Factors Influencing Initial Cup Stability in Total Hip Arthroplasty

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Abstract: 244 words
Main text: 3674 words
Number of Tables: 2
Number of Figures: 10

Abstract

Background: One of the main goals in total hip replacement is to preserve the integrity of the hip kinematics, by well positioning the cup and to make sure its initial stability is congruent and attained. Achieving the latter is not trivial.

Methods: A finite element model of the cup-bone interface simulating a realistic insertion and analysis of different scenarios of cup penetration, insertion, under reaming and loading is investigated to determine certain measurable factors sensitivity to stress-strain outcome. The insertion force during hammering and its relation to the cup penetration during implantation is also investigated with the goal of determining the initial stability of the acetabular cup during total hip arthroplasty. The mathematical model was run in various configurations to simulate 1 and 2 mm of under-reaming at various imposed insertion distances to mimic hammering and insertion of cup insertion into the pelvis. Surface contact and micromotion at the cup-bone interface were evaluated after simulated cup insertion and post-operative loading conditions.

Findings: The results suggest a direct correlation between under-reaming and insertion force used to insert the acetabular cup on the micromotion and fixation at the cup-bone interface.

Interpretation: while increased under-reaming and insertion force result in an increase amount of stability at the interface, approximately the same percentage of surface contact and micromotion reduction can be achieved with less insertion force. We need to exercise caution to determine the optimal configuration which achieves a good conformity without approaching the yield strength for bone.

Keywords: total hip replacement, cup fixation, interface conformity, micromotion, finite element analysis

1. Introduction

Each year, as many as 200,000 total hip replacements (THR) are performed in the United States [1], and approximately 7% of those require revision arthroplasty within 8 years of the initial procedure [2]. Revision of hip arthroplasty occurs in up to 25% of all arthroplasties performed in the US and has a less favorable outcome than primary THR [3]. It's worth mentioning that THA survivorship at 15 years for revision is at 69% [4]. It is also important to note that Total hip replacement is a successful and cost effective procedure that offers immediate relief of pain and considerable improvement of life daily function to patients suffering with osteoarthritis of the hip [5–13]. Risk factors for THR revision are patient-related (e.g., gender, neuromuscular disorder status, bone quality) or surgery-related (e.g., surgical approach of primary THA, orientation of the cup, component malpositioning, femoral head size, neck head offset, and surgeon experience) [14–18]. Callanan et al [19] reported an increased risk of acetabular cup malposition, particularly for minimally invasive approaches, low volume surgeons, and obese patients. Wetters [20] found that 40% of the patients with at least one episode of instability after revision THA were subject to recurrent instability. (see Figure 1 and 2 for discolation and instability related to THA). Osteolysis and aseptic loosening are the most common reasons for revision hip arthroplasty [3], and prosthetic loosening is most likely related to the technique used for implant fixation [21,22].

Many orthopedic surgeons agree that stability and duration of the implant depend more on the implantation technique than the type of implant used [23–26]. Several studies have compared the effect of various implantation methods, such as degree of under-reaming [27] or the use of screw fixation [28,29], on cup stability. Although the amount of under-reaming is left to the discretion of the surgeon, most surgeons under-ream the acetabulum by 1.0 or 2.0 mm [30,31].

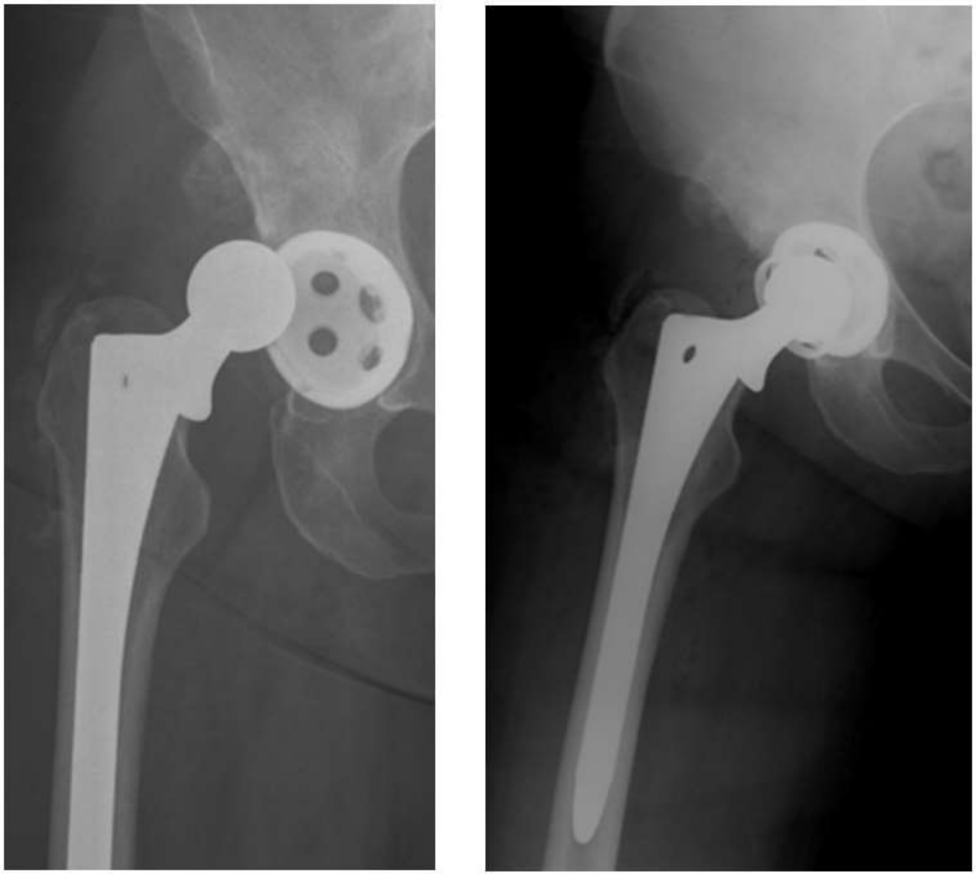

Fig. 1. Dislocation of the THA due to a vertically placed acetabulum component B. Acetabular shell maintained converted into a constrained liner.

