, and torque to failure was the highest recorded torque before fracture. Fracture pattern was noted after failure. A sample size of 6 specimens per group provided 80% power to detect a between-group difference of 1 Nm per degree in stiffness, using an estimated SD of 0.7 Nm per degree. In our statistical analysis, continuous variables are reported as means and SDs. Data from our torsional stiffness and load-to-failure testing were analyzed with unpaired 2-sample t tests, and $P < .05$ was considered statistically significant.

Results

We did not detect a statistical difference in notch depth, notch-to-implant distance, or femoral length between the CR and PS groups. Mean (SD) notch depth was 6.0 (1.3) mm for CR and 4.9 (1.0) mm for PS (P = .13); mean (SD) distance from the proximal end of the notch to the implant was 13.8 (1.7) mm for CR and 11.1 (3.2) mm for PS (P = .08); and mean (SD) femoral length was 46.2 (0.1) cm for CR and 46.2 (0.1) cm for PS (P = .60).

Mean (SD) torsional stiffness for the first 3 precycles was 6.2 (1.2), 8.7 (1.5), and 8.8 (1.4) Nm per degree for the CR group and 6.0 (0.7), 8.4 (1.4), and 8.6 (1.4) Nm per degree for the PS group; the differences were not statistically significant (**Figure 2A**). In addition, there were no statistically significant differences in mean (SD) stiffness at failure between CR, 6.5 (0.7) Nm per degree, and PS, 7.1 (0.9) Nm per degree (P = .24; **Figure 2B**) or in mean (SD) final torque at failure between CR, 62.4 (9.4) Nm, and PS, 62.7 (12.2) Nm (P = .95; **Figure 2C**).

All fractures in both groups were oblique fractures originating at the proximal angle of the notch and extended proximally. None extended distally into the box. Fracture locations and patterns were identical in the CR and PS groups of femurs (**Figure 3**).

Discussion

Periprosthetic fractures after TKA remain rare. However, these fractures can significantly increase morbidity and complications. Anterior femoral notching occurs inadvertently in 30% to 40% of TKAs.[6,17] The impact of femoral notching on supracondylar femur fracture is inconsistent between biomechanical and retrospective clinical studies. Retrospective studies failed to find a significant correlation between anterior femoral notching and supracondylar femur fractures.[6,17] However, findings of biomechanical studies have suggested that a notch 3 mm deep will reduce the torsional strength of the femur by 29%.[7] Another study, using 3-dimensional finite element analysis, showed a significant increase in local stress with a notch deeper than 3 mm.[15]

To our knowledge, no clinical studies, including the aforementioned Cochrane review,[10] have specifically evaluated the difference in risk for periprosthetic fracture between different TKA models in the presence of notching.[11] The biomechanical differences between implant designs could be a confounding factor in the results of past studies. More bone resection is required in PS designs than in CR designs. The position of the PS intercondylar cutout, much lower than the top of the patella flange, should not increase susceptibility to fractures more than in CR designs, but this hypothesis, though accepted, has not been validated biomechanically or addressed specifically in prospective or retrospective clinical analysis. In the present study, we used a biomechanical model to replicate an external rotation failure mechanism and quantify the differences in torsional strength and load to failure between CR TKA and PS TKA models in the presence of anterior femoral notching. Our results showed no significant differences in torsional stiffness, stiffness at failure, or torque at failure between the CR and PS design groups in the presence of anterior femoral notching.

In this study, all femoral fractures were oblique, and they all originated at the site of the cortical defect, not the notch—a situation markedly different from having bending forces applied to the femur. Previous biomechanical data indicated that bending forces applied to a notched femur cause fractures originating at the notch, whereas torsional forces applied to a notched femur cause fractures originating at the anterior aspect of the bone–component interface.[7] The difference is attributable to study design. Our femurs were held fixed at their proximal end, which may have exacerbated any bending forces applied during external rotation, but we thought constraining the proximal femur would better replicate a fall involving external rotation.

