The Journal of Arthroplasty Vol. 28 No. 1 2013

Biomechanical Analysis of Acetabular Revision Constructs

Is Pelvic Discontinuity Best Treated With Bicolumnar or Traditional Unicolumnar Fixation?

Jeremy M. Gililland, MD, Lucas A. Anderson, MD, Heath B. Henninger, PhD, Erik N. Kubiak, MD, and Christopher L. Peters, MD

Abstract: Pelvic discontinuity in revision total hip arthroplasty presents problems with component fixation and union. A construct was proposed based on bicolumnar fixation for transverse acetabular fractures. Each of 3 reconstructions was performed on 6 composite hemipelvises: (1) a cup-cage construct, (2) a posterior column plate construct, and (3) a bicolumnar construct (no. 2 plus an antegrade 4.5-mm anterior column screw). Bone-cup interface motions were measured, whereas cyclical loads were applied in both walking and descending stair simulations. The bicolumnar construct provided the most stable construct. Descending stair mode yielded more significant differences between constructs. The bicolumnar construct provided improved component stability. Placing an antegrade anterior column screw through a posterior approach is a novel method of providing anterior column support in this setting. **Keywords:** hip arthroplasty, revision hip arthroplasty, acetabular revision, discontinuity.

With total hip arthroplasty increasing in number annually and being performed in younger patients, the challenge of adequately addressing severe pericomponent bone loss in repeat revisions will become more frequent [1]. The most difficult of these bony deficits is pelvic discontinuity in the setting of acetabular bone loss. Pelvic discontinuity is a severe form of acetabular deficiency defined as a complete separation of the superior and inferior hemipelvis. In published series, the rate of discontinuity encountered in revision arthroplasty ranged from 1% to 8% of all acetabular revisions preformed [1-7]. With annual primary total hip arthroplasties expected to exceed 550 000 and revisions to be almost 100 000 by the year 2030, the incidence of acetabular revision in the setting of pelvic discontinuity will become more common [1]. To date, most treatment options have been anecdotal; the results of which have only been presented in a limited number of relatively small case series [1-8]. Historically, pelvic discontinuity was bone grafted with bulk allograft and then stabilized with a revision construct [9,10]. However, high failure rates of autograft and allograft treatment of pelvic discontinuity have driven current revision strategies [3,5,8,11]. In lieu of large structural bone grafts, ingrowth materials such as tantalum and trabecular metal have been used to span the discontinuity and provide internal fixation to the superior and inferior hemipelvis fragments. Stability is essential as an ingrowth component can only be successful if it is adequately stabilized against the host bone to allow host bone ingrowth to occur. Excessive motion at the bone implant interface may jeopardize the ability to achieve bony ingrowth into the revision component, leading to eventual failure of the reconstruction.

Mechanical constructs may provide the important initial stability in the setting of pelvic discontinuity, and they range from a simple revision acetabular component supported by a posterior column plate to a more complex custom fabricated triflange acetabular component. More complex hardware combinations necessitate larger exposures, which are often associated with higher rates of infection and overall complications [4,11]. Therefore, study of the initial stability provided by various constructs is a vital undertaking in the effort

From the Department of Orthopaedic Surgery, University of Utah, Salt Lake City, Utah.
Submitted December 20, 2011; accepted April 23, 2012.
The Conflict of Interest statement associated with this article can be found at http://dx.doi.org/10.1016/j.arth.2012.04.031.
Reprint requests: Jeremy M. Gililland, MD, Department of Orthopaedic Surgery, University of Utah School of Medicine, 590 Wakara Way, Salt Lake City, UT 84108.

0883-5403/2801-0028$36.00/0
http://dx.doi.org/10.1016/j.arth.2012.04.031

to maximize union rates and minimize complications in these difficult cases.

There are currently no biomechanical evaluations in the literature comparing acetabular reconstruction constructs for the treatment of pelvic discontinuity. Therefore, this study evaluated the biomechanics of 3 types of constructs:(1) the established concept of a cup supported by a posterior column plate, (2) a more recently adopted cup-cage construct with a cup supported by a reconstruction cage, and (3) the novel concept of a cup supported by a posterior column plate and an antegrade anterior column screw. The cup-cage construct involved the use of a large ingrowth metal acetabular cup as a void filler stabilized with an overlying reconstruction cage with a cemented polyethylene liner [2,12,13]. Construct 3 was developed based on the concept of bicolumnar fixation for transverse acetabular fractures to provide increased initial stability without the need for extensile exposures that have been used in prior dual plating constructs [4,11].

This study had 2 primary goals. The first was to evaluate the mechanical stability of the 3 acetabular revision constructs by quantifying the displacement of the superior and inferior hemipelvis fragments with respect to the acetabular component. The second was to quantify the stiffness of each construct, based on the slope of the load-to-failure curve.