For greater amounts of under-reaming, such as 3.0 or 4.0 mm, no significant difference in stability has been reported; however, in cases of 4.0-mm or greater under-reaming, press-fit insertion of acetabular cups has resulted in acetabular wall fracture [32,33]. For cementless acetabular cups without screw fixation, one study reported polar gap distances at the apex of the cup in 17.8% of patients immediately following surgery [34]. Excessive retrovision may result in posterior dislocation whereas excessive abduction may result in lateral dislocation [35,36]. Radiographic assessment may not be sufficient to evaluate hip prosthesis positioning and measurement of anteversion. THA instability continues to be a major issue [17,37,38] and is critically important to patient outcome; however, we do not fully understand the influence of the cup-bone interface, reaming conditions, bone quality (osteoporosis), and initial conditions surrounding cup penetration and insertion hammering forces on cup stability and THR outcome. In this study, our objective was to investigate the mechanics of the cup/bone interface, reaming, and insertion forces on the stability of the cup under different loading conditions.

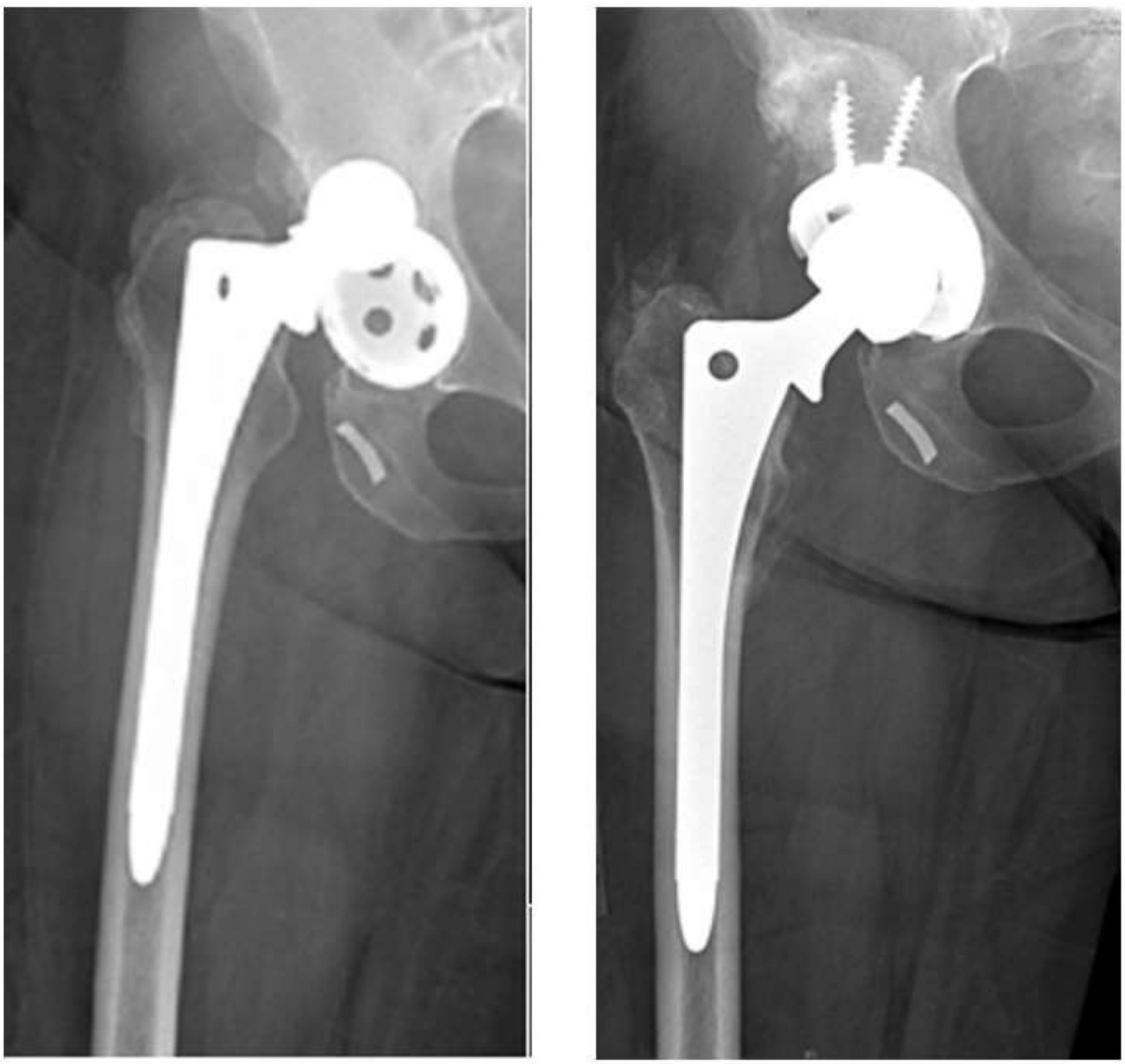

Fig. 2. Dislocation of the hip with failure of constrained liner (broken locking ring)   B. Final revision

acetabular shell placed in anatomic position with unconstrained liner (the constrained liner didn't address the pathology of the initial placed vertical cup).

