Increasing reaming depth enhances implant stability while minimizing bone strain

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Aims

Increasing the interference fit of the acetabular component can increase primary stability, but it introduces excessive periacetabular strain during impaction, which can lead to fractures. An optimal outcome following cementless acetabular component impaction is maximal primary implant stability with minimal periacetabular bone strain. The aim of this study was to investigate whether a simple modification to a surgeon's reaming technique can achieve this desirable outcome.

Methods

A custom drop rig simulated impaction strikes, seating acetabular components of either 1 mm or 2 mm interference fit into synthetic sawbones with cavities reamed to either a true hemisphere or a hemisphere with an enhanced reaming depth of 2 mm or 4 mm. Synthetic bone strain was recorded using strain gauges, and push-out tests were conducted to assess implant stability. Polar gaps were measured using optimal trackers.

Results

Compared to a true hemispherical cavity, enhancing the reaming depth significantly increased the primary stability of the implant ($p < 0.001$) while reducing both the periacetabular strain and strain deterioration for both 1 mm and 2 mm interference fit components. A 4 mm reaming depth enhanced the primary stability of 1 mm press-fit components to a level almost equivalent to a 2 mm press-fit, albeit reducing strain to the bone. Enhancing reaming depth did not significantly affect polar gap.

Conclusion

Enhancing cavity reaming depth is a simple technique to increase the implant primary stability of press-fit uncemented acetabular components, while avoiding any excess in periacetabular strain and the associated fracture risk.

Article focus

- An optimal outcome following cementless acetabular component impaction is maximal primary implant stability with minimal periacetabular bone strain.
- This in vitro study investigated the effect of enhanced reaming depth on these two outcomes.

Key messages

- Increasing reaming depth is an effective strategy to enhance primary implant stability while reducing periacetabular strain.

Strengths and limitations

- This is the first study, to our knowledge, to evaluate the primary stability and periacetabular strain of acetabular components following impaction resulting from reaming depth.
- The results provide surgeons with a simple technique to enhance component stability without compromising bone integrity.
- This study used synthetic bone, and hence the magnitudes may vary in live tissue; the transition between subchondral and cancellous bone in vivo is not accounted for.

Introduction

Maximizing primary implant stability while minimizing acetabular fracture risk is an optimal outcome of acetabular component impaction.[1,2] Implant primary stability is an important predictor of long-term stability and can be directly affected by intraoperative surgical technique.[3-5] Ideally, a balance needs to be struck whereby stability is maximized without compromising fixation through excessive bone strain and even risk of fracture.

Techniques to avoid implant micromotion are favourable for bone ingrowth.[6,7] Several strategies exist to enhance implant primary stability, such as increasing the implant interference fit or insertion of a screw.[8,9] However, these strategies are not without their complications. Screw insertion is associated with periacetabular osteolysis, neurovascular injury, and increased cost.[10,11] Considerations need to be made, however, of the resultant strain that is experienced by the surrounding bone, as this can result in periprosthetic fracture.[2,12] In cases where fracture is not the result, excessive strain can lead to large deformation of the acetabular component,[13] prevent liner seating,[14] cause excessive wear,[1] and even damage the component.[15] An ideal strategy to enhance fixation and minimize micromotion would enhance primary stability while minimizing periacetabular strain using a simple, reproducible technique.[5]

Acetabular reaming plays an important role in achieving component stability while minimizing unnecessary strain. During acetabular preparation, surgeons must ream an acetabular cavity. As well as deciding on how much to deepen the cavity and hence medialize the component,[16] surgeons need to consider the cavity shape. While it is usual to use the acetabular reamer to ream a hemisphere, this may not be the optimal cavity shape with regard to component stability and periprosthetic fracture risk.[17] Rather than reaming a true hemisphere, surgeons may choose to enhance reaming depth prior to component impaction. Enhancing reaming depth could allow the implant to sit deeper before bottoming out on the dome, helping to ensure rim press-fit.[18] It is unclear how this aspect of reaming technique can be used to maximize implant stability and its impact on periacetabular bone strain during impaction.

The effect of acetabular cavity shape on implant stability and periacetabular strain, particularly with regard to reaming depth, is not well understood, but it is important for several reasons. Firstly, understanding the relationship would enable surgeons to make informed decisions when reaming the acetabular cavity, to enhance stability and minimize periacetabular strain in light of the patient's characteristics. Secondly, it may lead to a reduction in complications associated with acetabular component impaction as surgeons may modify their implant interference fit choice and impaction force, particularly if stability or periacetabular strain are optimized by their reaming technique.