Materials and Methods

A pelvic discontinuity model was created in custom artificial osteoporotic composite hemipelvises (Sawbones, model no. 3405-10pcf; Pacific Research Laboratories, Inc, Vashon, Wash). A custom jig was used to create a standardized transverse osteotomy through each acetabulum to simulate pelvic discontinuity. Large acetabular reamers and a 1-in hole saw were used to remove most of the acetabular and periacetabular bone stock to simulate periprosthetic bone loss. A saw was then used to create a standardized transverse gap defect simulating a pelvic discontinuity (Fig. 1). By using this jig, the amount and position of bone removal as well as the level of the transverse osteotomy were kept constant between all specimens in the study.

Three acetabular revision constructs were assembled in 6 different models for each construct (n = 6 models per construct; total, 18 models): (1) 60-mm Regenerex Ringloc Acetabular Component (Biomet, Warsaw, Ind) with an 8-hole, 3.5 × 94 mm posterior column reconstruction plate (Synthes USA, Paoli, Pa), (2) 45-mm Protrusio Cage (DePuy, Warsaw, Ind) and a cemented polyethylene liner over a 60-mm Regenerex Ringloc Acetabular Component, and (3) 60-mm Regenerex Ringloc Acetabular Component with an 8-hole, 3.5 × 94 mm posterior column reconstruction plate (Synthes USA) and a 4.5 × 120 mm anterior column screw (Synthes USA). When building the constructs, uniformity was critical to provide valid comparisons. Therefore, all screws were placed in realistic intraoperative trajectories, and jigs were used to drill all screw holes. Each construct used the same 2 acetabular screws placed into the posterosuperior quadrant in realistic intraoperative trajectories (Fig. 2). No additional acetabular screws were placed because minimizing direct acetabular component stabilization amplified differences between the indirect component stabilization provided by the surrounding revision constructs.

As mentioned previously, construct 3 was developed, based on the concept of bicolumnar fixation for transverse acetabular fractures, to provide increased initial stability without the need for extensile exposures that have been used in prior dual plating constructs. The anterior column fixation in this construct is provided by a 4.5-mm anterior column screw placed in antegrade rather than the traditional retrograde fashion. The surgical technique for placement of this screw uses a standard posterior approach for addressing pelvic discontinuities. First, dissect over the superolateral acetabular rim to visualize the starting point of the screw, which is just lateral to the anterior inferior iliac spine (Fig. 3). A long 3.5-mm drill is placed percutaneously through the gluteal musculature using visual triangulation and fluoroscopic assistance to guide the trajectory of the drill. The drill is then brought down to the starting point where the tip of the drill can be visualized through the posterior approach, anterior, and superior to the acetabulum. Once the appropriate starting point is

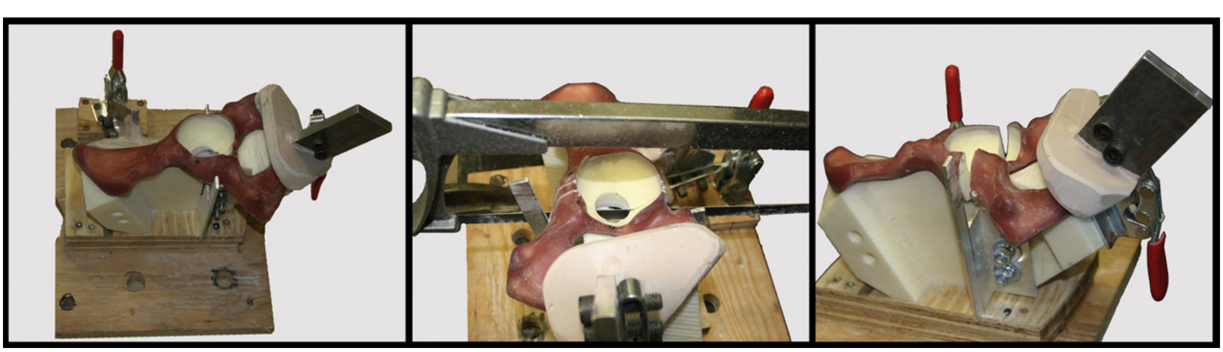

Fig. 1. Custom jig used to create a reproducible pelvic discontinuity in the hemipelvis model.