We developed a finite element model that would allow for patient-specific analysis of cup insertion factors and cup stability, and then validated the model in vitro. Cup stability was measured as a function of polar gap distance, surface conformity, and micromotion at the cup-bone interface immediately following THR.

2. Methods

2.1 In Vitro Experimental Study

Five pelvis were obtained from fresh frozen cadavers, stored at approximately -20°C. No bony or musculoskeletal abnormalities were noted upon visual inspection prior to hip prosthesis implantation. Each pelvis was dissected using a posterolateral approach to expose the acetabular rim. The acetabulum was sequentially reamed by 1-mm increments until cartilage was fully removed and cancellous bone was clearly exposed and visible within the acetabulum. A Titanium Pinnacle hemispherical acetabular cup (Johnson & Johnson, DePuy, Warsaw, IN) with a diameter 1 mm greater than the final reaming diameter was press-fit inserted. Following cup insertion, all soft tissue was removed and the pelvis was divided at the sacral and pubic joints to preserve the integrity of the two hemi-portions. The hemi-pelvis was mounted in Bondo polyester resin (Bondo Corp., Atlanta, GA) so that the ilium was fully constrained and the ischium was constrained with an adjustable steel stand to mimic the physiological constraints of the pubic symphysis and the sacro-iliac joint.

A reference system using the bony landmarks of the hemi-pelvis was utilized to ensure that load application and sensor placement was well-documented. The acetabulum was divided into a four-quadrant system defined by Wasielewski [39], with the anterior superior iliac spine (ASIS) marking the starting point of Line 1, which extends through to the ischial tuberosity. The $\overline{XY}$ plane lies parallel to the rim of the cup and the y-axis lies parallel to the projection of Line 1 onto the $\overline{XY}$ plane. The z-axis lies perpendicular to $\overline{XY}$ and extends through the center of the acetabular cup. Finally, the x-axis was created using the cross product of vectors along the x- and y-axis (Figure 3).

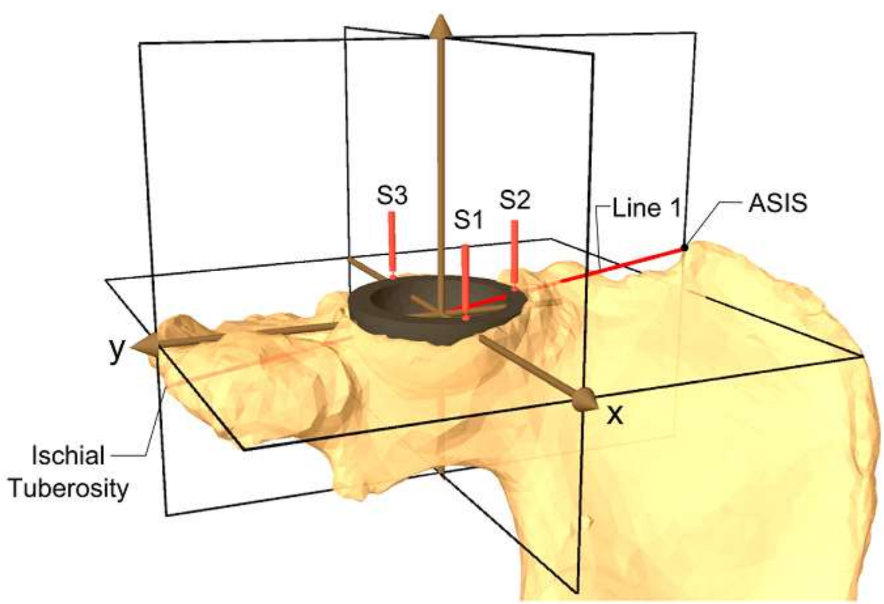

Fig. 3. Reference System adopted with Origin in the cup center, Y axis defined as the projection on the plane of the cup of the line joining the ASIS and the Ischial tuberosity and XY plane containing the cup edge.

Bergman [40] reported peack contact values ranging from 2.42 to 2.50 BW for slow and fast walking speed, for our study we used the peack forces observed during fast walking and assumed a body weight of 61 Kg to correspond to our speciment. A loading electromechanical system (Instron Model 5569, Instron Corp., Canton, MA) was used to apply a force from 0 N to 1500 N in increments of 100 N at a speed of 5mm/min for a total of 5 cycles along the z-axis with a moment arm along the x-axis of 30 mm to mimic immediate post-operative loading conditions. Data was collected using AC gauging linear variable differential transformers (LVDT) (333 Miniature AC gauging LVDT sensors, Trans-tek, Ellington, CT). Sensors 1, 2, and 3 were placed in contact with the rim of the acetabular cup perpendicular to the $\overline{XY}$ plane to record micromotion along the z-axis [41]. Micromotion recorded by the LVDTs was transmitted to a data acquisition system designed in LabVIEW Virtual Instrument (VI) to register and record the micromotion registered in real time.