In this dry bone study, we evaluated the effect of different reaming depths on implant stability and periacetabular strain. We hypothesized that deeper reaming results in increased implant stability, reduced peak periacetabular strain, and a reduction in strain deterioration.

Methods

Setup

An in vitro drop rig which mimics in vivo pelvic boundary conditions was used to impact acetabular components into solid rigid polyurethane foam blocks (Sawbones Europe AB, Sweden) (Figure 1).[19,20] The foam blocks were of 15 lb ft^{-3} density. Solid rigid polyurethane foam was selected as its uniformity, consistent properties, and cutting behaviour make it an ideal alternative test medium for human cancellous bone in the testing of orthopaedic instruments.[9,21] A 15 lb ft^{-3} density was used as it replicates the exposed, cancellous bone which is expected to be encountered in a reamed acetabulum.[19] Furthermore, fractures and implant loosening are more of a concern in lower-density bone.[2] The sawbones were computer numerical control (CNC) milled to create 53 mm diameter cavities. This included two 'notches' to simulate the ischial and ilial column pinching effects, which is a previously validated anatomically representative geometry (Figure 2).[19]

The cavity was centred either at the level of the surface of the block, 2 mm below the surface of the block, or 4 mm below the surface of the block (Figure 3).

The drop rig recreates surgeon mallet impaction in a repeatable manner considering in vivo soft-tissue boundary conditions. The synthetic sawbone was mounted to a spring that provides the displacement and acceleration rate of impaction similar to the human acetabulum during surgery, resembling the soft-tissue damping. This test setup has been described and validated previously.[19]

Hemispherical acetabular components were manufactured which have a stiffness similar to commercially available press-fit titanium devices. The components have a 3 mm shell thickness and Ti6A14V alloy. The components are attached to the drop rig via a central spigot projecting from the inside of the component. Such components have been used previously in a validated study with a rough outer surface (Rz > 500 μm) applied to the components representing a typical implant fixation surface.[22] Then, 55 mm and 54 mm component sizes were chosen to recreate a 2 mm and 1 mm press-fit, respectively, which represents that used in typical clinical practice.[3,6,23,24]

Testing

A strike velocity of 2.75 m/s and strike mass of 1.2 kg was used to represent a medium-strike velocity and strike mass based on previous literature and testing.[20,25–27] Five repeats were performed for each of the three reaming depth configurations for both 1 mm interference fits and 2 mm interference fits. For each setup, ten strikes were performed recording both strain and displacement data, resulting in a total of 180 tests.

Implant stability: strain measurement

Strain gauges (350 ohm, SGT-3F/350TY11; Omega, UK) were applied to the cavity of the sawbone periphery 2 mm from the edge on the superior surface using a validated method to measure hoop stresses. This area was chosen as it has been validated and shown to experience the highest levels of strain.[28,29] A USB data logger was used to acquire the strain gauge data for one second after each strike (NI9327; National Instruments, Hungary). Labview (National Instruments, 2014) was used to capture the raw data which were subsequently analyzed using MATLAB (Mathworks, USA).