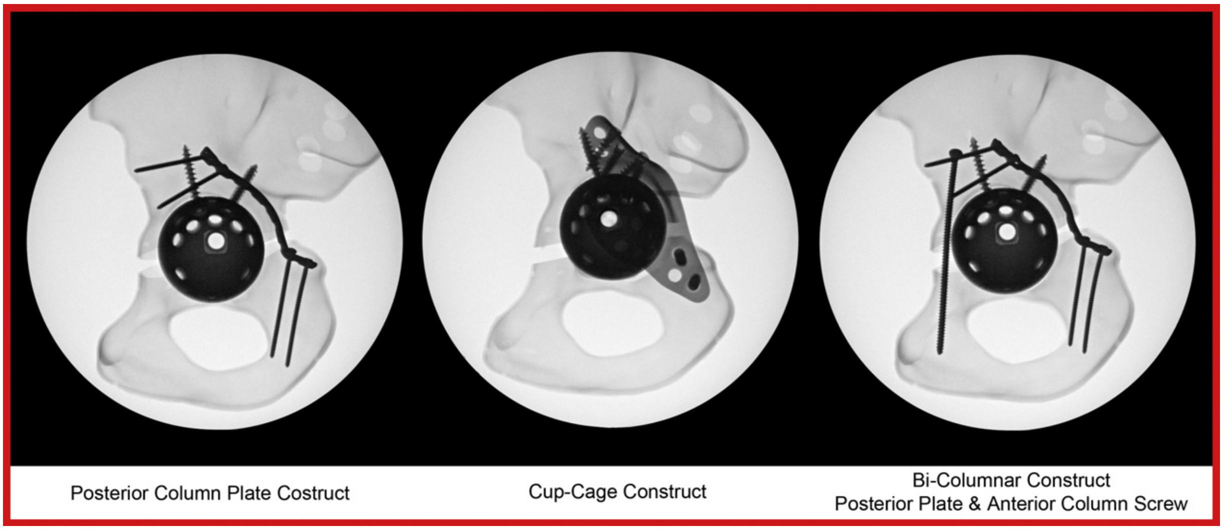

Fig. 2. Fluoroscopic images of the 3 revision constructs tested: (1) posterior column plate construct, (2) cup-cage construct, (3) bicolumnar construct: posterior column plate and antegrade anterior column screw.

verified under direct visualization, the drill is then advanced down the anterior column using a rolled-over (iliac) inlet fluoroscopic view to verify that the screw is not going out superiorly and a rolled-over (obturator) outlet view to verify that the screw is not going out medially through the inner table. A 4.5-mm non–self-tapping fully threaded screw is then placed down the anterior column using the same steps described for drill placement. This technique has been used safely and effectively at our institution.

The stabilized models were then mounted in a custom jig that attached the hemipelvis through the sacroiliac and pubic symphyseal joints via semideformable spacers to allow for slightly flexible interfaces. This configuration simulates a pelvic ring out of the hemipelvis model (Fig. 4). Of note, the mounting jig was tuned using an intact hemipelvis model by varying the stiffness of the material at the pubic symphyseal interface until published normal physiologic motion at the pubic symphysis was approximated under normal loads of walking [14,15].

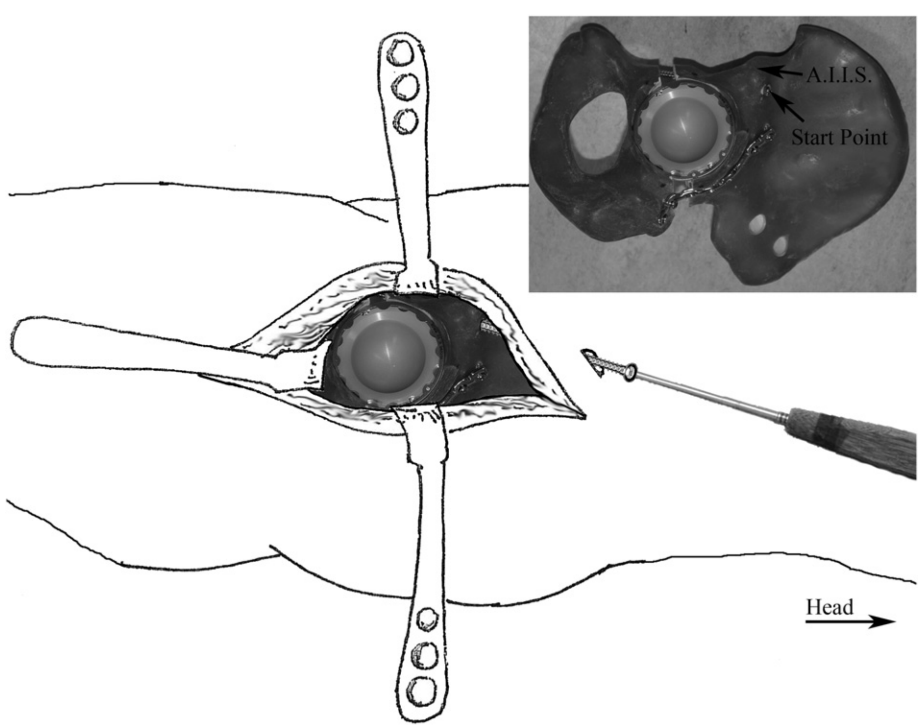

Fig. 3. Drawing of posterior approach showing placement of percutaneous antegrade anterior column screw. Inset image showing appropriate starting point for the antegrade anterior column screw on a Sawbone model.

