Acetabular Component Deformation with Press-Fit Fixation

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Abstract: Acetabular component deformation secondary to forces encountered during insertion is a potential consequence of the press-fit technique. This study characterized the stiffness of Pinnacle 100 cups (DePuy, Warsaw, Ind) via mechanical testing and used this information with intraoperative measurements of cup deformation to calculate the in vivo forces acting on cups inserted during hip arthroplasty in 21 patients. We found that 90.5% of cups had measurable compression deformity, averaging 0.16 ± 0.16 mm. The corresponding forces acting on these cups averaged 414 ± 421 N. For hard-on-hard bearing surfaces, such in vivo deformation of acetabular shells may result in negative clinical consequences such as equatorial loading with increased wear and potential seizing of components, chipping of ceramic inserts, or locking mechanism damage. **Key words:** hip arthroplasty, acetabular component, deformation, press-fit fixation.

Press-fit fixation is currently a common method for implanting noncemented acetabular components and has been shown to provide good initial stability [1-4]. With the press-fit fixation technique, a hemispherical porous-coated acetabular component, typically 1 to 4 mm larger than the last reamer used to prepare the acetabulum, is forcefully impacted into the acetabulum [5]. The objective of this technique is to press-fit the component into the host-bone, eliminating the need for supplemental fixation such as screws or spikes. The optimal technique produces a tight peripheral rim fit and minimizes gaps at the dome of the cup [3,5]. Although this technique has been shown to provide successful long-term fixation [3,6-8], concerns regarding its use exist. One obvious concern relates to the risk of acetabular fracture during impaction of an oversized component. Acetabular fracture has been demonstrated in cadaveric specimens implanted using the press-fit technique with components oversized by 2 to 4 mm [9], as well as in patients' hips surgically implanted with components oversized by 1 to 3 mm [10].

Recently, we have become aware of another potential consequence of the press-fit technique: acetabular component deformation secondary to forces encountered during insertion. This was first drawn to our attention when performing hip arthroplasty in younger patients with hard bone using metal-on-metal components where inserts are machined with a Morse taper to fit into the cup. In these patients, we noted that the metal inserts would not fully seat in the machined taper of the acetabular component and would toggle on the tight anteroposterior axis until firmly implanted,

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Fig. 1. Ultamet (Depuy, a Johnson & Johnson Co., Warsaw, IN) cobalt-chrome insert with a Pinnacle 100 acetabular cup.

indicating that some type deformation to the cup had occurred. As a result, we began a 2-part study to characterize this phenomenon. The first part of the study was an in vitro analysis, the purpose of which was to (1) simulate, via mechanical testing, the pelvic compression of an acetabular cup during total hip arthroplasty; (2) quantify the associated cup deformations; and (3) develop load vs deformation curves for specific cup sizes. The second part of the study was an in vivo analysis, the purpose of which was to (1) quantify postimplantation acetabular cup deformation in hip arthroplasty patients and (2) use the load vs deformation curves developed in vitro to assess the forces acting on cups implanted in vivo.

Materials and Methods

Part I: In Vitro Testing

Ten Pinnacle 100 titanium acetabular cups (DePuy, Warsaw, Ind) (Fig. 1) between 48 and 66 mm in size were analyzed on a servohydraulic testing machine (MTS Bionix 858 system, Eden Prairie, Minn). Custom load platens (Fig. 2) were manufactured to provide rim loading of the acetabular cup during testing. Rim loading was used to simulate the tight peripheral rim fit by the cortical bone along an axis that runs from the anterosuperior to the posteroinferior margin of the acetabulum. The Pinnacle acetabular cups are manufactured with rim derotation insets designed