2.2 Patient-Specific Finite Element Model Development

Diagnostic images were acquired with CT scanning using a BrightSpeed (GE Medical Systems) scanner (slice thickness of 0.625 mm, pixel size of 0.422 mm, field view of 216 mm); images were taken of the complete cadaveric pelvis prior to cup implantation in order to develop a patient-specific 3D reconstruction of the intact hemi-pelvis geometry, unaltered by the presence of the titanium acetabular cup. The 3D reconstructions of the five hemi-pelvis were built using the segmentation tools of the Mimics Suite (Materialise, Leuven, Belgium) and material property segmentation was conducted using the local bone mineral density between cortical and subchondral bone.

For each modeled pelvis, using a tridimensional reconstruction, we have measured five morphological parameters reported in Table 1. The morphological data of the sample with the closest values to the calculated average has been selected to create the model used for the proposed Input Parameter Variation Study.

Of the selected sample, an additional CT scan was performed after biomechanical testing to create a 3D reconstruction of the final geometry of the bone and cup. A fitting procedure was conducted to match the peripheral surfaces of the bone between the two models. Assuming that the acetabular floor of the model that was reconstructed from the tested specimen was unaltered by cup insertion, a best fit sphere was created using points located in this region, and used to under-ream the acetabulum of the intact model (Figure 4).

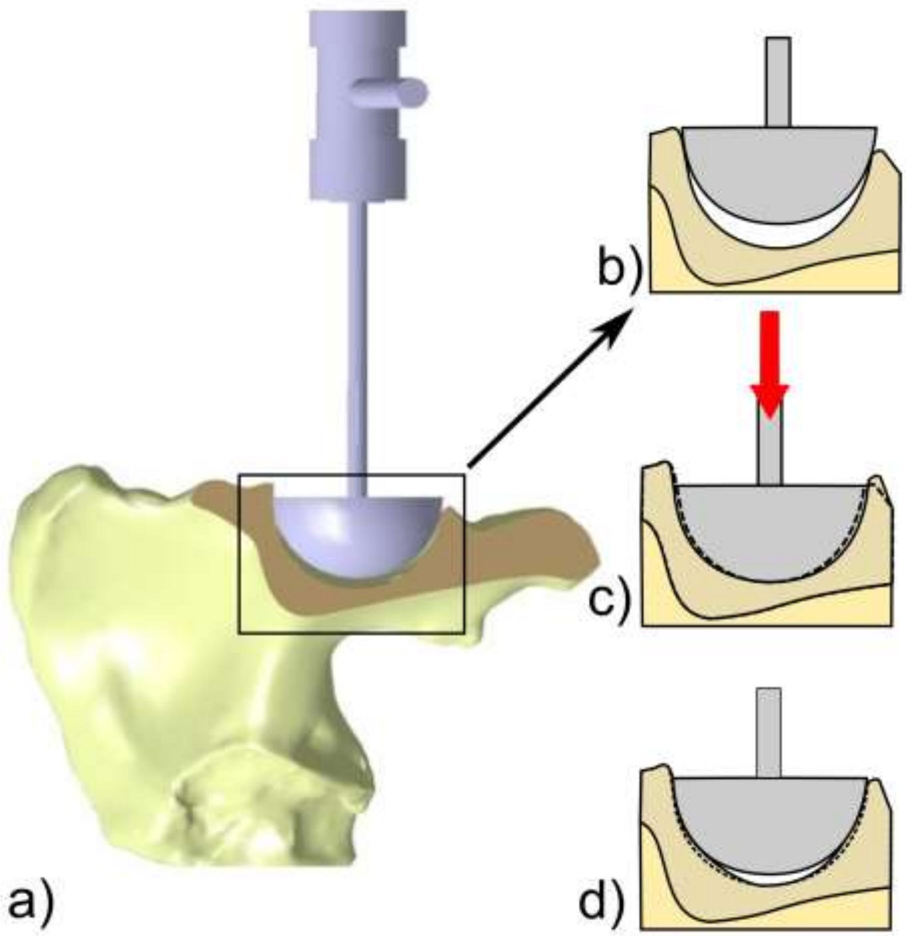

Fig. 4. Positioning and hammering steps in THA: a) Rendering of a Cup-bone interface and cross section of pelvis wall; b) Initial setting before hammering; c) Hammering and bottoming of the cup; d) Bouncing back at equilibrium position.

The relative position of the cup prior to testing and after cup implantation in relation to the unaltered reconstruction was obtained by relocating the cup to a position dictated by the values obtained from the LVDT sensors, taking into account the permanent deformation recorded at the end of the experiment. The equilibrium position of the cup at this stage was defined as the location of the cup after insertion but before any post-operative load had been applied. The polar gap distance between the apex of the cup and the floor of the reamed acetabulum was measured on the reconstructed model and determined to be 0.681 mm during cup equilibrium (Figure 5). A possible hammering distance was achieved by moving the cup along a line of action perpendicular to the plane of the cup rim until minimal contact was achieved between the cup and acetabular wall. The initial polar gap distance before implantation between the apex of the cup and the floor of the reamed acetabulum was 1.606 mm.