Bergmann et al [16] reported hip joint reaction force vectors in terms of flexion, abduction, and rotation and the magnitudes of these joint reaction forces during daily activities for 4 patients with instrumented femoral prostheses. The mean data from that study were used in the present study to simulate walking and descending stairs positions. Two custom wedges positioned our pelvic jig such that that a vertical load delivered by a femoral component ball, attached to the piston of the 5-kN tension-compression load cell (Model 2518-103; Instron Corp, Norwood, Mass), was directed into the acetabulum approximating the kinematics of peak loads seen during normal walking and descending stairs. For walking mode, the joint reaction force vector was placed in 6.5° of flexion and 7.2° of abduction. According to Bergmann et al, this approximates the direction of the peak hip joint reaction force during the heel strike phase of normal walking gait. For descending stairs, the joint reaction force vector was placed in 9.2° of extension and 9.5° of abduction. These loading conditions have been used successfully in a previous biomechanical analysis of the hip joint [17].

The descending stairs loading condition produced the highest hip joint reaction forces in Bergmann et al [16]. Stair ascent has been used previously to evaluate femoral stem fixation due to large torsional moments [16]. However, because the present study evaluated the acetabular component fixation, peak joint reaction forces were used because the acetabular component is somewhat shielded from the torsional moments by the low-friction bearing surface separating the 2 components. The stair descent condition was also used to detect differences due to the joint reaction force vector, which is significantly different than the walking condition. The larger anteriorly directed peak loads, as seen in stair descent, would provide more information regarding construct stability than stair ascent, which was similar to the walking condition.

Fig. 4. Photograph of custom Instron mounting jig attaching the hemipelvis through the sacroiliac and pubic-symphyseal joints via semideformable spacers.

Infrared diode arrays (flags) for the Optotrak motion capture system (Model 3020; Northern Digital, Waterloo, Ontario, Canada) were used to track the motion of the anterosuperior, anteroinferior, posterosuperior, and posteroinferior portions of the hemipelvis fragments. These flags were placed near the bone-implant interface, and digitized points were recorded directly on the bone-implant interface and referenced to the closest corresponding flag (Fig. 4). A flag was also attached to the posterior aspect of the acetabular component and was used to track the motion of the acetabular component with respect to the 4 digitized points at the bone-implant interface.

The loading protocol included 10 cycles per specimen of a sinusoidal loading from a baseline load of 50 N to a peak load of 1900 N at a rate of 0.5 Hz. The peak load (1900 N) was the average peak hip joint load seen in stair descent for a 75-kg patient (261% body weight) [16]. This loading was used for stair descent and for walking mode to simplify the experimental setup and because the peak loads seen with walking were only slightly less (233%-251% body weight depending on walking speed [16]). A slightly larger load in the walking condition overestimates the micromotion in walking but provides for direct interpretation of the differences between walking and stair descent as a function of only joint reaction force direction. Preconditioning was performed by running each specimen through 25 cycles of sinusoidal loading, then releasing the load and allowing the specimen to settle for 1 minute. An additional 25 cycles were run, and data were collected from the last 10 of these cycles. Pilot testing determined that this protocol decreased the variability from the constructs settling. The same protocol was used to test each specimen in both walking mode and descending stairs mode, and the testing order (walking or descending) was randomized between specimens.

Once cyclic loading was complete, a load to failure cycle was performed, while the load-displacement curve from the Instron was recorded. Two of each construct (3 constructs × 2 = 6 models) was tested to failure in normal walking mode with the Instron set on displacement control at a rate of 0.5 cm/s. *Failure* was defined as visible or audible fracture of any portion of the hemipelvis model. Only 2 constructs were used for each failure because hardware deformation occurred during these tests and hardware quantity limited additional testing.

A custom MATLAB (MathWorks, Inc, Natick, Mass) script was used to transform the Cartesian Optotrak data to a spherical coordinate system centered on the flat surface of the acetabular component (Fig. 5). By subtracting the motions of the cup at each digitized point from the corresponding point on the bony segment, motion at the cup-bone interface was quantified. Component motion data were defined in terms of compression or distraction from the center of the cup (radial displacement), horizontal shear (rotation about the circumference of the flat surface of the cup), and vertical shear (rotation normal to the flat surface of the cup). Average displacement was calculated from the 10 cycles at both the baseline load and the peak load.