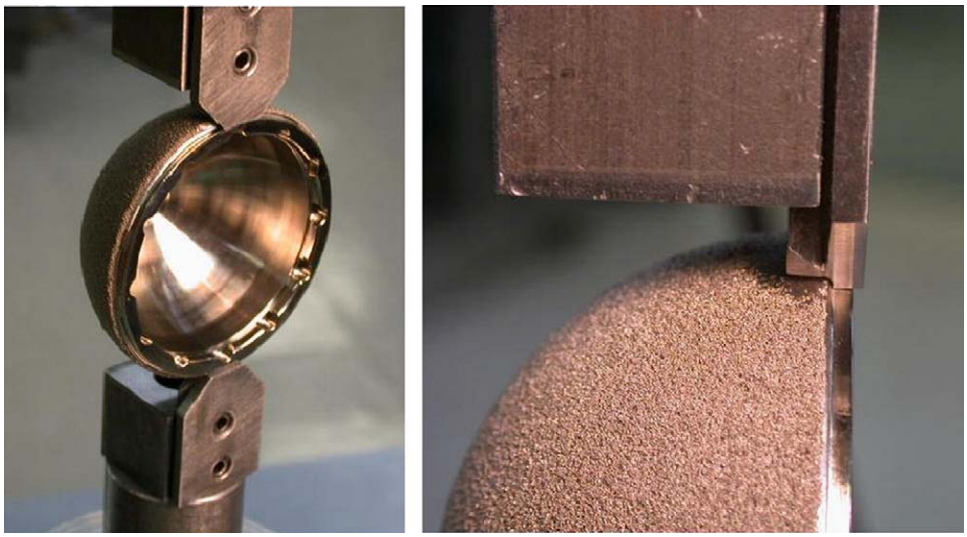

Fig. 2. Custom-designed load platens provided rim loading of the acetabular cup during mechanical testing.

Table 1. Cup Deformation and Compression Force Data

Cup Size (mm)	Bone Type	Diametrical Deformation (mm)	Calculated Force (N)
50	B	−.030	75
50	B	0.00	0
50	A	−0.47	1171
52	A	−0.10	242
52	C	−0.03	72
52	B	−0.04	96
52	A	−0.19	456
54	B	−0.41	957
54	A	−0.16	368
54	B	−0.24	552
56	B	−0.36	959
56	A	−0.57	1539
56	B	0.00	0
56	B	−0.01	27
56	A	−0.15	405
56	A	−0.15	405
56	B	−0.06	162
56	A	−0.19	513
58	B	−0.07	210
58	B	−0.12	360
60	A	−0.03	106

to interdigitate with tabs on mating polyethylene liners. The cups were mounted into the load platens such that 1 pair of diametrically opposed derotation insets was aligned with the axis of loading. The contact surface between the platen and the cup was a 12.5-mm wide arc, 3 mm deep, along the circumference of the rim (Fig. 2). The load platens were stepped front-to-back to facilitate clamping at the rim and increase clamping security. At the area of contact between the platens and the acetabular cup, the sintered bead coating was removed via grinding from the exterior surface to maximize seating on the rim and minimize slippage.

Mechanical testing was performed to determine the stiffness, defined as diametrical load per unit deformation, of each cup size. Measurements of the undeformed cup diameters were made with digital calipers (Mitutoyo, Model CD-6″BS, Aurora, Ill) at the inner surface of the cup rim before testing. Caliper accuracy was ± 0.01 mm. Each cup was subjected to compressive loads between 200 and 2000 N in increments of 200 N. After each load was applied, diameter measurements were taken between the opposing derotation insets at the cup rim. Deformation was defined as the difference in diameter between the loaded and the unloaded dimensions. Each loading cycle was run a total of 3 times with cup removal and reinstallation in the load platens between cycles. Stiffness values for individual tests were defined as the slope of the best-fit line for load vs deformation data in each load cycle. Stiffness values for the 3 load cycles were used to calculate the mean stiffness ± SEM for each cup size tested. Sigmaplot software (Systat Software, Inc, Point Richmond, Calif) was used for linear regression analysis of the load cycle data, and InStat software (GraphPad Software, Inc, San Diego, Calif) was used to determine mean and SE values for each series of tests.

