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Original Research

Deeper Central Reaming May Enhance Initial Acetabular Shell Fixation

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ABSTRACT

Background: The initial stability of press-fit acetabular components is partially determined by the reaming technique. Nonhemispherical (NHS) acetabular shells, which have a larger radius at the rim than the dome, often require larger reaming preparations than the same-sized hemispherical (HS) shells. Furthermore, deeper central reaming may provide a more stable press fit. Using a reproducible, in vitro protocol, we compared initial shell stability under different reaming techniques with HS and NHS acetabular components.

Methods: Cavities for 54-mm NHS and 56-mm HS acetabular components were premachined in 20-pcf Sawbones blocks. Acetabular cavities included diameters of 54, 55, "54+," and "55+". "+" indicates a cavity with a 2-mm smaller diameter that is 2-mm deeper. A 4750N statically applied force seated shells to a height that was comparable with shell height after an orthopaedic surgeon's manual impaction. Force required to dislodge shells was assessed via a straight torque-out with a linear load.

Results: Increased preparation depth (+) was associated with deeper shell seating in all groups. Deeper central reaming increased required lever-out force for all groups. Overall, HS and NHS implants prepared with 55 + preparation had the highest lever-out forces, although this was not significantly higher than those with 54+.

Conclusions: In 20-pcf Sawbones, representing dense bone, overreaming depth by 1-mm improved initial seating measurements. In both HS and NHS acetabular shells, seating depth and required lever-out force were higher in the "+" category. It is unclear, however, whether a decreased diameter ream increased seating stability (55+ vs 54+). Clinically, this deeper central reaming technique may help initial acetabular stability.

Introduction

It has been shown that cementless fixation is achieved when there is minimal motion at a bone-implant interface [1], also termed initial stability. Failure to obtain initial stability may cause eventual failure of the cementless fixation and lead to loosening. Acetabular loosening is estimated to cause approximately 19% of total hip arthroplasty revisions [2-4]. Therefore, it is critical to identify and address factors affecting the initial stability of acetabular components.

Surgical technique plays a major role in achieving stability, yet several competing reaming techniques are introduced during training. Surgical protocols suggest a reaming range based on the implant design and the surgeon's subjective assessment of bone quality. Implant geometry, stiffness, reamer design, bone quality, and surgeon experience can all influence the chosen reaming technique. Some of the techniques of acetabular preparation include (1) line-to-line reaming, (2) underreaming, (3) overreaming, and (4) elliptical reaming. Elliptical reaming is achieved by reaming a second time and going deeper using a 1-mm-smaller reamer than the initial reamer used.

A core acetabular reaming principle is to prepare the pelvis with hemispherical (HS) reamers to create a uniform surface to increase contact area for the implant-bone interface. When an implant is seated into this preparation, initial contact is made at the rim

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(Fig. 1a). As the implant seats deeper, press fit increases at the rim until contact at the dome occurs (Fig. 1b). Once initial contact at the dome is achieved, further impaction forces may drive the acetabular shell slightly deeper into the prepared cavity (Fig. 1c).

Most press-fit acetabular shells are HS. The tested HS shells are designed to be implanted with a reamed prep 1-2 mm smaller than the implant size, achieving a 1- to 2-mm HS press fit. The other option for the shell design is nonhemispherical (NHS) "polar-flattened" or elliptical. The studied NHS acetabular shells are designed to be implanted with a reamed prep 0-1 mm larger than the stated implant size, achieving a 0.8- to 1.8-mm press fit at the rim (when prepared at minimum vs maximum recommended reaming preparation), while still achieving press fit at the dome. Using a standard HS reaming strategy, there will be less press fit at the dome than at the rim when implanting a NHS shell. Although HS and NHS designs use different design theories, both the shells have been successful clinically [5].

A strategy of "elliptical reaming" could theoretically be applied to aid in seating acetabular implants. Reaming deeper with a smaller reamer can allow the implant to sit deeper before bottoming out on the dome, helping assure rim press fit. It is unclear how this type of elliptical reaming technique can be used to maximize implant stability.