The 3 constructs were compared with the continuous variable of bone-implant interface motion as the primary outcome measure and the continuous variable of stiffness as the secondary outcome measure. A paired *t* test statistical comparison was used to detect differences between constructs within a specimen. The *P* values were adjusted for 3 multiple comparisons using Hommel's multiple comparison procedure [18]. Because the groups are compared to answer 2 separate questions, the initial stability and stiffness, each was considered its own "family of comparisons." Therefore, adjustments for 3 comparisons were done separately for both outcome measures, consistent with the family-wise error rate controlled for with the Hommel's multiple comparison procedure. Hommel's procedure controls for the type I error without the need for first performing an analysis of variance [19].

A priori power analyses and sample size determination were not performed because comparable biomechanical data have not been previously published for estimated effect sizes. However, after 6 specimens per construct, the primary measure of displacement met a posteriori statistical power of 0.8. Alternatively, this study was not adequately powered for the secondary outcome of construct stiffness, as only 2 specimens from each construct were available. Given that localized displacement is more physiologically relevant to healing than overall construct stiffness, this limitation was deemed admissible for descriptive purposes.

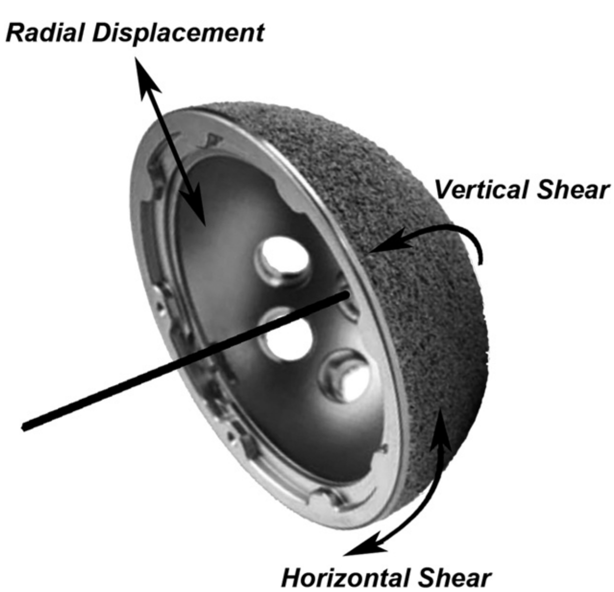

Fig. 5. Transformed spherical coordinate system centered on the flat surface of the acetabular component. All cup-bone interface motions are described in terms of these 3 component motions.

Results

At the anterosuperior bone-cup interface, in both walking and descending stair modes, the bicolumnar construct limited overall motion more than the posterior plate construct (stairs, 252 vs 640 μm; P = .003; walking, 242 vs 425 μm; P = .045) and the cup-cage construct (stairs, 252 vs 680 μm; P = .002; walking, 242 vs 1378 μm; P = .03). When analyzing data obtained with stair mode and focusing on individual component motions, the bicolumnar construct yielded less vertical shear than the cup-cage construct (108 vs 410 μm; P = .017) and less horizontal shear than the posterior plate construct (197 vs 442 μm; P = .001). Upon analysis of walking mode, both the bicolumnar construct (79 vs 427 μm; P = .009) and the posterior plate construct (66 vs 427 μm; P = .018) yielded significantly less radial displacement than the cup-cage construct (Fig. 6A and B).

At the anteroinferior bone-cup interface, the bicolumnar construct yielded less overall motion than the cup-cage construct in descending stair mode (369 vs 856 μm; P = .027). No significant differences were detected between constructs at this point in walking mode. No significant differences were seen in stair mode when breaking out the separate component motions (Fig. 6C and D).

At the posterosuperior bone-cup interface, no differences in overall motion were detected in walking or descending stair mode. No component motion differences were detected between constructs at this point when testing in walking mode. When breaking out the component motions seen in stair mode, there was significantly less vertical shear in the bicolumnar construct as compared with the posterior plate construct (199 vs 704 μm; P = .023) (Fig. 6E and F).

At the posteroinferior bone-cup interface, in stair mode, the bicolumnar construct limited overall motion more than the both the posterior plate construct (252 vs 402 μm; P = .009) and the cup-cage construct (252 vs 784 μm; P = .011). When breaking out the component motions in stair mode, the bicolumnar construct yielded less vertical shear than both the posterior plate construct (86 vs 272 μm; P = .023) and the cup-cage construct (86 vs 554 μm; P = .033). In walking mode, the posterior