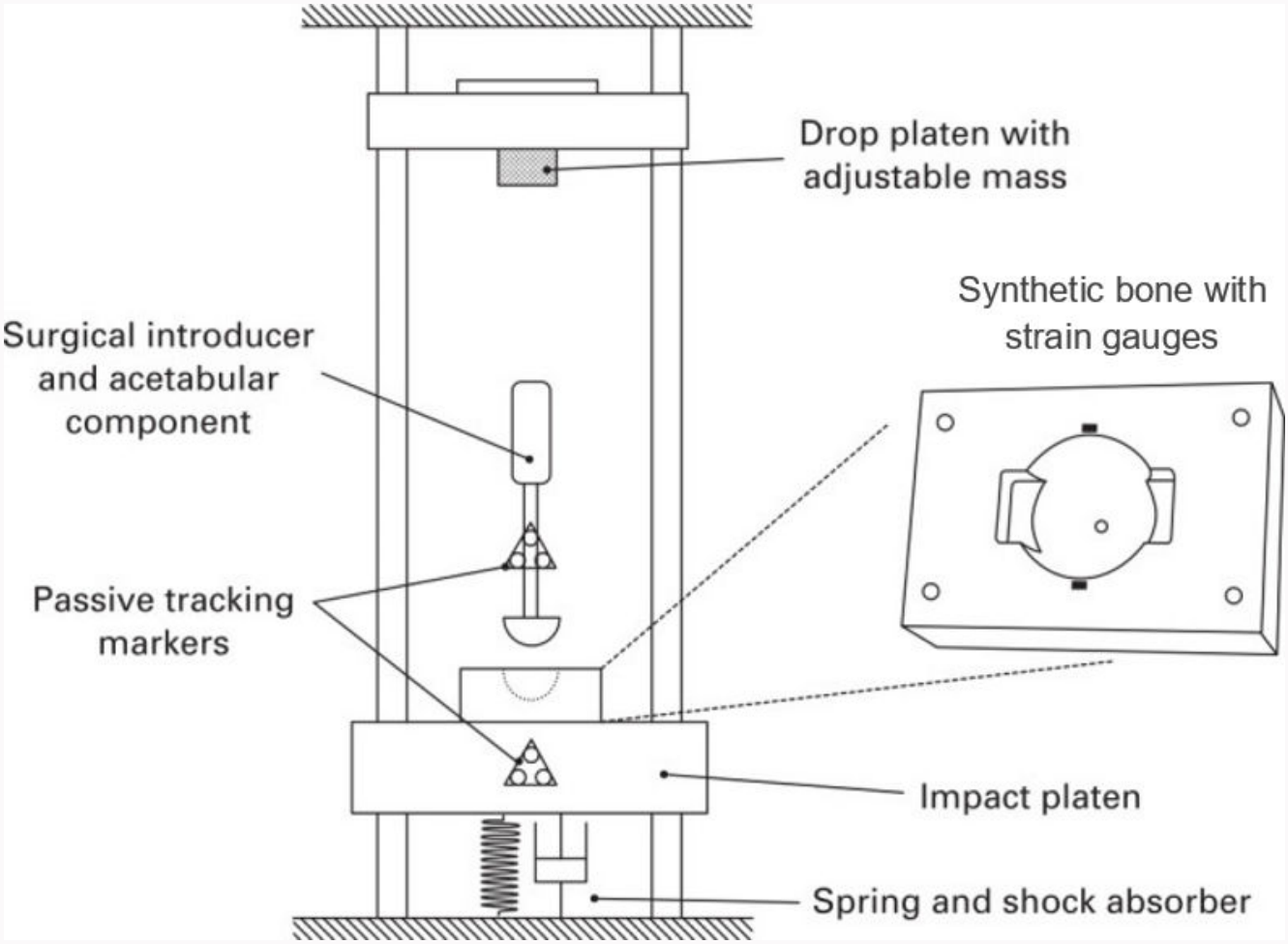

Fig. 1
Experimental setup.

Strain deterioration was determined by the difference between maximum and minimum strain over ten strikes divided by number of strikes between these two values. With increasing strikes, the sawbones demonstrate a predictable strain pattern with an increase in strain initially up to peak followed by a deterioration. This represents increased stability of the implant after which subsequent strikes represent over impaction and hence may reduce implant stability.[19] A larger strain deterioration would represent a situation whereby over impacting the implant leads to a bigger reduction in bone strain, and hence a larger compromise in implant stability.

Component seating

Infrared motion trackers were fixed to the component introduced and the sawbone models. A motion capture camera (Polaris Vega; NDI, Canada) recorded the position of the trackers after each strike to determine the position of the acetabular component following each strike. Ten strikes were performed, following which a depth gauge micrometre was used to determine the final polar gap. These data alongside the motion capture positions were used to calculate polar gap after each strike. The number of strikes after which the cap failed to progress > 0.1 mm was determined as the 'optimal number of strikes' for the impaction setup.

Pushout tests

A uniaxial load testing machine (model 5565; Instron, UK) was used to perform pushout tests following impactions with the 'optimal number of strikes.' A 6 mm diameter steep pin was used at a rate of 0.5 mm/min through a 7 mm entry hole at the back of the synthetic bone block which was held rigid, and movement was restricted in all directions during testing. This was chosen as a surrogate for fixation strength due to its replicability and correlation with micromotion tests.[30]

Statistical analysis

Risk of fracture of each of the six testing configurations (1 mm interference fit with 0 mm, 2 mm, and 4 mm depth cavities; 2 mm interference fit with 0 mm, 2 mm, and 4 mm depth cavities) was evaluated by correlating the peak bone strains and the impaction energy. Implant stability was evaluated by correlation pushout strength and impaction energy. Strain deterioration for each configuration was also evaluated to determine the relative implications of over impaction. Normality was tested using a Shapiro-Wilk test. Data were analyzed with a two-way repeated measures analysis of variance (ANOVA) with a post-hoc Tukey multiple comparison test. All analysis was performed in SPSS v24 (IBM, USA). Statistical significance was set at $p < 0.05$.