In this in vitro study design, we compared 4 reaming techniques in 2 acetabular component designs to assess initial stability of the shell press fit on initial impaction. We hypothesize that the optimal reaming strategy to achieve the best initial stability differs for each cup design and that deeper elliptical reaming can ease insertion forces while improving initial stability.

Material and methods

Implant size selection

For this study, 56-mm HS shells and 54-mm NHS shells from the same manufacturer were used. Both shells' substrates are forged Ti6Al4V and are coated with Cp Ti and HA with 0.8 coefficient of friction on bone and similar solid thickness, and these 2 shells require the same preparation size range. Fifty-six-millimeter HS and 54-mm NHS shells were chosen to minimize any effect of material, size, design, and cavity size on our results. In addition, they are commonly used shell sizes in our institution.

Owing to the different shapes of HS and NHS shells, different acetabular cavity preparations are required to achieve a similar press fit. In this study, we aimed for a 1- to 2-mm press fit, which corresponds to the manufacturer's guidelines. To achieve these values, the HS shell selected would need to be reamed 1-2 mm smaller than the labeled shell diameter. The NHS shell selected would need to be reamed 0-1 mm larger. For this reason, a 56-mm HS and a 54-mm NHS could both be implanted into 54- and 55-mm reamed cavities (Table 1). In clinical practice, the degree of desired press fit and ream size is clinician dependent.

Acetabular cavity preparation

To represent dense patient bone, 20-pounds-per-cubic foot (pcf)-dense Sawbones blocks (Sawbones, USA; Pacific Research Laboratories, Vashon, WA) were used to design a repeatable ranking study. Clinically, acetabular bone density is not homogenous, and the peripheral wall thickness around the shell rim is not consistent. One study found average apparent bone density in acetabular bone cores to be 15.6 pcf (ilium), 12.5 pcf (ischium), and 13.1 pcf (pubis) from male donors [6]. The Sawbones blocks selected in this study were designed to remove variability of the acetabular anatomy and ensure more repeatable homogenous properties than could be achieved in a cadaveric study. Sclerotic bone is expected to be denser locally but would likely require a more heterogeneous material model to assess. To reduce potential manufacturing and surgical variability, simulated acetabular cavities were prepared via routinely calibrated CNC (computer numeric−controlled) machining from a single sheet of Sawbones material.

Acetabular cavities were machined to simulate 4 desired reaming strategies: (1) 54-mm-diameter HS ream, (2) 55-mm-diameter HS ream, (3) "54+" elliptical ream, and (4) "55+" elliptical ream. These "nominal+" elliptical cavities simulated an additional 2-mm smaller-diameter ream that was reamed 2 mm deeper (Fig. 2). For example, the 54+ configuration simulated a 54-mm HS ream followed by a 52-mm HS ream an extra 2 mm deep.

Six cavities were machined for each reaming strategy and acetabular shell combination (Table 2). The HS acetabular preparations and implant combinations selected represent the manufacturer's suggested range of acetabular reaming preparations for the specified implants.

Shell impaction trials

Each Sawbones block was machined as prescribed and inspected with a routinely calibrated coordinate measuring machine. Blocks were held in a 45° fixture on top of dampening material to represent surgical impaction angulation. An instrumented mallet and a force sensor in the block system were used to collect force input received by the system when a single surgeon (T.R.H.) implanted a shell in each block (Fig. 3). One shell was implanted for each reaming strategy and shell combination to create a baseline achieved seating height. This correlates with previously used in vitro experimental setups [7].

Once each cavity and shell combinations were impacted, shell seating was measured using a height gauge, with seating height measured from each of the 4 removal slots to the top of the Sawbones block (Fig. 4). Seating was also visually confirmed through the dome hole.

Static-seating load calculation

Static seating was chosen as a shell insertion method for its reproducibility when compared with manual mallet or mechanical drop tower impaction (Fig. 5). Static seating applies a consistent load to an object at a constant rate. For this study, we chose to implant shells at a uniform rate of 0.1 mm/s to determine the force required to seat a shell at the same achieved seating height the surgeon impacted it to using a MTS Mechanical Test Frame; MTS Corporation, Eden Prairie, MN. This static seating method was internally validated and showed that manual mallet and drop towers had more significant outcome variability.