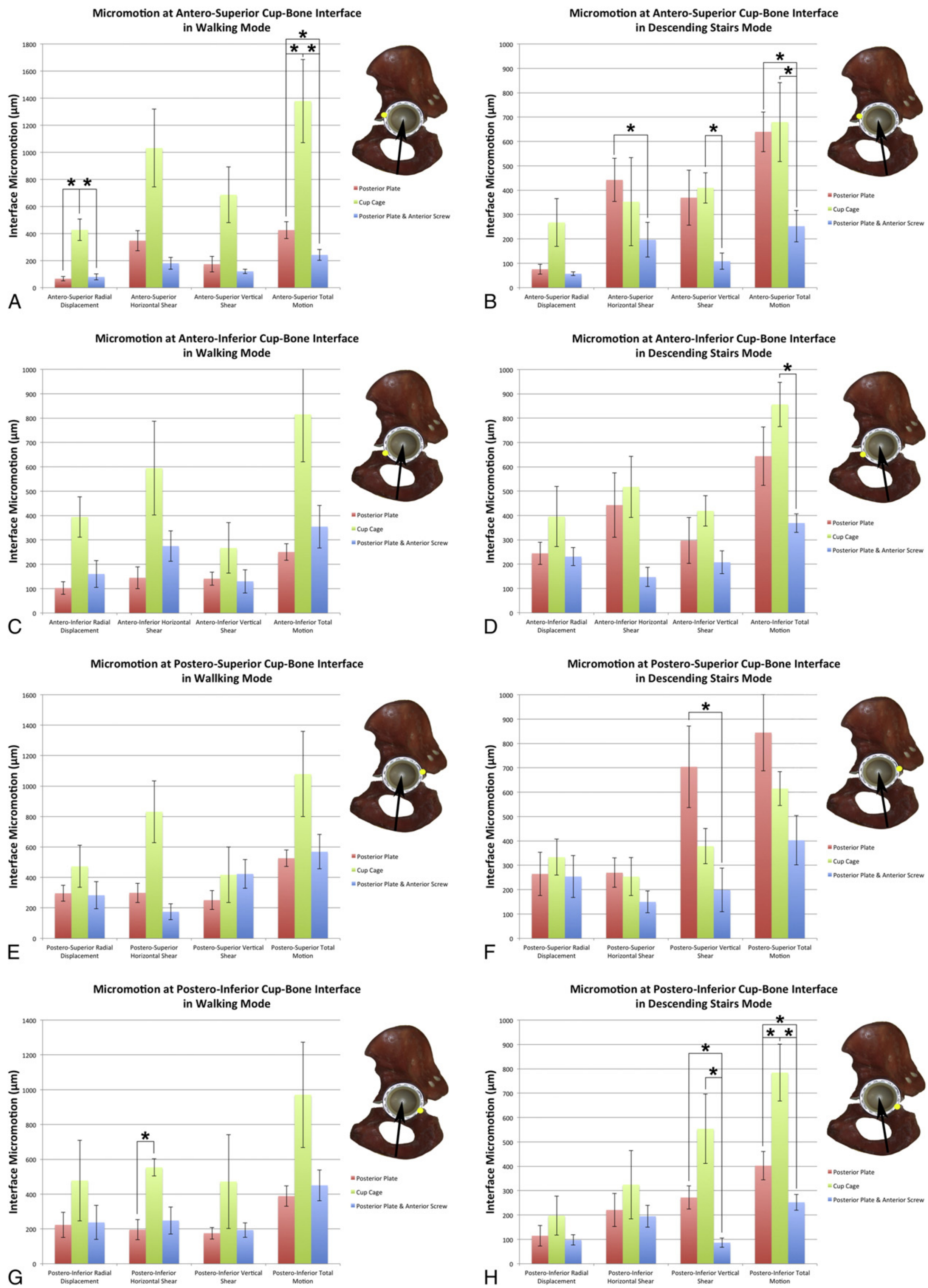

Fig. 6. Graphical representation of the cup-bone interface motion at each point for each construct both in walking and descending stair mode. Error bars indicates SEM. *$P \leq .05$.

plate construct yielded less horizontal shear than the cup-cage construct (196 vs 553 μm; P = .036) (Fig. 6G and H).

There was no statistically significant difference in the stiffness of the constructs from the load to failure data. In addition, all of the specimens failed by a fracture propagating through the sacroiliac mounting holes. Despite this common failure mode, slight, though statistically insignificant, differences were noted in the slopes of the force displacement curves before construct failure. Although statistically insignificant, there was a trend toward the bicolumnar construct being the stiffest (mean stiffness, 769 N/mm; SEM, 7) followed by the cup-cage construct (mean stiffness, 687 N/mm; SEM, 61), and the posterior plate construct (mean stiffness, 607 N/mm; SEM, 30).

Discussion

One of the most challenging problems in revision arthroplasty of the hip is acetabular bone loss. Pelvic discontinuity is among the most severe of acetabular defects and is defined as a complete separation of the superior and inferior hemipelvis. Historically, pelvic discontinuity was bone grafted with bulk allograft and then stabilized with a revision construct [9,10]. Unfortunately, the use of large allografts has shown poor results, even when supported by constructs such as a reconstruction cage. In several series with short-term to midterm follow-up of pelvic discontinuity treated with structural allografts, complication rates have been high, with loosening occurring in 15% to 28% [3,5,7,11]. In another series with good results at 5 to 10 years, late failure was seen in 60% of large allografts and 30% of large autografts at 16.5-year follow-up [20].