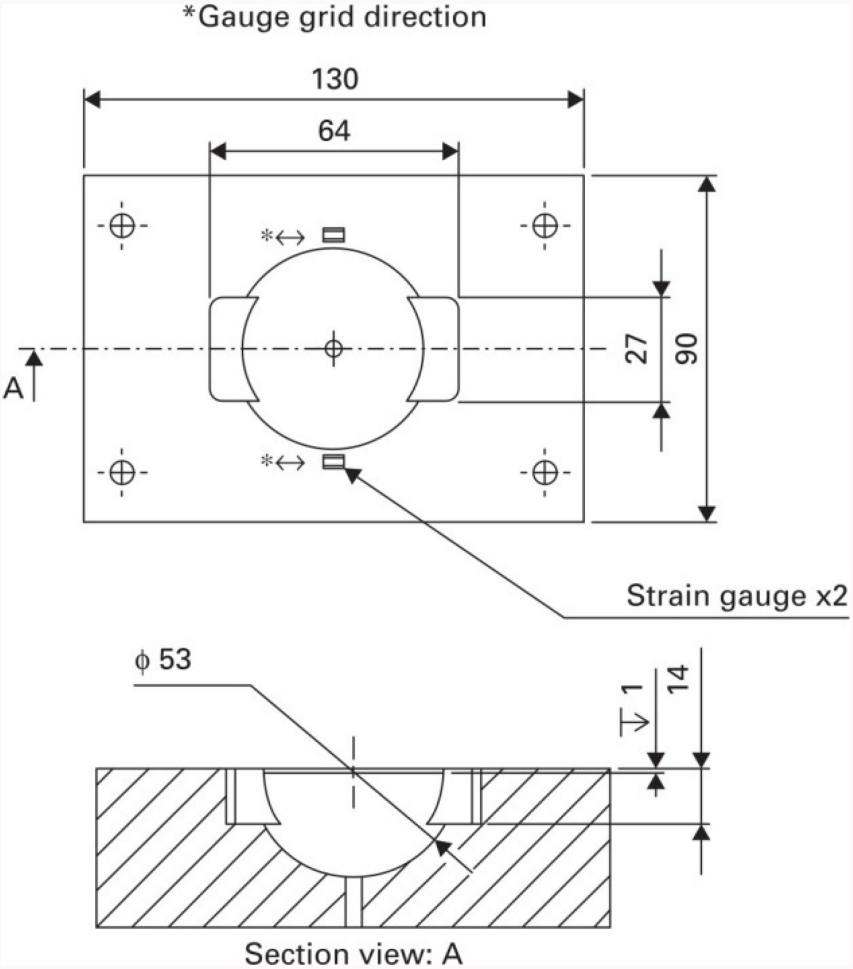

Fig. 2
Synthetic bone cavity cutouts and strain gauge placement.

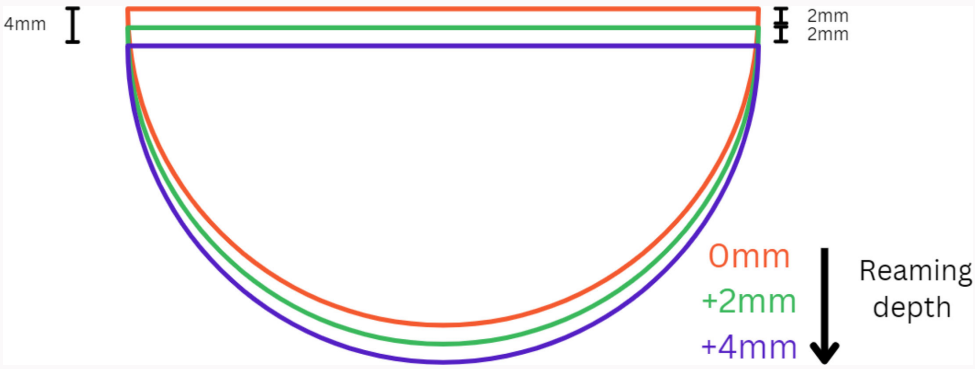

Fig. 3
Enhanced reaming depth configurations.