To calculate our desired seating force, achieved seating height described in the shell impaction trials section was collected. The achieved seating height for each shell and prep combination was used as the goal seating height. While conducting initial validation studies, it was observed that elastic deformation of the Sawbones blocks would relieve and lift the shell higher in the prepared cavity, by approximately 15% of the seating height achieved by the

Table 1
Suggested reaming ranges for 56-mm HS and 54-mm NHS.

	Minimum suggested reamed diameter (mm)	Maximum suggested reamed diameter (mm)
56-mm HS	54 ("2 mm under")	55 ("1 mm under")
54-mm NHS	54 ("line-to-line")	55 ("1 mm over")

NHS, nonhemispherical; HS, hemispherical.

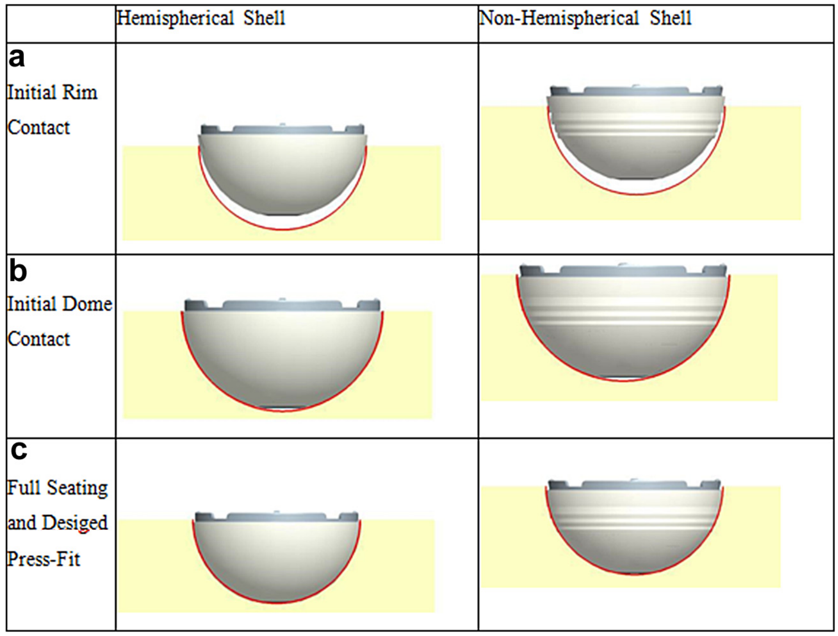

Figure 1. Theoretical spherical contact on shell seating (red: acetabular shell, yellow: reamed cavity): a) at initial rim contact; b) at initial dome contact; and c) at full seating and designed press-fit.

surgeon. To account for this, the shells were statically seated 15% deeper than the achieved seating height of the surgeon for all cavity combinations. Once the elastic deformation relaxed, 0.26 mm was the largest deviation between static seating height and the seating height achieved by the surgeon in all paired groups. The peak static force required to achieve this seating height across all groups was 4222N ± 527N. The average static-seating force required to achieve our initial goal height was augmented by 1SD of additional force to ensure proper seating. Clinically, if seating is difficult, a surgeon may impact additional times. The static load based on surgeon data used in this test is static displacement of (4222+1SD). This static-seating force will therefore ensure parts are fully seated without applying an inhuman seating force.

Static seating

A total of 6 samples for each group were statically seated at 0.1 mm/s until a peak force of 4750N was achieved using the MTS. Samples were levered out immediately after seating to minimize effects of stress relaxation on the initial stability of the shells.

Shell stability and lever-out force measurement

After the acetabular shells were statically seated into the prescribed simulated acetabular cavities, a threaded rod was attached to the dome hole insertion feature.

Each shell was then levered out with a displacement control single-axis load at a known distance (Fig. 6) that yielded 0.1 degree/second motion about the shell center using the MTS. The plot of angular displacement vs applied moment was assessed, which provided a yield moment, as measured by the 1-s offset intersect (Fig. 6) that was defined as the moment each shell began to move in the prepared cavity.

