Effect of Distal Stem Geometry on Interface Motion in Uncemented Revision Total Hip Prostheses

Kevin L. Kirk, DO, Benjamin K. Potter, MD, Ronald A. Lehman, Jr., MD, and John S. Xenos, MD

Abstract

In this study, we compared differences in motion at the bone-prosthesis interface in femora in which a fluted, tapered, or cylindrical distal stem design had been implanted in a revision total hip arthroplasty model.

Paired, fresh-frozen, cadaveric femora underwent resection of the proximal femur to simulate the proximal femoral bone loss often present during revision total hip arthroplasty and implantation with either a fluted, tapered stem or a clinically proven cylindrical stem. Specimens were then preloaded and subjected to a synchronous axial and torsional load with continuous monitoring of axial displacement and rotation.

For the fluted, tapered stem, mean axial and rotational displacements were 13.33 µm and 9.81 µm, respectively, compared with 18.37 µm and 13.40 µm for the cylindrical stem (both *P*s<.05). Therefore, the fluted, tapered stem design that was tested demonstrated superior initial biomechanical stability compared with that of the clinically proven cylindrical design tested. However, both stems demonstrated motion below the threshold necessary for bony ingrowth. Knowledge of the initial biomechanical properties of different stem designs may assist the revision joint surgeon in choosing the optimal prosthesis for implantation.

A critical factor for the success of cementless femoral components is to achieve rigid initial fixation so that micromotion between the implant and bone is low along the entire length of the bone.[1-3] Rigid implant stability, avoiding motion at the bone-prosthesis interface, is essential for ingrowth of bone into porous surfaces.[4,5] Therefore, the relative magnitude of early implant-bone movement is important in the design and use of implants chosen for both primary and revision total hip arthroplasty (THA).

Preliminary fixation is achieved first by surgical preparation of the bone cavity to optimally fit the implant and then by impaction of the prosthesis, leading to a solid press-fit between implant and bone with the intention of creating minimal relative motion. Initial stability, the amount of motion present immediately after surgery, depends on the geometric and mechanical properties of the prosthesis, the accuracy of the preparation of the bone bed, and the quality of the patient's bone.[6] Development of femoral components that provide increased rigidity of initial fixation may provide for optimal clinical results.

Choice of revision hip implant depends on several factors, including extent of loss of proximal femoral bone, quality of remaining host bone, the patient activity level, and surgeon experience. The initial stability of cementless femoral implants relies partially on distal fixation. In revision THA, distal fixation is critical, as proximal bone stock is often insufficient, particularly if an extended trochanteric osteotomy has been used. In some patients, the degree of proximal bone loss necessitates achieving adequate stability by distal fixation alone. Kendrick and colleagues,[3] in a comparison of distal stem design and the torsional stability of cementless femoral stems, demonstrated that a solid, fluted stem was the only design to show sufficient resistance to torsional forces to stabilize a femoral prosthesis solely through distal fixation within the medullary canal.

We hypothesized that a difference in distal stem design would affect the amount of early motion between the stem and the surrounding bone. The purpose of our study was to compare differences in motion at the bone–prosthesis interface in femora in which a fluted, tapered, or clinically proven, cylindrical distal stem design had been implanted in a revision THA model.

Materials and Methods

We selected a newer fluted distal stem design component (Link® MP™ Hip Stem; Link America, Pine Brook, NJ) and a cylindrical distal stem design (Solution System® Hip Stem; Depuy, Warsaw, Ind) because the senior surgeon (Dr. Xenos) commonly used them at our institution and because the latter stem has an excellent long-term clinical track record, functioning thus as a proven clinical control. The Link stem is made of titanium alloy and has a tapered, fluted distal portion of the femoral stem and a 70-micron pore texture with

Dr. Kirk is Foot and Ankle Fellow, Union Memorial Hospital, Baltimore, Maryland. He was a resident at the time the article was written.

Dr. Potter is Chief Resident and Dr. Lehman is Director of Adult and Pediatric Spine Surgery, Department of Orthopaedics and Rehabilitation, Orthopaedic Surgery Service, Walter Reed Army Medical Center, Washington, DC, and Department of Surgery, Uniformed Services University of the Health Sciences, Bethesda, Maryland.

Dr. Xenos is Adult Reconstructive Surgeon, Premier Orthopedics, Colorado Springs, Colorado.