More important for our study, an oblique fracture pattern was noted for both design groups (CR and PS), indicating the fracture pattern was unrelated to the area from which bone was resected for the PS design. All femur fractures in both design groups occurred proximal to a well-fixed prosthesis,

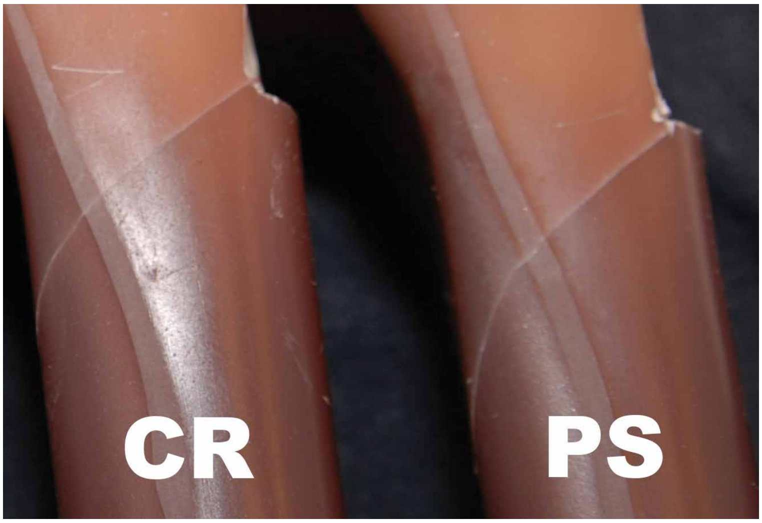

Figure 3. Fracture pattern in cruciate-retaining (CR, left) and posterior-stabilized (PS, right) femora when torqued to failure. Rest of femora exhibited identical fracture pattern and are omitted from image for clarity. Femora pictured distal on top, proximal on bottom, anterior on the right, and posterior on the left.

indicating they should be classified as Vancouver C fractures. This is significant because intercondylar fossa resection (PS group) did not convert the fractures into Vancouver B2 fractures, which involve prosthesis loosening caused by pericomponent fracture.[18] This simple observation validated our hypothesis that there would be no biomechanical differences between CR and PS designs with respect to the effects of anterior femoral notching. This lack of a significant difference may be attributed to the PS intercondylar cutout being much lower than the top of the anterior flange shielding the resected bone deep to the anterior flange.[7] In addition, given the rarity of supracondylar fractures and the lack of sufficient relevant clinical data, it is difficult to speculate on the fracture patterns observed in clinical cases versus biomechanical studies.[12]

The use of synthetic bone models instead of cadaveric specimens could be seen as a limitation. Although synthetic bones may not reproduce the mechanism of failure in living and cadaveric femurs, the mechanical properties of synthetic bones have previously been found to fall within the range of those of cadaveric bones under axial loading, bending, and torsion testing.[13,14]

As a uniform testing material, synthetic bones allow removal of the confounding variations in bone size and quality that plague biomechanical studies in cadaveric bones.[13,14] Interfemoral variability was 20 to 200 times higher in cadaveric femurs than in synthetic bones, which makes synthetic femurs preferable to cadaveric femurs, especially in studies with a small sample size.[13,14] In addition, a uniform specimen provides consistent, reproducible osteotomies, which were crucial for consistent mechanical evaluation of each configuration in this study.

The long-term clinical significance of anterior femoral notching in periprosthetic fractures is equivocal, possibly because most studies predominantly use CR implants.[6] This may not be an issue if it is shown that CR and PS implants have the same mechanical properties. Despite the differences between clinical studies and our biomechanical study, reevaluation of clinical data is not warranted given the biomechanical data we present here. Results of biomechanical studies like ours still suggest an increased immediate postoperative risk for supracondylar fracture after anterior cortical notching of the femur.[5,7] Ultimately, this study found that, compared with a CR design, a PS design did not alter the torsional biomechanical properties or fracture pattern of an anteriorly notched femur.

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