The resulting 10,940 tetrahedral solid (SOLID72) elements (3,466-node) model of the hemi-pelvis, and 3,466-element (865-node) model of the acetabular cup was imported into ANSYS 13.0 (Ansys Inc., Canonsburg, PA). All materials were assumed to be linear elastic with a Poisson's ratio of 0.3 and values of Young Modulus of 0.07 GPa for cancellous bone adopted from Barreto et al, [42] and assigned the cortical bone Young Modulus using the power relationship used by Taddei for the femur [43] with imposed value of 17 GPa for the highest values of Hounsfield Unit found. The average value of young modulus for cortical bone was of 13.8 GPa with a lower limit of 11.13 GPa. The metal alloy cup (Ti-6Al-4V) had an elastic modulus of 110 GPa with a Poisson's ratio of 0.3 [44]. A nonlinear, asymmetric, frictional, surface-to-surface contact interface was created between the bone and cup implant with 784 elements (CONTA 173) and 598 target elements (TARGET 170). An augmented Lagrange method was chosen to solve the contact model, with a coefficient of friction of 0.5 [45].

2.3 Parameter Fitting Characterization and Model Validation

A parameter fitting method was utilized to optimize the imposed hammering distance during insertion for the model such that it accurately represented the in vitro experiment. Input values of imposed hammering distances to the cup were sequentially varied to achieve the equilibrium polar gap distance from the in vitro experiment. The iliac crest and a portion of the ilium were constrained in all directions to mimic the conditions of the mechanical testing and to avoid translational movement of the bone, as shown in Figure 5. The simulation was done in two load step phases, the first of which mimicked cup insertion/hammering due to imposed hammering distance. The boundary conditions on the cup were such that it was allowed freedom of movement in only the direction it was to be displaced. In the second phase, all boundary conditions were removed from the cup, at which point the cup was able to rebound and adjust to an optimal conformity, finding the equilibrium position.

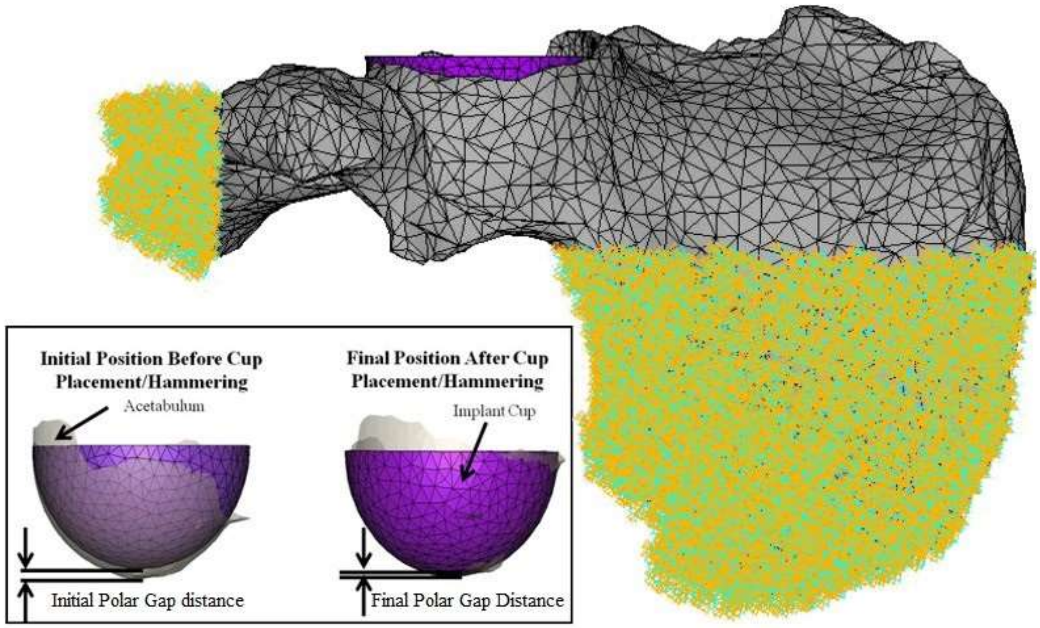

Fig. 5. Finite element model of the hemi-pelvis with constraints at the pubic symphysis and the sacro-iliac joint. Boundary constraints of the hemi-pelvis with the ilium and a portion of the ischium fully constrained and the gap distance of cup-bone interface before and after cup placement shown.

The location of the cup at the end of the second load step was regarded as the equilibrium position of the cup following press-fit implantation. The mathematical model was run, varying the imposed hammering distance until the predicted polar gap distance matched the experimental polar gap distance with less than 2.0% error. The polar gap distance was measured as the distance along the z-axis between a pair of nodes located at the apex of the cup and the floor of the reamed acetabulum. The imposed hammering distance to match the experimental results within 2% was 1.55 mm, which achieved a polar gap distance of 0.691 mm.

Following parameter fitting of the imposed hammering distance to accurately predict the equilibrium polar gap distance, the model was validated using displacement values of selected nodes located in the same position as the sensors from the in vitro experiment. The reference system defined was used to select the appropriate nodes. Validation was conducted in three load step phases. The first two load steps mimicked the insertion and equilibrium phases, respectively. In the third and final load step, a compressive load from 0 N to 1500 N along the z-axis and a moment arm of 30.38 mm was placed on the cup, as in the in vitro experiment. Throughout loading, the predicted values of micromotion matched the actual values with about 12% error or less. Furthermore, the final values of cup displacement at 1500 N of the three nodal points representing sensors 1, 2 and 3 were predicted with a percent error of 11%, 12%, and 2%, respectively (Figure 6).