Requests for reprints: Benjamin K. Potter, MD, Walter Reed Army Medical Center, 6900 Georgia Avenue, NW, Building 2, Clinic 5A, Washington, DC, 20307 (tel, 202-782-6574; fax, 202-782-6845; e-mail: kyle.potter@us.army.mil).

Am J Orthop. 2007;36(10):545-549.

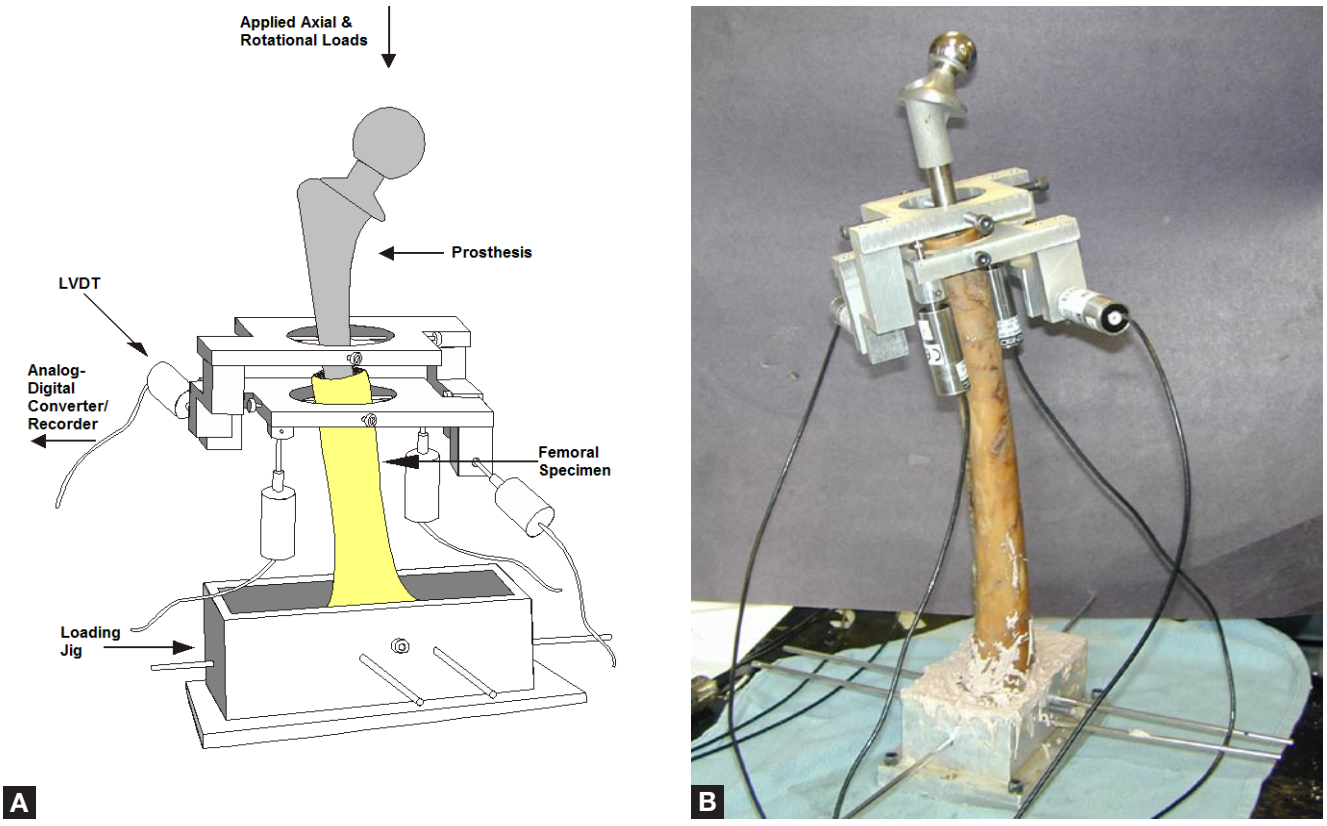

Figure 1. Diagram **(A)** and laboratory photograph **(B)** of the testing apparatus setup designed to measure axial and rotational bone–prosthesis motion through linear variable displacement transducers (LVDTs) attached to an analog–digital converter and recorder.

a 3° angle proximally. The Solution stem is made of cobalt chromium with an extensive porous coating and has a cylindrical distal stem design and tapered distal tip.