Part II: In Vivo Testing

Eight men and 13 women with an average age at surgery of 59.2 ± 10.5 years underwent total hip arthroplasty by a single surgeon with insertion of a Pinnacle 100 (DePuy) acetabular component using a 1-mm press-fit technique. Cup size ranged from 50 to 60 mm (Table 1). For each patient, measurements of acetabular component diameter before and after cup insertion were made in 2 perpendicular directions (between opposing derotation insets that roughly corresponded to the anteroposterior and superoinferior axes) using a custom telescoping gage and vernier calipers (MSC Industrial Supply, Inc, Melville, NY). In each case, the gage was inserted between opposing insets, and the gage width was locked in place. The gage was removed, and the distance between the locked gage ends was measured using the vernier calipers. Each measurement was performed 3 times, and a final measurement was calculated as the average of 3 separate caliper readings. Again, deformation was defined as the difference in diameter between the loaded and the unloaded dimensions. For this patient population, preoperative diagnosis was osteoarthritis in 14 hips, avascular necrosis in 4 hips, and posttraumatic arthritis in 3 hips. Bone quality, assessed using the Dorr classification system [11], was type A in 9 hips, type B in 11 hips, and type C in 1 hip.

Using the load deformation curves developed in the in vitro portion of the study, the intraoperatively assessed deformation measurements were used to calculate the forces acting on the inserted cups. Force was calculated as the product of the measured deformation (in millimeters) and the stiffness of the cup (in newtons per millimeter). The forces acting on the cup were then compared among bone types, among diagnoses, and between sexes. Statistical analysis was performed using SPSS software (SPSS V8.0, SPSS Inc, Chicago, Ill).

Results

Part I: In Vitro Testing

Diametrical deformation measured during mechanical testing ranged from 0 to 0.9 mm, with an

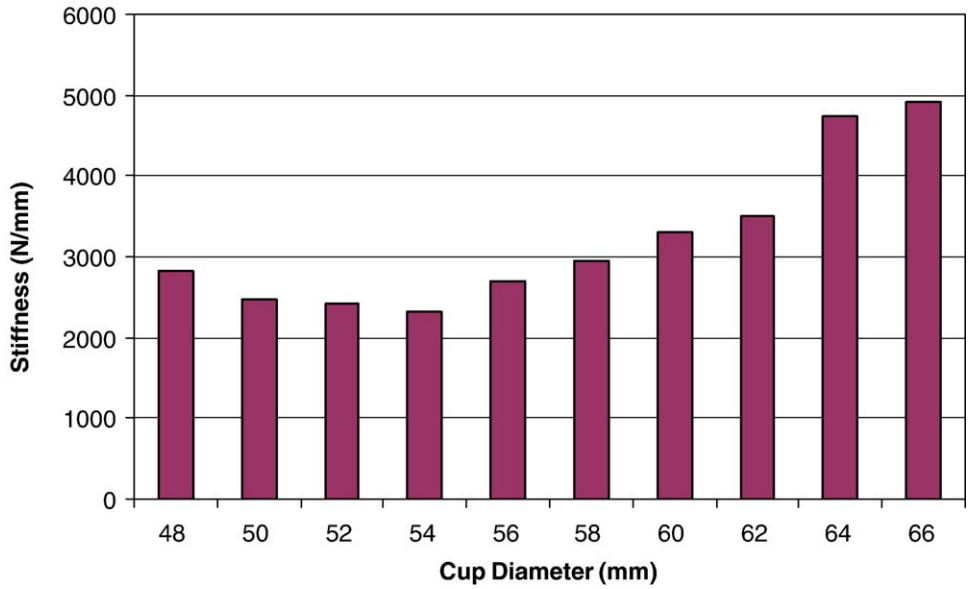

Fig. 3. Bar graph of stiffness values (mean ± SEM) for each of the 10 acetabular cups tested.

average deformation of 0.34 ± 0.21 mm. Linear correlation coefficients for load vs deformation curves ranged from 0.910 to 0.995, indicating that load vs diametrical deformation was strongly linear for all of the cups in the load range tested. Stiffness values, as assessed by the slope of the best-fit line for load vs deformation curves, ranged from 2333 ± 76 N/mm (mean ± SEM) for the 54-mm cup to 5205 ± 333 N/mm (mean ± SEM) for the 64-mm cup. Fig. 3 illustrates the measured stiffness values for each of the 10 cups tested.