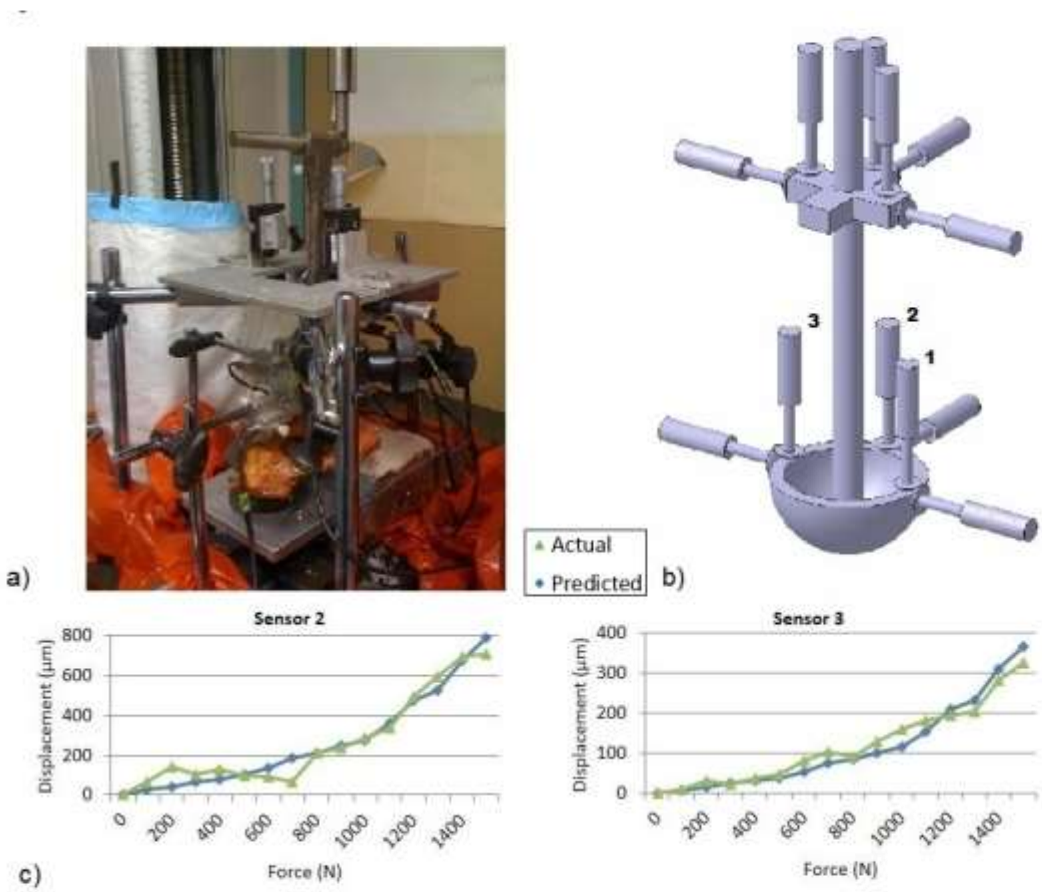

Fig.6. THA Experiment Setup: a) execution; b) Layout of the sensors adopted c) Comparison of predicted and experimental values from 0 to 1500 N.

2.4 Input Parameter Variation

Following model validation, we evaluated the influence of varying input parameters and the effect on stresses, surface conformity, polar gap distance, and micromotion at the interface. Three load steps were used to mimic the press-fit insertion, equilibrium, and loading phases. The nodes at the pubic symphysis and the sacro-iliac joint were constrained in all degrees of freedom to simulate in vivo conditions at each step. Various configurations were run as the first load step to account for 1 mm and 2 mm of under-reaming for press-fit insertion with varying hammering distance depths in order to achieve different amounts of contact at the cup-bone interface. During the second load step, constraints were removed from the cup to allow for equilibrium, and during the third load step, a force of 1500 N was applied to the cup to mimic post-operative loading.

Results

The distribution of von Mises stress throughout the entire hemi-pelvis was predicted following press-fit insertion of the cup for various configurations. High stresses were predicted in the superior region of the acetabular wall as well as a portion of the ischium.

Following the first phase of cup insertion, the insertion force of the cup was evaluated to observe the influence of the force needed to insert under-reamed cups set to varying target locations, as specified by the hammering distance imposed on each cup. The aim was to determine in which configuration a minimal force can be used to establish a stable contact at the cup-bone interface. The results show that insertion force increases as a function of the amount of under-reaming, as well as increasing target insertion distances Figure 7.