Ten pairs of fresh-frozen femora (20 femora) from the cadavera of 8 female and 2 male donors were used for implantation. All femora were stripped of soft tissues, placed sealed in polyethylene bags, frozen at -20°C, and grossly examined. X-rays were obtained to ensure there were no structural defects. Dual-energy x-ray absorptiometry scans were obtained (Hologro QDR 2000 scanner; specimens tested on a bed of rice in an acrylic tank), and bone mineral density (BMD) was recorded for each specimen.

The femora were thawed at room temperature before implantation. The femoral prostheses (Link or Solution stem) were randomly assigned to either the right or the left femur. When the right or left femur received the Link stem, the other femur from the same cadaver received the Solution stem. After the proper size of the prosthesis for maximum filling of the femoral canal had been determined with a template, a transverse osteotomy of the proximal femur was performed at a point 12 cm distal to the greater trochanter, and the proximal portion was discarded to simulate proximal femoral bone loss that may be present at the time of revision THA. This allowed biomechanical stability assessment on the basis of distal stem fixation only. Dr, Xenos, who has clinical experience with both devices, performed canal reaming and prosthesis implantation in the manner described by the manufacturers of each prosthesis. To reduce micromotion variability caused by length of cortical contact between specimens, both prostheses were templated and implanted to a depth of 10 cm in the diaphyseal femoral segments tested. X-rays were repeated after implantation to evaluate for any occult fractures not recognized during implantation.

Testing Protocol

The implanted femora were potted distally in 6° of valgus in an aluminum pot with the use of epoxy resin, and fixation screws were then mounted in a loading jig that was attached to a material testing machine (Bionix Test System model 858; MTS Systems Corp, Eden Prairie, Minn). Linear variable displacement transducers (LVDTs; Sensotec model 060-3611-02; Sensotec, Columbus, Ohio) were mounted to record relative bone-prosthesis motion in both the axial (longitudinal) and rotational (transverse) planes (Figure 1). Analog signals from LVDTs were continuously monitored and stored by means of a Labtech notebook operating system and an analog–digital converter sampling at 57 Hz.

Before the cyclic-loading of each specimen, an axial load of 200 N was applied in a ramp function to reduce the initial "seating" variability caused by implantation. The load applied for testing consisted of a sinusoidal load of 750 N at a frequency of 1 Hz with a synchronous torque of 11 N-m applied through the prosthetic femoral head by a polyethylene acetabular liner mounted to the testing machine. Each specimen was loaded for 200 cycles, as this was found to be sufficient to reach a steady state for bone–prosthesis micromotion.

Table. Age, Sex, Race, and Bone Mineral Density (BMD) of the Femora From Each Specimen Tested*

Age	Sex	Race	Fluted Stem BMD (g/cm^2)	Cylindrical Stem BMD (g/cm^2)
62	F	C	0.866	0.952
82	F	C	0.723	0.725
85	M	C	1.052	0.882
79	F	C	0.952	1.011
64	M	AA	0.869	0.846
93	F	C	0.727	0.695
79	F	C	0.803	0.820
81	F	C	0.988	0.960

*F, female; M, male; C, Caucasian; AA, African American.

Data Analysis

Data were obtained and recorded over the entire 200 cycles. As repeated loads on each specimen did not differ by more than 5%, the motion produced during the last 5 cycles was used for all determinations of motion at the bone–prosthesis interface.

Measurements of motion at the interface were compared between stems using the paired *t* test. The association between BMD and motion measurements was analyzed with the Pearson correlation coefficient. Data are presented as means and SDs.

Results

Of the 10 pairs of femora harvested, 8 completed the testing protocol. Two sets of femora were excluded because one pair fractured during stem implantation, and the other fractured at the bone-alumina pot interface during mechanical testing. Mean age of specimens that completed the entire testing protocol was 78.1 years (SD, 10.4 years; range, 62-93 years) (Table).

For the fluted stem, mean BMD was 0.87 g/cm^2 (SD, 0.12 g/cm^2; range, 0.72-1.05 g/cm^2), mean axial displacement was 13.33 µm (SD, 2.56 µm; range, 9.92-16.48 µm) (Figure 2), and mean rotation was 9.81 µm (SD, 2.32 µm (range, 6.01-12.58 µm) (Figure 3).