Failures of autograft and allograft treatment of pelvic discontinuity have driven current revision strategies, and ingrowth materials such as tantalum and trabecular metal are being used to span the discontinuity and provide internal fixation to the superior and inferior hemipelvis fragments. Clinical series using ingrowth materials are promising, with lower rates of loosening (0%-8%) at early to midterm follow-up [1,2,4,6]. Stability is crucial in these constructs, as excessive motion at the bone implant interface may lead to eventual failure of the reconstruction. More complex hardware combinations necessitate larger exposures and are often associated with higher rates of infection and overall complications [4,11]. Therefore, knowledge of the initial stability provided by various constructs is essential to maximize union rates and minimize complications in these difficult cases.

Overall, when comparing the 3 constructs evaluated in this study, the bicolumnar construct was the most stable in terms of limiting micromotion at the interface between the superior and inferior bone segments and the acetabular component. In general, there were not significant differences between the posterior plate construct and the cup-cage construct. Compared with walking mode, descending stair mode yielded more significant differences between constructs. This is secondary to the anterior directed joint reaction force in descending stair mode, which amplifies the inherent difference between the bicolumnar construct and the other 2 constructs; the bicolumnar construct stabilizes the anterior column, whereas the other 2 constructs do not.

Animal models have shown that bone-implant attachment occurs via fibrous connective tissue rather than bone when interface motions exceed 150 μm [21,22]. The magnitude of bone-implant interface motion in the present study exceeds the ideal limit for bone ingrowth to occur. Although the bicolumnar construct generally had the least amount of bone-implant interface motions, most of these motions still exceeded 150 μm. This implies that none of the evaluated constructs provides adequate initial stability to allow immediate full weight bearing in walking or descending stairs conditions. In our institution, it is protocol to make these patients touchdown weight bearing on the operative extremity for the first 6 weeks postoperative, and the data from this study support this practice.

The cup-cage construct yielded more interface motion than intuitively expected. When creating the constructs, uniformity between common portions of the 3 different constructs was critical to provide valid comparisons. Therefore, the same 2 posterosuperior acetabular screws were placed through the same holes in the acetabular component in each construct. No additional acetabular screws were used because minimizing direct acetabular component stabilization amplified differences between the indirect component stabilization provided by the surrounding revision constructs. This attempt at uniformity and decreased direct acetabular component stability led to the lack of anterosuperior iliac acetabular fixation and anteroinferior pubic fixation in the cup-cage construct, which may have led to hinging through the malleable posterior based cage. Current cup-cage systems allow for some modularity in that acetabular screws can be placed wherever needed by burring through the porous metal cup. This enables an increase in anterior column fixation in comparison with previous revision acetabular components with defined screw hole patterns, the importance of which seems to be evident in the present data.

Several limitations should be noted. First, this is a highly simplified model of a very complex and variable problem. The use of uniform custom osteoporotic Sawbone models and a standard discontinuity defect provided a valid platform with which to compare various constructs alone. Cadavers may provide a more in vivo simulation due to bony morphology and interface, but variability between specimens may have limited the ability to differentiate construct stability from implantation effects. In addition, previous studies have shown Sawbone models to be effective for investigating the stability of fracture fixation constructs in both the femur and pelvis [23-26].

Second, constructs were tested by quasistatic loading, which is a gross simplification of real hip joint kinematics. However, loading conditions tested in this study were based on in vivo data from instrumented femoral prostheses, and this testing protocol has been validated by previous biomechanical studies [16,17]. Third, the construct stiffness data do not have adequate statistical power to draw definite conclusions but do provide direction for future analyses with a higher number of samples. Finally, the Optotrak motion capture system has a listed accuracy of 0.1 mm. Our interface motions were found to be less than 1 mm and occasionally approached the accuracy threshold of 100 μm. However, most motion data points (75%) recorded were above this threshold, and therefore, the reliability of the Optotrak is acceptable (mean motion overall, 300 μm; range, 15-1811 μm).

By definition, both the anterior and posterior columns are deficient in the setting of pelvic discontinuity. The present data indicate that the reconstruction/stabilization of both columns is essential to limit motion at the bone-implant interface. The bicolumnar construct improved component stability in the presence of pelvic discontinuity compared with the posterior column plate and cup-cage constructs. Placing an antegrade anterior column screw through a posterior approach is a novel method of providing anterior column support in this difficult setting. Isolated posterior column fixation, regardless of whether it is through a simple posterior column plate or a more elaborate cup-cage construct, does not address the anterior column defect, which can lead to the acetabular component hinging off of the posterior fixation, excessive interface motion, and potential ingrowth failure. However, anterior column fixation through acetabular screws and/or an antegrade anterior column screw may be viable adjuncts to either construct. Future biomechanical studies are needed to further analyze constructs that address the defects in both columns, including cup-cage constructs coupled with anterior acetabular screws and anterior column screws.

