Feature Article

In Vitro Analysis of Periprosthetic Strains Following Total Knee Arthroplasty

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ABSTRACT

Clinical case studies have disclosed certain risk factors associated with periprosthetic fracture in elderly patients. How the mechanical strength of the distal femur is changed by total knee arthroplasty (TKA) has not been elucidated. Using elderly cadaveric femora, this study evaluated both periprosthetic strains and associated fracture patterns arising from an in vitro simulation of a fall onto the distal femur. The data showed a significant increase in anterior and posterior mechanical strain following TKA. Neither stemless nor stemmed versions of two cemented Howmedica prostheses (Rutherford, NJ) reduced distal femur strains to baseline values. However, neither produced a disproportionate frequency of periprosthetic fractures. Although not formally evaluated herein, bone geometry/density may contribute more profoundly to the occurrence of periprosthetic fracture than the implants tested.

Ipsilateral supracondylar fracture following total knee arthroplasty (TKA) represents a significant and devastating injury to the elderly. The estimated incidence is between 0.5% and 3% of >143,000 operations annually.1 This rate is substantially higher in rheumatoid arthritis patients.6

Little trauma is required to induce this injury.5,7 The etiology of these periprosthetic fractures is a fall or blow to the knee. Although Culp et al5 have suggested torsional force, the actual force dynamics and kinematics of this injury, remain ambiguous.

Numerous treatment strategies have been devised for this injury pattern. Nonoperative management requires lengthy convalescence with attendant risks such as nonunion or malunion, depletion of physical reserve, and thromboembolism. Operative treatment also may pose similar risks, as well as infectious complications.4,6,8,9

Implant design also may contribute to periprosthetic fracture. Current trends to produce lower profile devices with a prominent anterior flange may reduce stress in the anterior cortex, resulting in weakened bone structure secondary to stress shielding.10 Femoral components also may contain an intramedullary stem that is impressed into the bone canal. The effect of stemmed versus unstemmed systems on the mechanical strength in this area of the femur is unknown.

Few studies have addressed the biomechanical aspects of supracondylar fractures following TKA. Culp et al5 theoretically posited that a 3-mm violation of the anterior cortex of the femur decreases torsional strength by 27.2%. Most studies, however, have been clinical reviews describing predisposing factors, management suggestions, and expected outcomes.11-17

Current opinion holds that anterior notching as well as rheumatoid arthritis, osteopenia, osteoporosis, and neurologic disorders are major risk factors in the development of these fractures.3,5,8,9 Actual assessment of the combined effect of arthroplasty and implant

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design on the mechanical strength of the distal femur has not been addressed.

This in vitro study analyzed the changes to normal strain patterns in the distal femur associated with TKA. Both stemmed and unstemmed implants were compared for their effect on distal femur mechanical integrity as well as resultant fracture patterns ensuing from the simulated fall.

**MATERIALS AND METHODS**

**Testing Device**

Supracondylar fracture patterns were reviewed in both the literature and in radiographs of clinical cases. Fractures above the knee implant, indicative of a blow or fall to the knee, were apparent in all (Figure 1). This review, however, failed to reveal a single biomechanical stress responsible for the injury as compressive, shear, and torsional forces were all evident. To provide the most practical, reproducible fall simulation, a vertical impact device was developed (Figure 2).

The device consisted of a 50-kg drop sled constrained by a pair of 2.4-m vertical metal rails. The bottom of the rails spanned a 30×30×0.9-cm strike plate, that was secured to a steel reinforced 70×122×1.2-cm platform. A vertical steel cross beam secured the rails at the top of the assembly. Ball-and-roller guide wheels were located on the outboard edges of the sled and provided near frictionless contact with the guide rails. This configuration facilitated a continuum of drop heights, from 1.9 m to <2 cm.

To secure specimens to the sled, an adjustable pivot clamp was devised. The pivoting adjustment permitted selection of a fixed amount of angulation between the specimen and the strike plate. This pivoting mechanism was designed to allow from 0°-90° of inclination from the vertical.

All tests were conducted with a fixed angle of 15°. This permitted loading of the anteroinferior surface of the condyles of the specimen. Additionally, specimens were installed in the fixture to ensure bicondylar contact at the moment of impact. These parameters provided uniform force distribution between both condyles at the anteroinferior surfaces. As such, it represented a best approximation of in situ loading during a fall onto the knee.