For the cylindrical stem, mean BMD was 0.86 g/cm^2 (SD, 0.11 g/cm^2; range, 0.70-1.01 g/cm^2), mean axial displacement was 18.37 µm (SD, 3.14 µm; range, 14.31-22.58 µm), and mean rotation was 13.40 µm (SD, 1.06 µm; range, 9.89-17.17 µm).

Paired-samples *t* test demonstrated no significant difference between the BMD of the femora tested for the 2 prostheses (P = .69). However, a statistically significant difference existed for the bone–prosthesis interface motion between the 2 prostheses. The fluted distal stem design demonstrated less axial displacement (P = .001) and less rotation (P = .043). There was no association between BMD for axial displacement (r = 0.031, P = .91) or rotational displacement (r = 0.05, P = .85) for either stem.

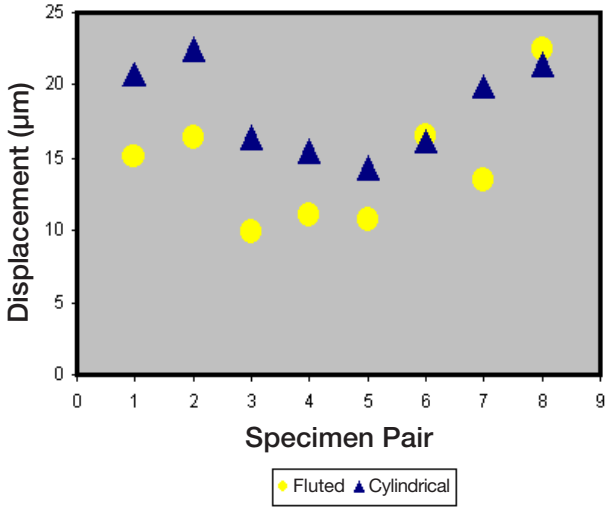

Figure 2. Axial displacement (µm) of prostheses.

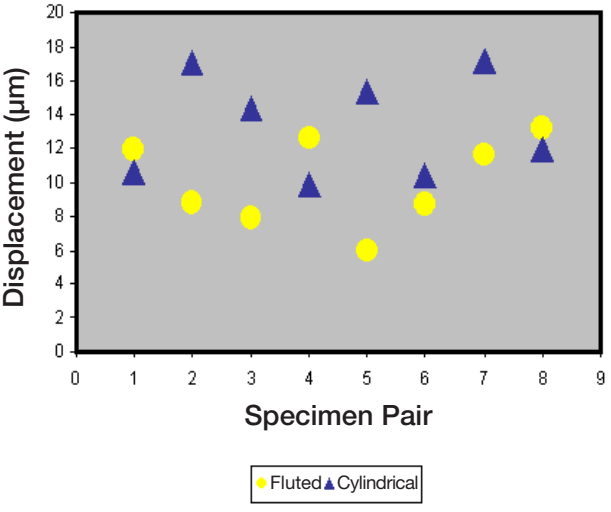

Figure 3. Rotational displacement (µm) of prostheses.

Discussion

Minimizing implant–bone micromotion is directly related to the initial stability of a femoral implant and may be related to its ability to function clinically over the long term. Moreover, there appears to be an important 2-way relationship between implant micromotion and bony ingrowth. Frequency and extent of bony ingrowth are reduced by excessive micromotion; conversely, the larger the amount of bony ingrowth, the more stable the implant.[7] Excessive micromotion of a cementless implant relative to bone is thought to be a major factor causing failure of fixation.[1,2,8,9] Minimizing the micromotion of an uncemented prosthetic component is a key requirement for obtaining bony ingrowth.[5] If the initial movement is excessive, bony ingrowth into the porous surface may not occur.

An appreciation of the range of initial implant-bone relative movement that would allow for bony ingrowth is important in the design and use of implants selected for both primary and revision THA. However, relatively few

data exist regarding the critical ranges of micromotion that would allow for stable bony ingrowth. In a cadaveric retrieval study, Engh and colleagues[7] demonstrated that the maximum relative axial motion between bone and implant that showed bony ingrowth was 40 µm, with fibrous ingrowth occurring with motions as high as 150 µm. However, implants stabilized with fibrous tissue ingrowth were distinctly less rigidly fixed to the femur than those that demonstrated evidence of bony ingrowth. In an animal study, Pilliar and colleagues[5] found similar results, suggesting that the amount of acceptable displacement between the interface of the implant and bone should be less than 28 µm for bony ingrowth and up to 150 µm for fibrous ingrowth.