Results

Pushout

Central reaming depth increased the mean push out force for both the 1 mm and 2 mm interference groups ($p < 0.001$, two-way ANOVA with post-hoc Tukey multiple comparison test) (Figure 4). For components with a 1 mm interference fit, the mean pushout force increased by 26% between 0 mm and 4 mm reaming depths (292 N (SD 38) vs 368 N (SD 12); $p = 0.005$, two-way ANOVA with a post-hoc Tukey multiple comparison test) while a 2 mm interference increased the pushout force by 35% between 0 mm and 4 mm (397 N (SD 32) vs 537 N (SD 46); $p < 0.001$, two-way ANOVA with a post-hoc Tukey multiple comparison test).

As expected, increasing interference also increased the mean pushout force, by 36% for the 0 mm (292 N (SD 38) vs 397 N (SD 32); $p < 0.001$), 21% for the 2 mm (346 N (SD 36) vs

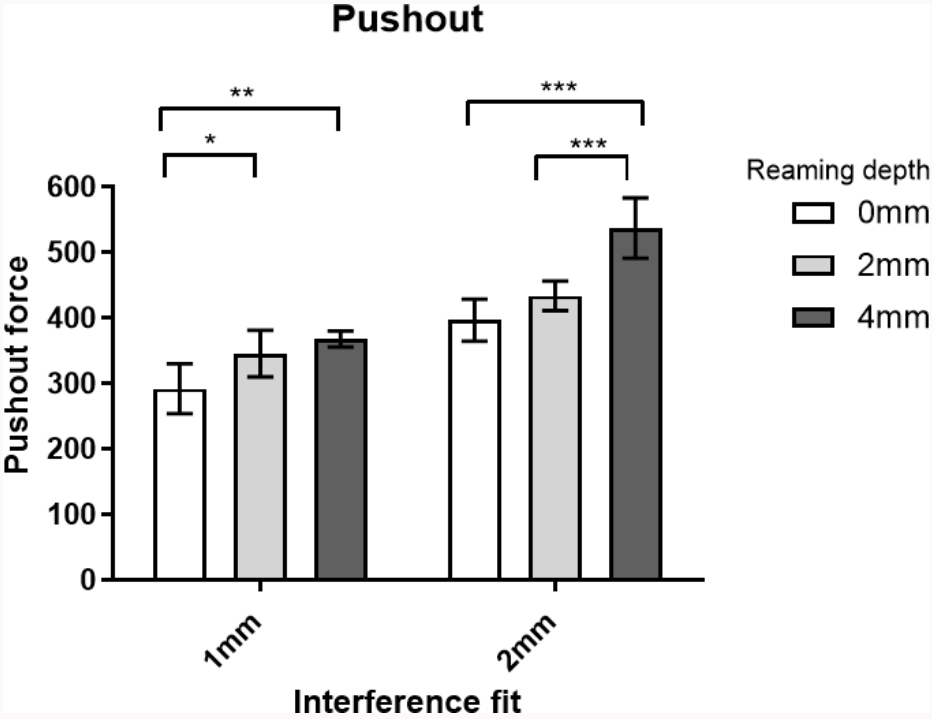

Fig. 4
Mean push force for 1 mm and 2 mm interference fit components with each reaming depth following impaction. *, **, and *** indicate p-values of < 0.050, < 0.005, and < 0.001, respectively; two-way analysis of variance with a post-hoc Tukey multiple comparison test.

434 N (SD 22); $p < 0.001$), and 46% for the 4 mm (368 N (SD 12) vs 537 N (SD 46); $p < 0.001$, two-way ANOVA with a post-hoc Tukey multiple comparison test) reaming depths.

Enhancing reaming depth did not significantly affect the mean polar gap for either the 1 mm under-ream group (0.03 mm (SD 0.02) vs 0.03 mm (SD 0.03) vs 0.17 mm (SD 0.08)) or the 2 mm under-ream group (0.54 mm (SD 0.06) vs 0.62 mm (SD 0.2) vs 0.60 mm (SD 0.12)).

Peak strain

Mean peak strains were highest in the 0 mm reaming depth group for both interference fits. There was no increase in mean peak strain with enhanced reaming depth. Increasing the reaming depth by 4 mm significantly reduced the peak strain in the 1 mm interference group (106.7 (SD 42.8) vs 47.8 (SD 10.2); $p = 0.040$).