Part II: In Vivo Testing

Intraoperatively, 90.5% of cups (19/21) had measurable compression deformity. Diametrical deformation ranged from 0.0 to 0.57 mm and averaged 0.16 ± 0.16 mm (Table 1). The corresponding forces acting on these cups calculated from these diametrical compressions ranged from 0 to 1539 N, with an average of 414 ± 421 N (Table 1). There was no correlation between compression force and patient age at surgery (P = .30, Spearman ρ), preoperative diagnosis (P = .94, Kruskal-Wallis nonparametric test), or patient sex (P = .55, Mann-Whitney nonparametric test). However, compressive force was marginally related to bone type. Cups press-fit into type A bone experienced 577.8 N of compressive force, whereas cups press-fit into type B bone experienced only 307.6 N of force (P = .07, Mann-Whitney nonparametric test). Power analysis demonstrated that we would need 30 patients per group for this difference to be significant at the .05 level (power = 0.8, α = .05, tails = 1). The single cup press-fit into type C bone experienced 72 N of compressive force.

Discussion

This study demonstrates deformation of porous-coated acetabular components as a consequence of the press-fit technique. Because the stiffness of each cup varies, however, direct comparison of diametrical deformation between cups in individual patients is inappropriate. Instead, we viewed the amount of deformation as an indirect measure of the force being applied to the cup during press-fit insertion. Calculating the applied force by multiplying diametrical deformation and cup stiffness resulted in normalized values, allowing comparisons among patients.

We found that the cups inserted in vivo experienced compressive forces up to 1.5 kN. In our population, these forces were not related to patient age, sex, or diagnosis but were marginally related to bone type. As we expected, cups inserted into harder bone (type A) experienced greater mean compressive forces than cups inserted into less dense bone (types B and C). This supports our clinical observations in which we experienced some toggle of the metal inserts on the anteroposterior axis before impaction into the Morse taper of the acetabular shell.

This phenomenon of acetabular component deformation during insertion may not have been a relevant issue in the past because most press-fit total hip arthroplasties were performed with metal shells subsequently coupled with polyethylene liners. Polyethylene liners, being more flexible compared with metal or ceramic liners, could easily deform to seat in a shell that had experienced some shape change during insertion. Moreover, it may not be a problem with stiff Co-Cr metal inserts as we observed that impaction of such inserts into the cup appeared to reexpand the cups, likely because of compression of the bone.

However, use of thin metal liners with press-fit shells may present unforeseen problems. Today, 1-piece metal-on-metal shells for resurfacings and standard arthroplasties are being manufactured with thinner walls to maximize femoral head sizes. Deformation of thin cups during insertion could lead to changes in bearing geometries, such as clearance and sphericity, which could adversely affect fluid-film lubrication and wear and, in extreme cases, eliminate the clearance and cause the joint to seize [12]. A recent cadaver and foam-model study by Jin et al [12] assessed this phenomenon for an experimental prototype thin metallic resurfacing shell having an outer diameter of 60 mm and a wall thickness between 2.3 mm at the equator and 4 mm at the pole. That study demonstrated diametrical cup deformation between 60 and 100 μm at the rim for a 0.5-mm to 1.0-mm interference fit during insertion. Compared with diametrical clearances of between 80 and 120 μm generally specified for fluid-film lubrication and adequate tribiologic performance of metal-on-metal implants, these deformations can be considered excessive [12]. A subsequent finite element study on similar implants demonstrated that cup deformation increased as (1) cup wall thickness decreased, (2) the interference increased, and (3) the size of the cup increased [13]. These studies underscore the importance of assessing cup deformation during press-fit insertion so as to ensure such deformation will not affect the tribiologic performance of the implants.