Continued controversy exists regarding cementless hip prosthesis design parameters, including metallurgy, ingrowth surface, extent of porous coating, and distal stem geometry. Relatively little research has focused on the contribution of the distal stem design to implant stability, especially in scenarios in which the only adequate bone stock available for rigid initial fixation is diaphyseal. In the present study, we elected to evaluate 2 curved stem designs because of the ability of curved stems to resist large torsional loads and to match the anatomical anterior bow of the femoral diaphysis.[4] The 2 designs differed in metallurgy (titanium vs cobalt chromium), pore size (70 vs 250 microns), and distal stem design (tapered, fluted vs nontapered, cylindrical). Otani and colleagues[8] demonstrated no significant differences in micromotion with axial or torsional load tests when comparing stems with different elastic characteristics. The present in vitro study evaluated only initial stability, and therefore the effects of pore size on subsequent bony ingrowth were not evaluated. Although differences in prosthesis surface texture could conceivably affect surface friction and prosthesis purchase, we postulated that any differences in bone-implant micromotion would be predominantly caused by distal stem geometry.

In this study, we attempted to limit the contribution of proximal stem geometry and femoral bone to implant stability by resecting the femur 12 cm from the greater trochanter. We expected that this proximal bone would significantly limit motion and therefore would confound our measurements. Further, diaphyseal fixation is more critical in cases of inadequate remaining proximal bone stock, and our revision THA model attempted to simulate this. The loading parameters were selected on the basis of minimum expected loads placed on a femoral prosthesis in the early postoperative period.[10,11] A statistical difference was present in the prosthesis-bone interface motion in axial displacement ($P = .0001$) and rotation ($P = .043$) in a comparison between a tapered, fluted stem design and a cylindrical distal stem design. As noted, these differences in motion may be explained by the distal stem design. Although the fluted stem has less surface contact with bone, the flutes are able to cut into the endosteal surface during insertion, thereby increasing the gross mechanical interlock of the implant. Theoretically, one might expect the cylindrical stem to provide for stronger interlock with the bone on the basis of its larger surface contact area. However, one possible explanation for our findings is that the cylindrical stem erodes the reamed surface during insertion, thus reducing the effective interface between the endosteal bone and implant.

On a biomechanical basis, our results suggest that the fluted stem provides better initial fixation and potentially a superior environment for subsequent bony ingrowth, thereby improving long-term fixation over the cylindrical stem. These findings are in agreement with those of Kendrick and colleagues,[3] who demonstrated that a fluted distal stem design provided superior resistance to rotational stresses in a comparative study of primary THA stems. However, both stems demonstrated motion below the threshold for bony ingrowth (ie, <28-40 µm).[5,7] Therefore, it would appear that either prosthesis is capable of providing adequate initial fixation.

However, initial biomechanical stability is only one factor in the ultimate clinical outcome of a femoral implant. Although the cylindrical stem demonstrated more micromotion when compared with the fluted stem, clinical results of the cylindrical prosthesis have been excellent, which was one reason that we included this stem in our testing protocol.[12-15] The tapered, fluted stem was designed to provide for more rotational stability. Our results support this design rationale, at least with regard to initial fixation. However, though early results have been promising,[16] no long-term clinical results have been documented regarding the success of the implant.

This study has several limitations. First, as in any in vitro study, loads applied do not reproduce actual in vivo forces; therefore, relative bone-implant motion is not an exact replica of a clinical scenario. Second, loads applied through the femoral head were of relatively small magnitude and duration. Larger differences in micromotion may have been measured if we had used higher loads with varying angles of application. Third, we did not include muscle simulation in the loading jig. It is likely that different amounts of micromotion may have been obtained with the inclusion of muscular forces. Last, our model did not include endosteal and circumferential bone loss, either or both of which may be present after femoral component failure. We felt that any technique that artificially created such defects would not likely be uniform and might have introduced unwanted bias. These limitations notwithstanding, we believe our results are valid and reproducible and provide valuable biomechanical information.