Increasing interference more than doubled the peak strains at 0 mm (106.7 (SD 42.8) vs 262.8 (SD 49.5); $p < 0.001$), while by 4 mm, a doubling of the interference from 1 to 2 mm resulted in an almost five-fold increase in strain 4 mm (47.8 (SD 10.2) vs 226.7 (SD 26.8); $p < 0.001$, two-way ANOVA with a post-hoc Tukey multiple comparison test) reaming depths (Figure 5).

Strain deterioration

Implant impaction demonstrated a peak bone strain followed by strain deterioration as expected, with a similar pattern for all interference fits and reaming depths (Figure 6). For the 1 mm interference fit, the peak strain occurred at the first strike for all reaming depths. With 2 mm interference, peak strain still occurred at first strike at 2 mm reaming depth, while the peak strain did not occur until second strike with a 4 mm reaming depth and third strike with a 4 mm reaming depth. Strain deterioration decreased significantly as reaming depth increased in both the 1 mm and 2 mm interference groups ($p < 0.001$). However, this difference could be caused by the rim of the component passing distal to the surface (where the strain gauge was), thus relaxing the strain. Impactions of a component with a 2 mm interference fit resulted in a higher mean strain deterioration compared to the 1 mm interference reaching significance in the 0 mm (16.9 $\mu\varepsilon$ (SD 1.4) vs 13.2 $\mu\varepsilon$ (SD 3.8); $p < 0.050$), 2 mm (15.2 $\mu\varepsilon$ (SD 2.5) vs 10.3 $\mu\varepsilon$ (SD 1.6); $p < 0.005$), and 4 mm (11.4 $\mu\varepsilon$ (SD 2.1) vs 5.6 $\mu\varepsilon$ (SD 1.8); $p < 0.001$) reaming groups (Figure 7).

Discussion

This dry bone study demonstrates that primary stability is optimized by enhanced reaming. Using this method, primary stability can be optimized while avoiding the increases in peak strains seen with increasing interference fit, and hence avoiding risking bone integrity. For a 1 mm press-fit, an increase to the reaming depth by 4 mm increased fixation strength to almost the same strength as a 2 mm press-fit, but with a fifth of the bone strain. The likely higher contact area between the implant and bone may explain our findings.[31] Increasing contact area would enhance primary stability as well as reduce the stress at any one point due to the increased area.

Several studies have evaluated technical factors that affect implant stability, however there is a lack of guidance regarding the effect of reaming depth. One previous study has evaluated the effect of central reaming depth on lever-out strength in sawbones, demonstrating increased stability by increasing reaming depth with a 2 mm interference fit but not a 1 mm interference fit.[18] The authors used a smaller 52 mm reamer to enhance central reaming depth

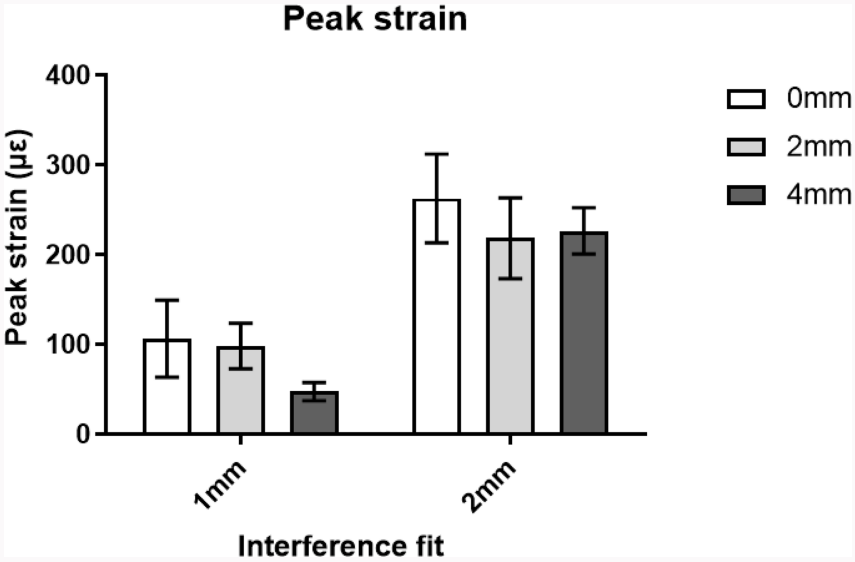

Fig. 5
Mean peak strain for 1 mm and 2 mm interference fit components with each reaming depth following impactions.