In addition to affecting the tribiology of metal-on-metal arthroplasty components, other potential problems include chip fractures of ceramic liners during insertion into deformed cups or the potential damage of locking mechanisms due to shell deformation.

We acknowledge several limitations to the current study. First, we tested only 1 type of cup. Clearly, our results are dependent upon the thickness and material of this component. We would expect results to vary with other shells and, therefore, suggest continued study of this phenomenon with multiple cup designs. Particularly we are interested in 1-piece metal-on-metal resurfacing designs with tight tolerances where cup deformation could result in major problems.

Second, because intraoperative deformation measurements were made by placing the telescoping gage into derotational insets, it is likely that we did not always measure the exact axis of maximum deformation. In such case, our measurements would not represent the maximum force experienced by the cup in vivo. Moreover, because of difficulties inherent to intraoperative measurements, in vivo measurement technique of cup deformation was slightly different from the in vitro measurement technique. We acknowledge that use of identical instruments and measuring techniques for in vitro and in vivo testing is optimal and would minimize potential sources of error in our study. However, the difficulty of measuring cup deformation intraoperatively precluded our taking of measurements in an identical fashion. The problems associated with intraoperative measurements included that the measurement device had to be customized to fit into the surgical wound at the time of cup implantation and that the measuring tools used in vivo had to be washed and sterilized between surgeries. Through many trials, we found that use of custom telescoping gage and vernier calipers gave us the most reliable in vivo data possible.

Third, we are not certain how long the deformation of the acetabular component lasts. Because viable bone has viscoelastic properties and the ability to relax and remodel, the deforming forces applied to these press-fit cups should subside over time. Although we were unable to examine this in detail because of time limitations during total hip arthroplasty, we were able in 3 hips to repeat deformation measurements 20 minutes after initial impaction of the acetabular component. In these cases, we found no change in the deformation with this short period.

Fourth, we acknowledge that use of the Dorr classification cannot truly quantify the quality of bone into which the cups were implanted. Such quantification would require dual energy x-ray absortiometry or computed tomography analysis. However, the Dorr classification, as used in this study, gave the authors insight as to a possible trend between bone quality and cup deformation—a trend that should be analyzed with more advanced technologies in future studies.

Finally, we acknowledge the limited number of cups studied intraoperatively may have precluded

our finding statistical relationships between force and patient age, sex, and diagnosis, as well as precluded our finding a stronger relationship between force and bone quality. It is possible that analysis of larger populations would uncover other patient variables correlated to this compressive force. For example, acetabular geometry could be one such variable. Surgically, we observed that Protrusio or deep acetabular geometries, which had an intact rim circumferentially around the component, seemed to experience less deformation. Conversely, mildly dysplastic shaped acetabuli, with no bony contact superiorly, seemed to experience more deformation presumably because forces were being applied to the cup only anteriorly and posteriorly. However, proof of such an observation would require quantification of additional variables, a more detailed analysis, and a larger population.

Despite these limitations, the current study demonstrates a potential consequence of the press-fit technique that may have unintended negative clinical consequences in patients with type A bone implanted with so-called hard-on-hard bearing surfaces. We advocate further study of this phenomenon with a larger population and a larger number of cup types so as to gain greater insight into the clinical variables correlated to force magnitude and the potential amount of deformation a cup could experience. The press-fit technique remains an excellent method of cup implantation, providing good long-term results while avoiding the risk of neurovascular injury and fretting and egress of particulate wear debris associated with the use of screws for implant fixation. However, greater understanding of cup deformation during press-fit insertion is important to orthopedic surgeons to aid their implant selection for specific types of patients, so as to avoid potential future problems.

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