Statistical procedures

Seating height was measured between all nominal and nominal+ reaming strategies for NHS and HS implants together using a one-way analysis of variance with post hoc Tukey test. Lever-out force was compared between acetabular cup type, reaming diameter, and reaming strategy using a one-way analysis of variance with post hoc Tukey test. P-values less than 0.05 were considered significant.

Results

For each construct, the + reaming preparation resulted in a significantly deeper seating height for both NHS and HS implants (See Fig. 7a). For HS implants, the 55+ reaming strategy required the strongest lever-out force (see Fig. 7b). This was statistically significant when compared with 54 and 55. The 54+ reaming strategy also resulted in a significantly higher lever-out force than 55 (but not 54) ($P < .001$). For NHS implants, the 55+ reaming strategy in NHS also resulted in the highest lever-out force, which was statistically significant when compared with all other NHS reams ($P < .001$) (see Fig. 7b). The 54+ reaming strategy also resulted in a statistically significantly higher lever-out force than 54 ($P < .05$), but 55 had a significantly higher lever-out force than 54+ ($P < .001$).

When comparing NHS and HS implants, the 55+ reaming strategy had higher lever-out force in 20-pcf Sawbones for both implant types ($P < .01$). There was no significant difference between the NHS 55+ and HS 55+ groups.

Discussion

Studies have indicated that more than 150 μm of micromotion at the bone-prosthesis interface may lead to fibrous rather than stable bone ingrowth in canine models [1]. Owing to this relationship, it is

Table 2
Implants with prep ranges to be tested.

56-mm HS	54-mm NHS
54	54
54+	54+
55	55
55+	55+

NHS, nonhemispherical; HS, hemispherical.

believed that initial stability of metallic press-fit acetabular shells is critical for long-term success of total hip arthroplasty. Numerous studies have been conducted in an effort to determine technical factors that may enhance initial acetabular cup stability [8-15]. Most surgeons attempt to achieve a press fit by varying the reamer size and technique based on the patient's anatomy and implant design, as per the manufacturer's guidelines and their subjective assessment of bone quality, while avoiding excessive reaming that can increase the risk of exceeding host bone yield strength and result in an acetabular fracture [10]. There is evidence that underreaming by just 1 mm can achieve the same amount of initial stability as a 2-mm underream with less insertion force [9]. Furthermore, excessive underreaming may lead to incomplete component seating or increased fracture risk [10]. Meanwhile, line-to-line, or even slight overreaming, is indicated for some NHS shells, as described previously. In this study, all shells were seated via a set static-seating force, but in clinical practice, physicians who overream must be cautious of incompletely seating a cup implanted into a smaller diameter ream.

Manufacturers' guidelines prescribe differing acetabular cavity reaming techniques to maximize the press fit of individual shell designs. The HS implants tested in this study are designed for press fit into cavities that are reamed 1-2 mm smaller than the shell labeled diameter, providing 1- to 2-mm press fit. The NHS implants tested in this study are designed for press fit into acetabular cavities that are reamed to 0-1 mm larger in diameter than the shell labeled diameter, providing 0.8- to 1.8-mm press fit. A previous cadaveric study demonstrated that with both 1 and 2 mm of underreaming, acetabular components were initially stable clinically, but more than 150 μm of micromotion could be generated with variable bending forces ranging from just 49.3N to 214.4N [12]. While we did not perform micromotion analysis, we found that 1-mm underreaming for HS implant (1-mm press fit) with a centrally deeper (elliptical) ream required the highest lever-out force. For the NHS implant, 1-mm overreaming (0.8-mm press fit at the rim) with an elliptical technique of deeper central reaming provided the highest lever-out force. The NHS implant with a 1-mm overreaming and elliptical ream required the strongest lever-out force of any combination. This indicates that a "tighter" press fit was achieved with a centrally deeper cavity.