**Specimen Preparation**

Ten pairs of cadaveric femurs, obtained from elderly donors, were obtained from a regional anatomic supplier (National Anatomical Services, Parlin, NJ). Pairing was based on visual similarity of specimen size and geometry, and did not necessarily represent donor pairing.

Specimen preparation included resection of the proximal femur at the level of the lesser trochanter and a plaster wrap of the remaining proximal half of the bone (Johnson and Johnson, New Brunswick, NJ). Following plaster hardening, this section of bone was encased in a 10-cm diameter polyvinylchloride drainage casing containing approximately 800 mL of a 5:1 mixture of West Systems 105 epoxy resin and West Systems 205 hardener (Gougeon Brothers, Bay City, Mich). The use of a plaster insulator between bone and epoxy was developed empirically to minimize tissue degradation during exothermic hardening of the epoxy. As approximately 2-3 cm of the plaster envelope extended beyond the level of the epoxy, the stress concentration at the bone epoxy interface also was lessened.

At the distal end of each specimen, anterior and posterior cortical surfaces were cleaned and finished in preparation for strain gauge attachment. Four 350Ω linear gauges (Measurements Group Inc, Raleigh, NC) were then cemented to the surfaces using a butyl-cyanoacrylate adhesive (3M Verbond, St Paul, Minn). This adhesive was selected for elastic accuracy rather than postyield measurement.

Posterior gauges were located on the posteromedial and posterolateral columns approximately 2.5 cm from the respective epicondyles. On the anterior surface, two gauges were aligned axially with the diaphysis and located 6 cm and 8 cm proximal to the intercondylar fossa.

Appropriateness of gauge locations was derived from geometric considerations of the specimens and photoelastic stress analysis (unpublished data). Some variation occurred in gauge placement between specimens as a result of variability in the locations and numbers of nutrient foramen in these regions.

Kinematic (70-mm stem) and Kinematic II (stemless) implants were
supplied by Howmedica Corporation (Rutherford, NJ). Arthroplasty consisted of resurfacing condylar areas in accordance with implant geometry per manufacturer recommendations.

Implants were cemented into place using Simplex P bone cement (Howmedica, Rutherford, NJ). Nine specimens were used for stemmed implant trials and 11 for stemless. However, cortical notching occurred in 1 stemless specimen. Although mechanically evaluated, the data were not included in the statistical analysis of strains or fracture patterns.

**Testing Protocol**

Trials consisted of static and dynamic specimen loading. Loading was furnished by the 50-kg sled mass. To account for elastic creep properties in bone, data for static loading were recorded 1 minute after the application of load per observations of Smith and Walmsley.\(^\text{18}\)

Dynamic tests were subdivided into nondestructive and destructive tests using drop heights of 1.3 and 20 cm, respectively. Typically, destructive tests provided little strain data owing to instantaneous catastrophic failure of the gauges. However, these tests were used to record fracture patterns associated with implant design. Throughout testing, specimens were liberally hydrated with saline solution to minimize tissue shrinkage.\(^\text{18}\)

Instrumentation to capture dynamic strains consisted of Model 3800 Strain Indicators (Measurement Group Inc, Raleigh, NC) connected in series to Tektronix Model TDS 320 digital storage oscilloscopes (Beaumont, Ore). Hard copies of the measurements were obtained with an inkjet printer connected to the oscilloscope output.

To validate the accuracy of strain data, a 3-point bending test was performed on a bovine femur. This consisted of attaching a strain gauge to the approximate center of a bovine femur and applying a load on the opposing side while supporting the ends. A 20,000-lb Universal Test Machine (Vega, Decatur, Ill) was used for force induction up to 4000 lbs. After measuring the strains, the bone was sectioned and the cross-sectional area was digitized so the moment of inertia could be numerically determined. The strain and moment of inertia data produced a Young's Modulus that was consistent with published values for bone.

The experimental design for strain evaluation consisted of a repeat measure strain means vector. Femurs were instrumented, statically and nondestructively loaded, and then modified via TKA and installation of the prosthesis.

Periprosthetic strains were then measured under static and nondestructive loads followed by catastrophic loading to induce fracture. This repeat measures assessment was an outgrowth and refinement of initial trials that relied on strain measurement of matched or similarly dimensioned specimens.