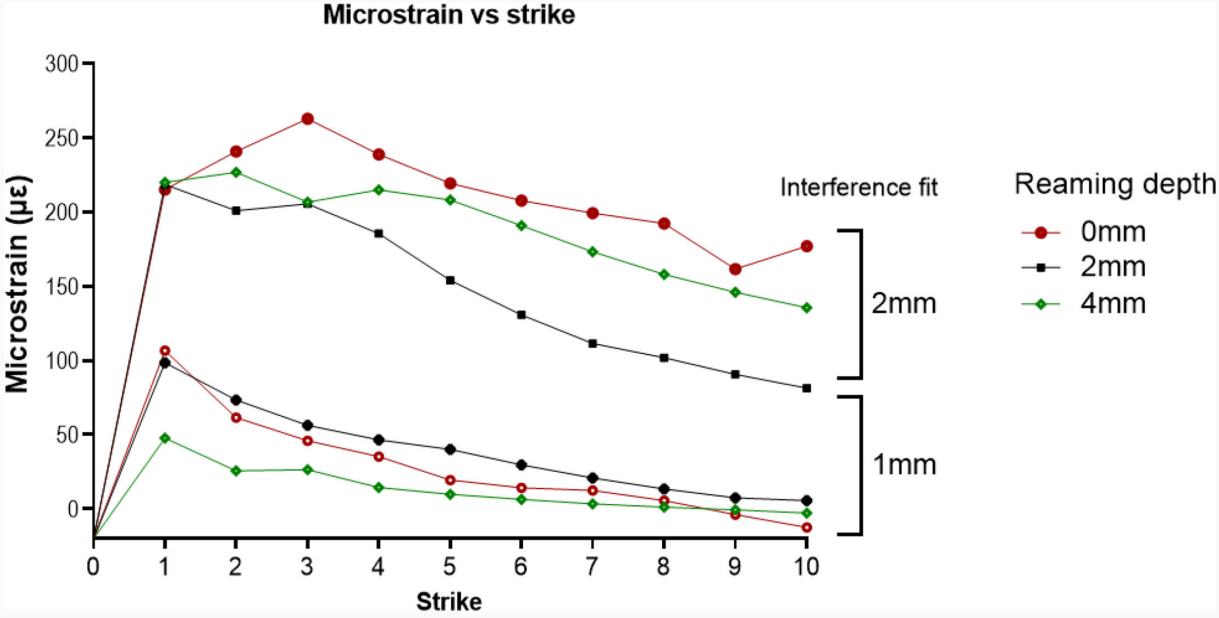

Fig. 6
Strain pattern for 1 mm and 2 mm interference fit components with each reaming depth with subsequent impactions.

of a 54 mm reamed cavity in 20 lb ft^{-3} sawbones by 2 mm, whereas our sawbone reaming depth was enhanced using the same size 53 mm reamer. We felt that this more realistically represented a commonly used technique and would also be more replicable intraoperatively. The authors also statically loaded the components, whereas the present study simulated impaction using a validated impaction rig mimicking surgical strikes taking into account soft-tissue boundary conditions.[30] Our study also evaluated periacetabular strain, which is an important determinant of fracture risk and micromotion.[19]

Using an impaction rig that provides the same displacement and acceleration rate following a strike as the human acetabulum during surgery ensures that a similar damping of soft-tissue seen in vivo is taken into account in vitro. Our research group has previously demonstrated that excessive impactions reduce primary stability, so we used this method to determine the optimal number of impactions for seating. By doing so, we aimed to perform a controlled experiment replicating in vivo conditions.[19,20,29] Despite this, there are several obvious limitations inherent to our study. We used 15 lb ft^{-3} density sawbones, attempting to replicate the low-density bone representative of the exposed cancellous bone after reaming. It also represents the low-density bone quality most likely to sustain intraoperative fractures and poor primary implant stability. However, it may not be representative of denser subchondral bone, and hence further work in higher-density sawbones would be beneficial. As this is an in vitro study, we are unable to take into