Elliptical reaming strategies increased the required lever-out force for every ream diameter and shell type. This was statistically significant for all groups except the 54+ HS group. This confirms our hypothesis that elliptical reaming increases lever-out

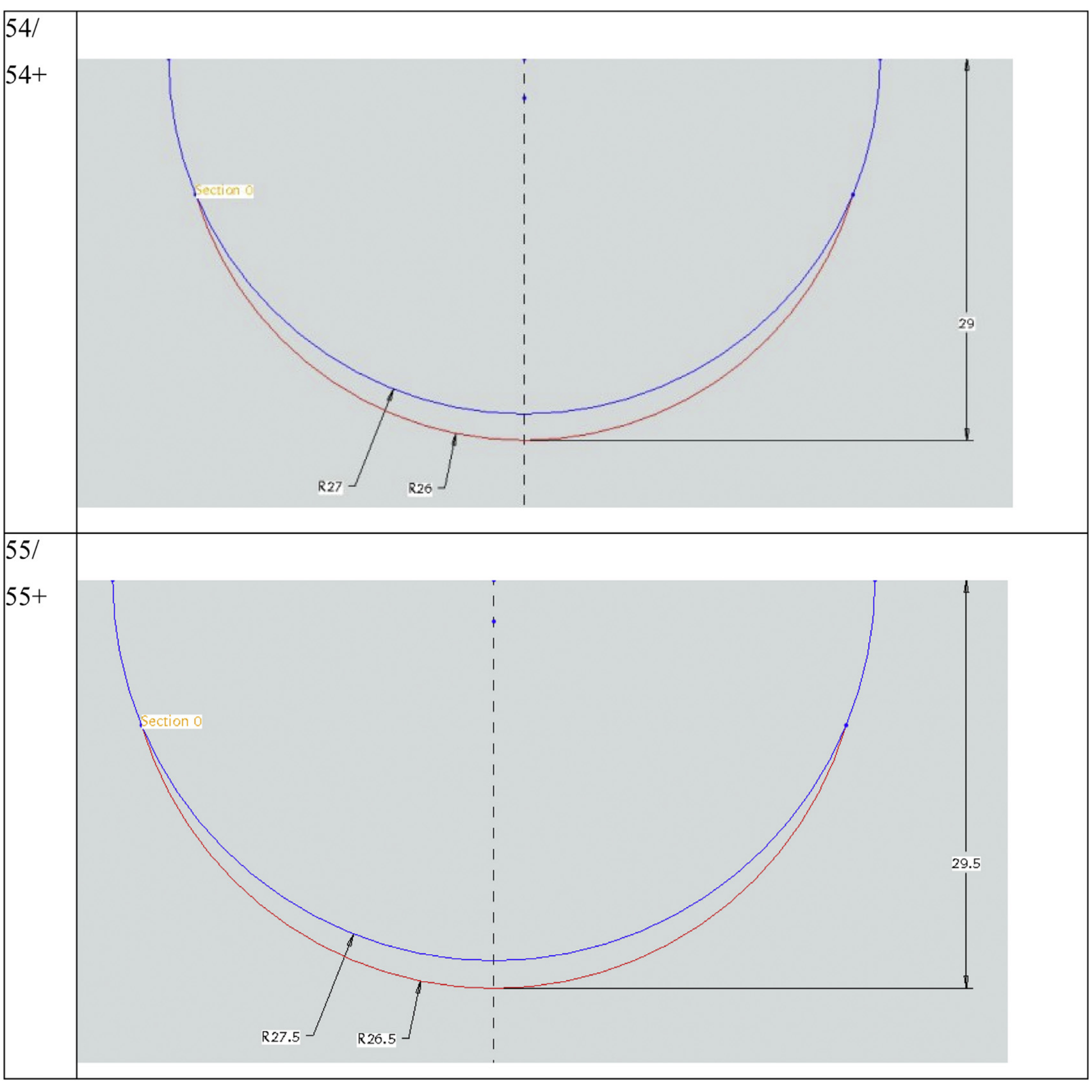

Figure 2. Example diagram of nominal (blue) and nominal+ elliptical (red) reaming strategies.

Figure 3. Experimental setup with instrumented mallet and force sensor impacting an acetabular component into a premachined Sawbones cavity.

force, which is a proxy for initial stability. Using an elliptical ream allows the peripheral rim to achieve the desired engagement with the acetabular cup before the shell's dome bottoms out. This allows greater contact with bone and, as indicated in this study, may increase initial stability.