Early trials consisted of two stemless (including a notched specimen) and one stemmed implants, and one control. This matching, however, inadequately accounted for slight differences in bone geometry as well as in strain gauge placement. Using the same femur to obtain both control and peri-prosthetic strain data eliminated this variation.

**Statistical Analysis**

Sixteen of 20 specimens were evaluated by the repeat measures strategy. Statistical analysis was based on these 16 specimens using a repeat measures analysis of variance. Post hoc tests were used to disclose strain differentials based on implant design as well as the most strain-labile site. The level of significance was set at .05.

**Fracture Patterns**

Fracture patterns were broadly categorized into two types: periprosthetic and diaphyseal. Differences in the proportionate types of fractures between implant types were determined by Fisher’s exact test. Fracture patterns for all 19 specimens (excluding the notched specimen) were included in this analysis using a level of significance of .05.

**RESULTS**

Figures 3-6 depict periprosthetic strains as a function of implant design, loading, and location. Steamed implants produced the highest mean compression and tension values for both static (−361±174 μE and 204±89 μE) and dynamic (−3437±733 μE and 2178±820 μE) loading conditions, respectively.

The lowest static strains, −174±85 μE and 94±48 μE, were recorded for stemless implant specimens. Under dynamic conditions, however, the control (nonsurgical) condition registered the smallest strains (−189±542 μE and 893±456 μE). Compared to stemmed implants, these control data were 82% and 140% lower.

As inferred graphically, dynamic loading induced strains that were from 7.1-12.5 times static levels (P<.001). A trend toward higher strain levels in stemmed implants also was evident, although the differences between implant type were not statistically significant (P=.129). This was further corroborated by the absence of an interaction effect between implant design and loading force (P=.152). However, TKA significantly increased strain levels (P<.001). This effect was exacerbated by dynamic loading (eg, arthroplasty-loading interaction, P<.001).

While some data overlap, mean strains derived from each gauge location were significantly different (P<.001). Generally, the largest anterior strains were observed at the proximal gauge location, whereas the highest posterior strains occurred at the lateral site. The differences were not attributable to contact bias between the lateral and medial condyles as set-up efforts ensured uniform contact.

While strain magnitude by location was not influenced by implant design (P=.996), arthroplasty significantly increased the strain at each site (P=.031). Further, changes in the strain distribution pattern also were evident.
The posteromedial site registered the largest static strain in controls, whereas the posterolateral aspect displayed the largest strains in post-arthroplasty specimens.

As suggested by the strain data, a notched stemless femur did not produce excessively high strains at any strain gauge location (Figures 3-6).

Shortly after these data were acquired, however, the specimen fractured after a drop that was nondestructive for other specimens.

Postcatastrophic impacts failed to reveal a trend in fracture patterns between stem and stemless implants (Table). Differences in the incidence of periprosthetic fractures between implant types were 6%-7%. These differences were not statistically significant (P=.575).

**DISCUSSION**

The data show that post-arthroplasty specimens exhibited the highest distal femur strains in response to an impact of the anteroinferior surface of the femoral condyles. In general, the areas of greatest stress were the posterolateral and anterior proximal aspects of the distal femur. Stemmed implants failed to restore some of the mechanical strength lost to arthroplasty. This was further evidenced by the similarity in the rates of supracondylar fractures associated with each type of implant following catastrophic impacts.

A limitation of this study was the failure to obtain radiographic and bone density estimates for each of the specimens. Martens et al. found that cross-sectional geometry profoundly affected the bending behavior of cadaver femurs in a 4-point bending test. This was manifested by large variations in bending parameters for the specimens, a function of bone inertial properties. They further delineated two distinct load-deformation curves for two classes of fractures: one that fractured centrally and another that fractured at the distal third of the specimen. They attributed the latter to the inherent susceptibility of these specimens to combined bending-shear force interaction in this region, a function of bone morphology. Thus, bone geometry and cross-sectional area may exert the most profound influences on fracture pattern rather than the implant design (eg, stemmed versus stemless).