Conclusions

Our study demonstrated that, at least for the prosthesis designs tested, a tapered, fluted stem provides superior initial biomechanical stability when compared with a cylindrical design. Both stems provide adequate initial stability to theoretically support subsequent bony ingrowth. These findings should be considered when selecting revision hip implants for cases in which stress shielding or loosening has caused proximal bone loss and when developing new

implant designs. As the long-term clinical results of the cylindrical prosthesis we evaluated have been excellent, our results support the need for further clinical studies to assess the clinical outcomes of the fluted, tapered distal stem design.

Authors' Disclosure Statement and Acknowledgments

The authors report no actual or potential conflict of interest in relation to this article.

The opinions or assertions contained herein are the private views of the authors and are not to be construed as official or as reflecting the views of the US Army or the US Department of Defense. Some of the authors are employees of the US government. This work was prepared as part of their official duties, and, as such, there is no copyright to be transferred.

We thank Oleg Vesnosky, PhD, LD, Timmie Topoleski, PhD, and Robin Howard, MS, for their assistance with this study.

References

1. Buhler DW, Berlemann U, Lippuner K, Jaeger P, Nolte LP. Three-dimensional primary stability of cementless femoral stems. *Clin Biol.* 1997;12(2):75-86.
2. Burke DW, O'Conner DO, Zalenski EB, Jasty M, Harris WH. Micromotion of cemented and uncemented femoral components. *J Bone Joint Surg Br.* 1991;73(1):33-37.
3. Kendrick JB, Noble PC, Tullos HS. Distal stem design and the torsional stability of cementless femoral stems. *J Arthroplasty.* 1995;10(4);463-469.
4. Engh CA, Bobyn JD, Glassman AH. Porous coated hip replacement the factors governing bone ingrowth, stress shielding and clinical results. *J Bone Joint Surg Br.* 1987;69(1):45-55.
5. Pilliar RM, Lee JM, Maniatopoulos C. Observations on the effect of movement on bone ingrowth into porous-surfaced implants. *Clin Orthop.* 1986 Jul;208:108-113.
6. Schneider E, Kinast C, Eullenberger J, Wyder D, Eskillsson G, Perren SM. A comparative study of the initial stability of cementless hip prostheses. *Clin Orthop.* 1989 Nov;248:200-209.
7. Engh CA, O'Commer D, Jasty M, McGovern TF, Bobyn JD, Harris WH. Quantification of implant micromotion, strain shielding, and bone resorption with porous-coated anatomic medullary locking femoral prostheses. *Clin Orthop.* 1992 Dec;285:13-29.
8. Otani T, Whiteside LA, White SE, McCarthy DS. Effects of femoral component material properties on cementless fixation in total hip arthroplasty. *J Arthroplasty.* 1993;8(1):67-74.
9. Callaghan JJ, Fulghum CS, Glisson RR, Stranne SK. The effect of femoral stem geometry on interface motion in uncemented porous-coated total hip prosthesis. Comparison of straight stem and curved stem design. *J Bone Joint Surg Am.* 1992;74(6):839-848.
10. Crowninshield RD, Johnston RC, Andrews JG, Brand RA. A biomechanical investigation of the human hip. *J Biomech.* 1978;11(1-2):75-85.
11. Davy DT, Kotzar GM, Brown RH, et al. Telemetric force measurements across the hip after total arthroplasty. *J Bone Joint Surg Am.* 1988;70(1):45-50.
12. Engh CA, Glassman AH, Griffin WL, Mayer GD. Results of cementless revision for failed cemented total hip arthroplasty. *Clin Orthop.* 1988 Oct;235:91-110.
13. Krishnamurthy AB, MacDonald SJ, Praposky WG. 5- to 13-year follow-up study on cementless femoral components in revision surgery. *J Arthroplasty.* 1997;12(8):839-847.
14. Moreland JR, Bernstein ML. Femoral revision hip arthroplasty with uncemented, porous coated stems. *Clin Orthop.* 1995 Oct;319:141-150.
15. Paprosky WG, Greidanus NV, Antoniou J. Minimum 10-year results of extensively porous-coated stems in revision hip arthroplasty. *Clin Orthop.* 1999 Dec;369:230-242.
16. Kwong LM, Miller AJ, Lubinus P. A modular fixation option for proximal bone loss in revision total hip arthroplasty: a 2- to 6-year follow-up study. *J Arthroplasty.* 2003;18(3 suppl 1):94-97.

This paper will be judged for the Resident Writer's Award.