Deeper seating leads to theoretically greater press fit and greater contact surface area at the interface between the implant and bone. As expected, increasing the size of the simulated acetabular cavities decreased the seating height of the acetabular shells in all groups tested. A significant decrease in seating height was measured for both HS and NHS implants when seated into 55- vs 54-mm cavities. There was a small but statistically significant decrease in seating height between 54 and 54+ and 55 and 55+ with both implants. It is possible this deeper seating height was a result of the additional 1 mm vertically removed and closer bone contact. One potential risk of deeper reaming is that medializing the acetabular component may overmedialize the mechanical center of rotation of the hip [13]; however, it is unclear whether the 1-2 mm of centrally deeper reaming here would reach clinical significance. Depending on individual anatomy, acetabular cup medialization with compensatory femoral offset is often desirable to decrease joint reactive force [16,17].

Our laboratory study design controls for numerous variables, providing valuable insights into the best acetabular preparation techniques for 2 popular cup designs in dense bone. However, there are still limitations inherent to our study. We use lever-out force to approximate the clinical concept of initial stability, but several other factors, such as bone quality, debris interposition, and a myriad of other surgical variables, influence in vivo stability. Second, for purposes of reproducibility, we used a static-seating design rather than a force impaction protocol more similar to surgical impaction. This may introduce other variables that are not considered in our model. Third, we assume the tolerances of the implants and the reamer diameters to be constant, whereas in the clinical scenario, reamers may be dulled [14] and implants may deform under excessive press fit [15]. Fourth, this benchtop study was performed on 20-pcf Sawbones blocks. This model represents dense patient bone. The reamer-to-implant relationships may change in rank with the less dense bone that is sometimes seen in patients undergoing total hip arthroplasties. Fifth, the nature of this study has intrinsic variability. Minor batch-to-batch variations in

Figure 4. Seating height measurement.

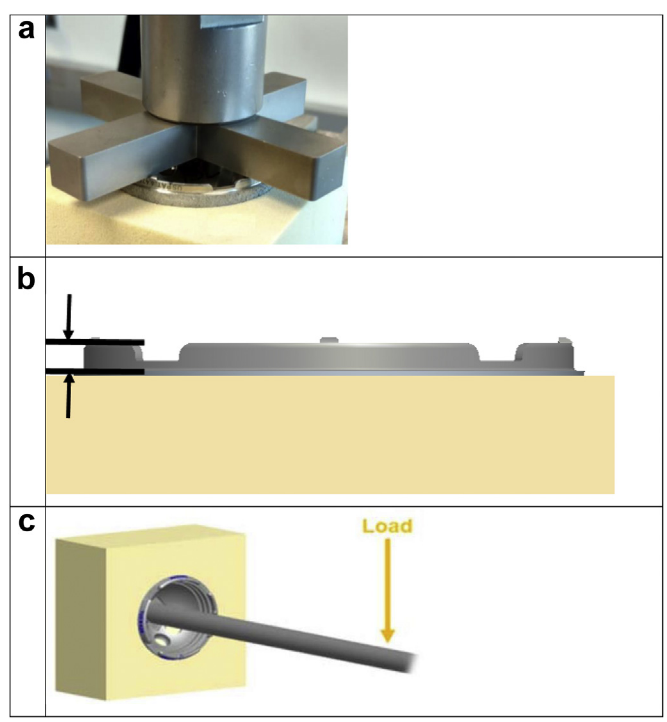

Figure 5. (a) Shell static-seating fixture. (b) Location of height measurement taken on the shell rim (4×). (c) Shell torque-out schematic.

Figure 6. Lever-out method.