Acknowledgments

The authors thank the Orthopaedic Research and Education Foundation for their support of this project. The authors thank Biomet, Synthes USA, and DePuy for providing the instrumentation and implants used in this study. We would like to thank Greg Stoddard for his assistance with our statistical analysis.

References

1. Kurtz S, Ong K, Lau E, et al. Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. J Bone Joint Surg Am 2007;89:780.
2. Paprosky W, Sporer SO, Rourke MR. The treatment of pelvic discontinuity with acetabular cages. Clin Orthop Relat Res 2006;453:183.
3. Weeden SH, Schmidt RH. The use of tantalum porous metal implants for Paprosky 3A and 3B defects. J Arthroplasty 2007;22(6 Suppl 2):151.
4. Berry DJ, Lewallen DG, Hanssen AD, et al. Pelvic discontinuity in revision total hip arthroplasty. J Bone Joint Surg Am 1999;81:1692.
5. Deboer DK, Christie MJ, Brinson MF, et al. Revision total hip arthroplasty for pelvic discontinuity. J Bone Joint Surg 2007;89:835.
6. Sporer SM, Paprosky WG. Acetabular revision using a trabecular metal acetabular component for severe acetabular bone loss associated with a pelvic discontinuity. J Arthroplasty 2006;21(6 Suppl 2):87.
7. Stiehl JB, Saluja R, Diener T. Reconstruction of major column defects and pelvic discontinuity in revision total hip arthroplasty. J Arthroplasty 2000;15:849.
8. Eggli S, Muller C, Ganz R. Revision surgery in pelvic discontinuity: an analysis of seven patients. Clin Orthop 2002;136.
9. D'Antonio JA, Capello WN, Borden LS, et al. Classification and management of acetabular abnormalities in total hip arthroplasty. Clin Orthop Relat Res 1989;126.
10. Paprosky W, Magnus RE. Principles of bone grafting in revision total hip arthroplasty. Acetabular technique. Clin Orthop Relat Res 1994;147.
11. Christie MJ, Barrington SA, Brinson MF, et al. Bridging massive acetabular defects with the triflange cup: 2- to 9-year results. Clin Orthop Relat Res 2001;216.
12. Gross AE, Goodman S. The current role of structural grafts and cages in revision arthroplasty of the hip. Clin Orthop Relat Res 2004;193.
13. Gross AE, Goodman SB. Rebuilding the skeleton: the intraoperative use of trabecular metal in revision total hip arthroplasty. J Arthroplasty 2005;20(4 Suppl 2):91.
14. Garras DN, Carothers JT, Olson SA. Single-leg-stance (flamingo) radiographs to assess pelvic instability: how much motion is normal? J Bone Joint Surg Am 2008;90:2114.
15. Walheim G, Olerud S, Ribbe T. Mobility of the pubic symphysis. Measurements by an electromechanical method. Acta Orthop Scand 1984;55:203.
16. Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. J Biomech 2001;34:859.
17. Anderson AE, Ellis BJ, Maas SA, et al. Validation of finite element predictions of cartilage contact pressure in the human hip joint. J Biomech Eng 2008;130:051008.
18. Wright S. Adjusted P-values for simultaneous inference. Biometrics 1992;48:1005.
19. Dunnet C. When and how to do multiple comparisons. New York: Hall; 2006.
20. Shinar AA, Harris WH. Bulk structural autogenous grafts and allografts for reconstruction of the acetabulum in total

hip arthroplasty. Sixteen-year-average follow-up. J Bone Joint Surg Am 1997;79:159.
21. Jasty M, Bragdon C, Burke DO, et al. In vivo skeletal responses to porous-surfaced implants subjected to small induced motions. J Bone Joint Surg Am 1997;79:707.
22. Pilliar RM, Lee JM, Maniatopoulos C. Observations on the effect of movement on bone ingrowth into porous-surfaced implants. Clin Orthop Relat Res 1986;108.
23. Adamczyk MJ, Odell T, Oka R, et al. Biomechanical stability of bioabsorbable screws for fixation of acetabular osteotomies. J Pediatr Orthop 2007;27:314.
24. Fujishiro T, Nishikawa T, Niikura T, et al. Impaction bone grafting with hydroxyapatite: increased femoral component stability in experiments using Sawbones. Acta Orthop 2005;76:550.
25. Harman MK, Toni A, Cristofolini L, et al. Initial stability of uncemented hip stems: an in-vitro protocol to measure torsional interface motion. Med Eng Physics 1995;17:163.
26. Widmer BJ, Peters CL, Bachus KN, et al. Initial stability of the acetabular fragment after periacetabular osteotomy: a biomechanical study. J Pediatr Orthop 2010;30:443.