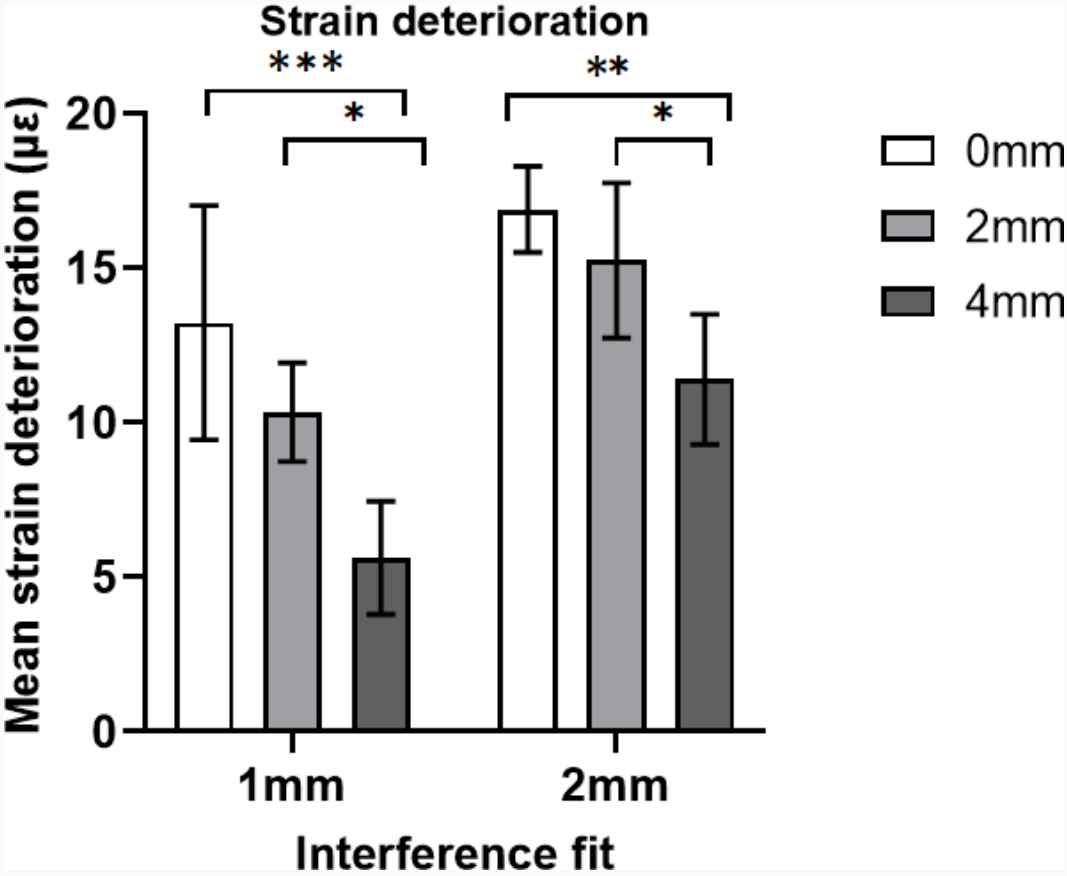

Fig. 7

Mean strain deterioration for 1 mm and 2 mm interference fit components with each reaming depth following impactions. *, **, and *** indicate p-values of < 0.050, < 0.005, and < 0.001, respectively; two-way analysis of variance with a post-hoc Tukey multiple comparison test.

account the inherent anisotropy of arthritic acetabular bone. Having a uniform testing protocol, however, with uniform and validated sawbones enables us to control numerous variables, enhancing the validity of our conclusions. Clinically, increasing central reaming depth may have biomechanical consequences. This not only includes the graduation of subchondral bone to cancellous bone which occurs with increasing reaming depth, but also the increased bone loss. Without increasing component diameter, increased depth will also result in component medialization, which will impact on global offset, potentially resulting in a requirement to change femoral offset.[16,32] In addition, deepening the socket can increase the risk of impingement. These factors must be considered by the surgeon when increasing reaming depth to enhance implant stability. We used push-out force to measure primary stability, which is a well-validated method. However, in vivo, several other factors need to be considered with regard to anatomy, bone quality, and soft-tissue debris, among others. Furthermore, we assume that the reamers and sawbones are uniform when a small amount of tolerance is inherent. To combat this, all sawbones were CNC machined, and we performed numerous repeats. Finally, we used only one component type for consistency. Further work evaluating the effect of enhanced reaming depth with peripherally expanded components may be beneficial.

In summary, the goal of maximizing primary implant stability while minimizing fracture risk is achievable by enhancing reaming depth. Enhancing reaming depth by as little as 2 mm enhances the primary stability of press-fit uncemented acetabular components while avoiding any excess in periacetabular strain and the associated fracture risk.

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Data sharing

The data that support the findings for this study are available to other researchers from the corresponding author upon reasonable request.

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