Courtney et al. showed a generalized decrease in bending strength and fracture energy in the femoral neck of specimens derived from older donors. They related this difference to decreased bone-mineral density and cross-sectional area, indicating that these specimens possessed no more than half of the mechanical strength needed to sustain a typical fall. They further suggested the direction of the
fall, in concert with the amount of soft-tissue cushioning, may dictate the production of a hip fracture.

Akin to femoral neck fractures, the direction and associated momentum of a fall would influence the potential for periprosthetic fracture at the knee. As rigorous detail was absent, it was postulated that some measure of hip flexion is associated with the event and arrived at 10°-15° from vertical. An additional supposition included a fairly high coefficient of friction on knee impact that would culminate in a combination of torsion and shear at the condyles.

Our test set-up mimicked this conceptualization of knee fracture kinematics, ultimately producing impact on the anteroinferior surface of the condyles of the test specimens. This set-up, however, may or may not adequately represent the in vivo event. Furthermore, the contributions of the patella to this fracture pattern could not be ascertained.

The subject of cortical notching has not been fully appreciated either biomechanically or clinically. Sisto et al.16 noted few supracondylar fractures for patients with measurable notching on radiographic images.3 We found that strain data were not remarkably high for a specimen with a defect of at least 5 mm, yet fracture occurred at a nondestructive drop height. Strain data, however, may have been of limited usefulness as only one specimen was evaluated, and strain gauge placement was inevitably altered by the presence of the notch. (Gauges could not be directly affixed to exposed trabecular bone).

Sisto et al.16 cite remodeling as the mechanism that counteracts notching, thereby reducing the risk of periprosthetic fracture. However, individuals with chronic bone metabolic disease render this scenario dubious. To the extent that disease compromises remodeling, any loss of bone stock would be detrimental. This was made clear by our study, which suggests arthroplasty significantly increased mechanical strain in the distal femur.

**CONCLUSION**

Under nondestructive loading conditions, mechanical strain to bone in the distal femur can be increased up to 140% following arthroplasty. Whereas arthroplasty significantly elevated dis-

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**TABLE**

<table>
<thead>
<tr>
<th>Implant</th>
<th>No. (%) Periprosthetic</th>
<th>No. (%) Diaphyseal</th>
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<tbody>
<tr>
<td>Stemless (n=10)</td>
<td>4 (40)</td>
<td>6 (60)</td>
</tr>
<tr>
<td>Stem (n=9)</td>
<td>3 (33)</td>
<td>6 (66)</td>
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*P=.575.

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**Figure 5:** Comparison of strains originating in the posteromedial location. Static loads were supplied by the 50-kg impact sled. Nondestructive dynamic loads were imparted by a 1.3-cm drop of the sled. Median values are depicted by the dark horizontal bars within the box plots.

**Figure 6:** Comparison of strains originating in the posterolateral location. Static loads were supplied by the 50-kg impact sled. Nondestructive dynamic loads were imparted by a 1.3-cm drop of the sled. Median values are depicted by the dark horizontal bars within the box plots.
tal femur strains, no significant difference occurred between stemmed and unstemmed implants. Further, neither implant affects the frequency of periprosthetic fractures in vitro.

Although bony ingrowth in vivo might be expected to lessen this observed reduction in mechanical strength, healing time, as well as metabolic bone disease, could result in a period of fracture susceptibility. Although not evaluated herein, we concur with other investigators that bone density may have a profound influence on the incidence of these fractures, perhaps exerting a greater effect than implant design.

REFERENCES


EDITORIAL DISCUSSION

ORTHOPEDICS: With periprosthetic bone loss in the distal femur following TKA, and with your findings of significantly increased strain in the cadaver model, why isn't the prevalence of supracondylar fractures following TKA much higher than it is?

PARRY et al: The risk of these fractures is multifactorial, requiring the development of a force that exceeds the yield point of the periprosthetic bone. Bone geometry and density help determine resistance to fracture. Further, not all falls potentiate a sufficient force level to induce a break. Soft tissue, the patella, or both, may be highly influential in acting to deflect or absorb energy. Finally, the highest risk of fracture occurring would certainly be during the immediate postoperative period, a time when ambulation is governed by the rehabilitation protocol, as well as pain and effusion. As symptoms subside, ambulation levels increase, thereby increasing the risk of a fall. However, symptom resolution takes time during which bony ingrowth is initiated about the implant. Overall, the probability of sustaining this type of fracture progressively lessens with convalescence time.