Sawbones, time between shell seating and lever out, and general laboratory conditions can all influence future reproducibility. Sixth, this study used 2 different cups from the same manufacturer. There is likely an ideal reaming geometry unique to every combination of shell geometry [18], manufacturer material, coating [19], bone property, and individual patient and surgeon. Seventh, this study does not characterize the cavity geometry achieved by a reamer in the clinical setting. Our investigation assesses how changes in prepared cavity geometry may affect shell seating and initial stability. Clinically successful reamers have been shown to underream at the rim relative to the dome [20]. This study did not assess how clinically reamed cavities behave and how their cavity-to-shell relationship affects seating and stability. Finally, the results shown in this study are for 56-mm HS and 54-mm NHS shells. It is likely that different designs or sizes could have different relationships to press-fit design and stability. This may be of interest for future studies.

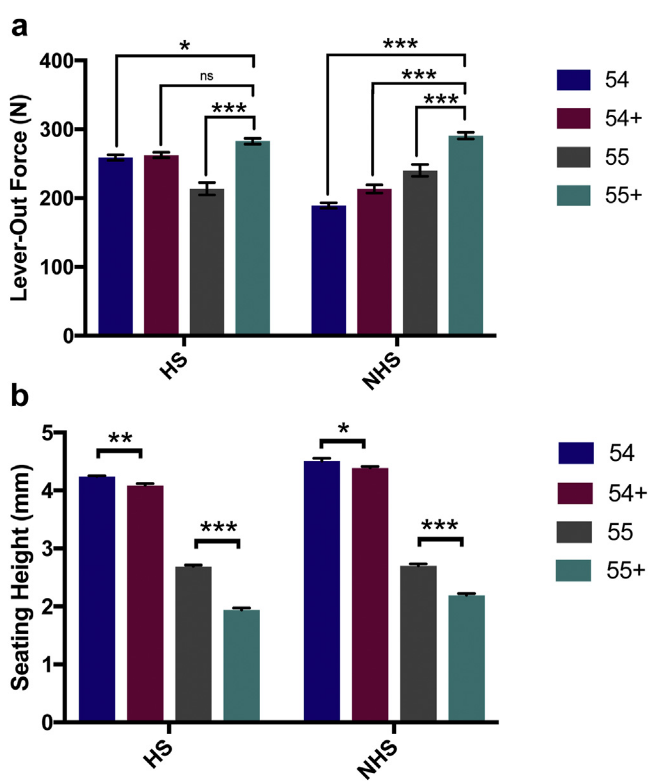

Figure 7. (a) Seating height of hemispherical (HS) and nonhemispherical (NHS) acetabular cup using different reaming techniques with standard error bars. (b) Lever-out force required to remove cups with standard error bars. Asterisks denote statistical significance between the groups.

Conclusions

Ultimately, there are different cavity preparation suggestions for different cup designs to maximize initial shell stability. Overall, this pilot study of an in vitro Sawbones system was reproducible and may be used in future studies to test initial acetabular shell seating and stability with different shell designs and different cavity reaming or machining strategies. Although it is not possible to draw direct clinical conclusions based on this model, our study indicates "elliptical" cavity geometry may increase initial stability of acetabular implants. Further experiments may include the testing of different densities of Sawbones and animal or human cadaveric bone specimens, replicating the cavity a reamer would create or different cavity preparation strategies in an effort to maximize initial in vitro acetabular shell stability with the ultimate goal of enhancing clinical in vivo acetabular press-fit shell stability.

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Conflict of interest

Mr. Davignon reports that he is an employee of Stryker. Dr. Geller reports royalties and speaker fees from Smith & Nephew; research support from Smith & Nephew, Orthopaedic Scientific Research Foundation, and OrthoSensor; and is a board member for the Journal of Arthroplasty and Journal of Bone and Joint Surgery. Dr. Cooper reports consultant fees from DePuy, A Johnson & Johnson Company, Joint Purification Systems, Inc., KCI Medical Canada, Inc., KCI USA, Inc., OnPoint Knee, Inc., and Zimmer-Biomet; research support from KCI and Smith & Nephew; and is a board member for the American Academy of Orthopedic Surgeons, the Journal of Arthroplasty, and the Journal of Bone and Joint Surgery. Dr. Shah reports consultant fees from Link Orthopaedics; is an unpaid consultant for OnPoint; research fees from KCI; and is a board member for the US Food and Drug Administration.

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