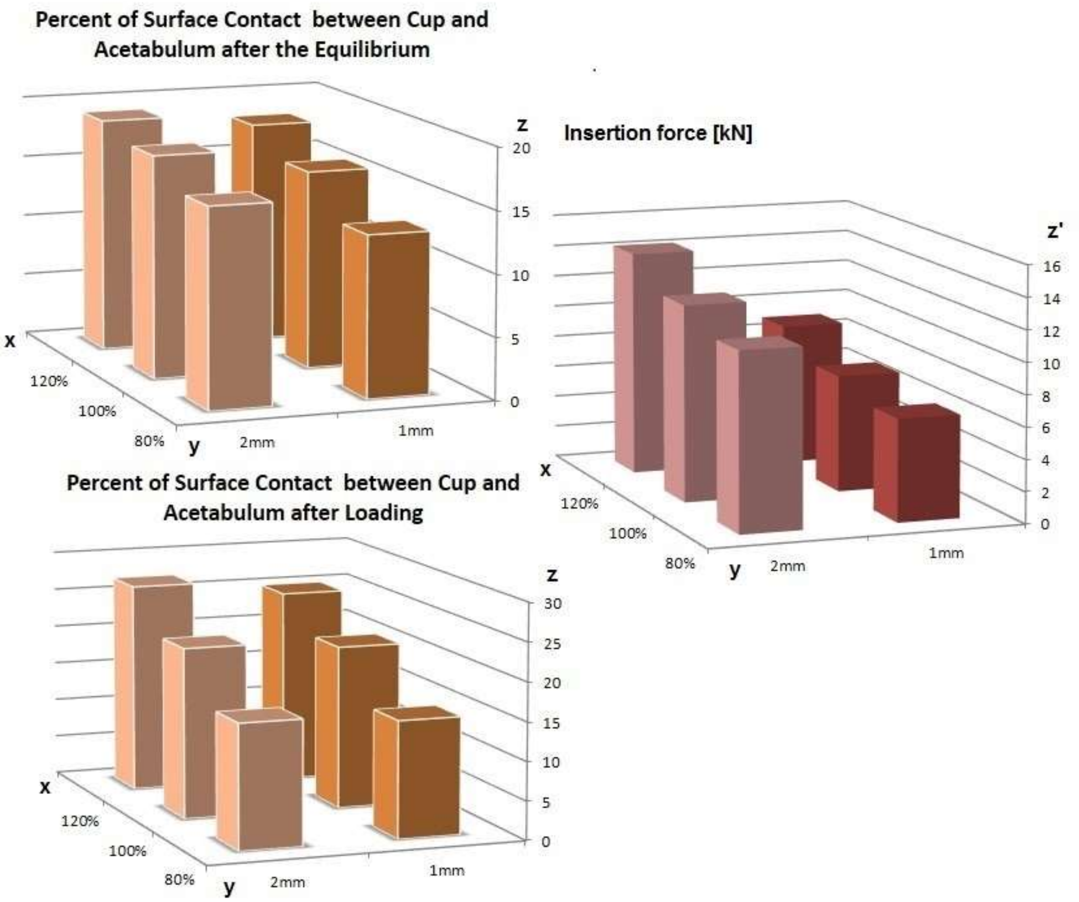

Fig. 7. Percent of contact surface at cup-bone interface (z) before and after cup insertion and Insertion force (z') as function of x (hammering distance expressed as percentage of the initial polar gap distance) and y (value of under-reaming).

In order to evaluate the surface conformity at the cup-bone interface, the percentage of nodes with a contact penetration depth of less than 0.1E-6 mm at the interface was calculated at the end of each load step. Despite the degree of under-reaming, the hammering distance and therefore the insertion force needed to press-fit the cup had a greater influence on the amount of surface conformity at the interface. Under-reaming by 2 mm achieved only a slight increase in good surface conformity immediately following cup insertion. An adjustment of the conformity of the peripheral bone of the acetabular wall due to over-sized cup insertion was observed, while the bone at the floor of the reamed acetabulum remained largely unaffected by cup insertion. Gap distances at the apex of the cup and the nearest point of the floor of the reamed acetabulum were also compared and were consistent with the results of clinical cases, in which polar gap distances of 0.5 mm were reported following the implantation of press-fit cups. Larger polar gap distances were noted for configurations with a greater amount of under-reaming. A comparison of three imposed hammering scenarios between the 1 and 2 mm under-reaming conditions is shown in Figure 7 as function of percentage of surface in contact before and after the equilibrium phase and insertion force.

For 1 mm of under-reaming the percentage of cup surface in contact with the bone varied from 14.5 to 25.1% for the imposed target distances These values were reduced to 12.9 and 18.13% after the equilibrium phase was reached.

For the 2 mm under-reaming the percentages of surface in contact were slightly greater with values ranging from 16.2 to 26.9% after loading and 15.8 to 19.35% after the cup reached the equilibrium phase. Between the first and second phases of cup insertion, the percentage of surface in contact decreased by an average of 3% for all configurations. For each configuration, higher stresses were present on the peripheral rim of the acetabular cup and bone (Figure 8), with the highest stresses noted at the superior region of the acetabular wall.

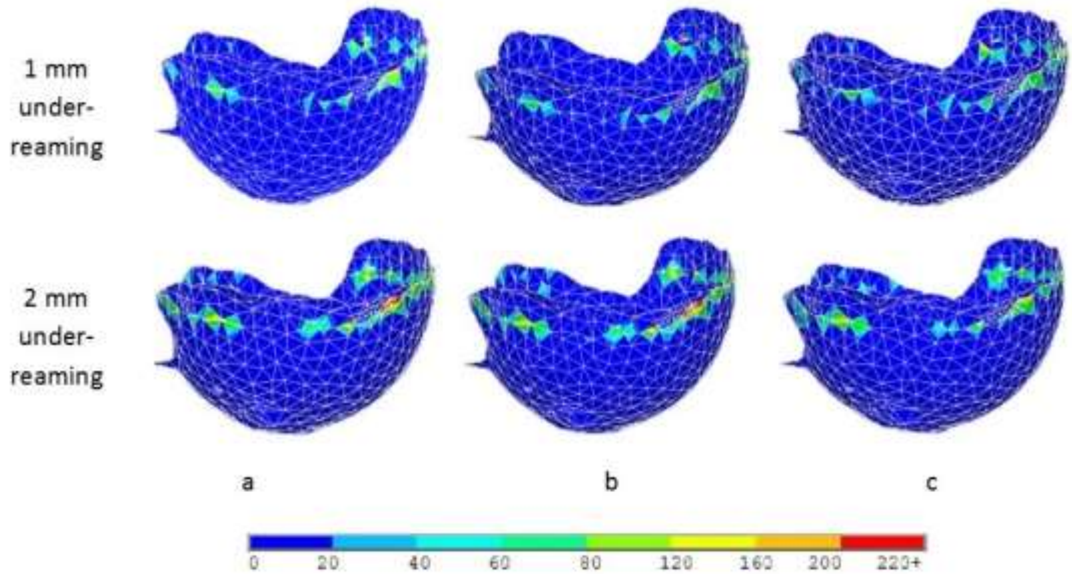

Fig. 8. Total stress (MPa) on the acetabular wall at the cup-bone interface for 1 and 2 mm of under-reaming during (a) 80% (b)100%, and (c) 120% of initial gap distance.

In the cases presented with 1 mm of under-reaming, 95% of elements at the interface experienced a stress less than 20 MPa, with no elements exceeding a stress of 220 MPa. Under-reaming by 2 mm resulted in 92% of stress less than 20 MPa, with 0.17% of elements depicting a peak stress greater than 220 MPa. Because 95% of the stress varies between 0 and 55 MPa at the interface, we conclude that the contact at the interface is nearly homogenous (Table 2). In addition to a decrease in surface conformity during the equilibrium phase, the gap distance at the apex of the cup almost doubled, despite the hammering force or decrease in under-reaming. Up to a loading of 1500 N, the surface of contact decreased by less than 1% and the gap distance at the apex of the cup decrreased by 2% providing further evidence that the cup- bone interface is stable during the loading phase (Figure 9).

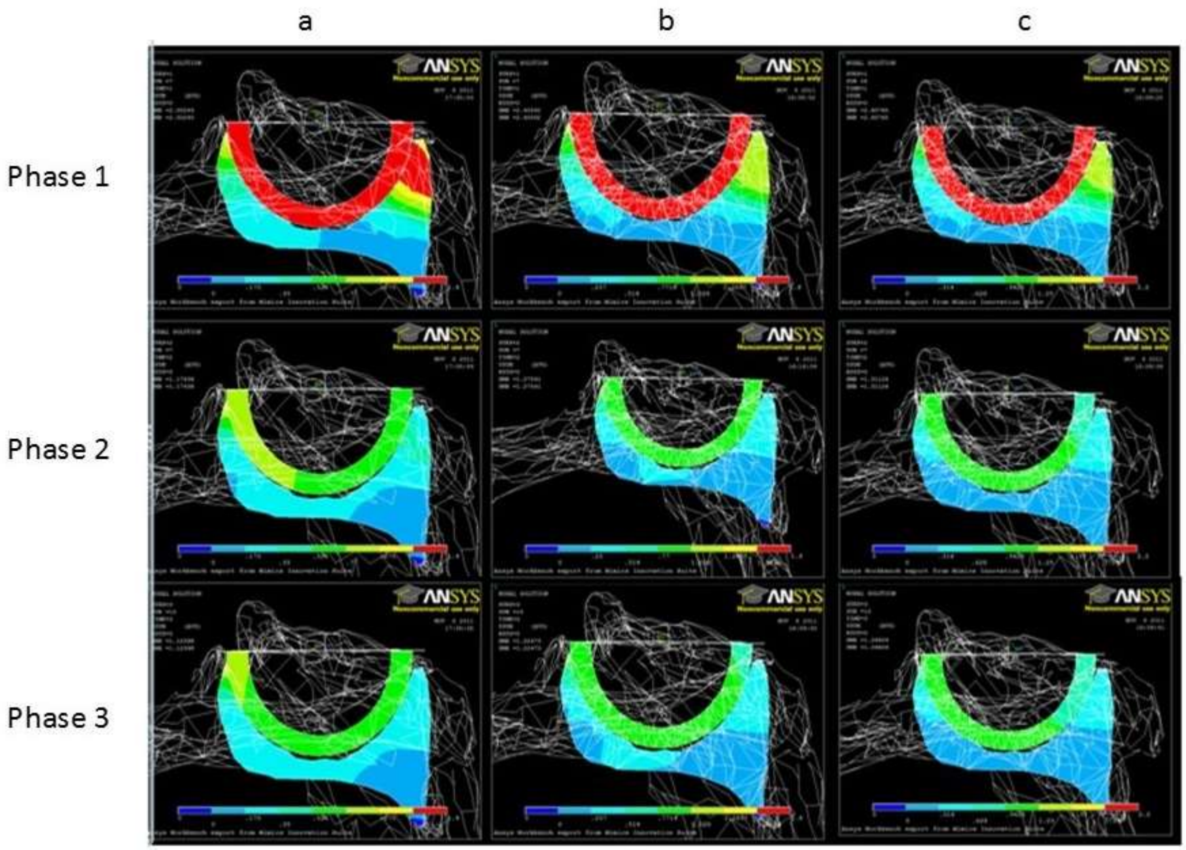

Fig. 9. Cross section view of the acetabular cup with 2mm of undereaming during the three loading phases for three imposed hammering distances a,b and c of 80,100 and 120% of intial gap distance.

The criterion for surface conformity at the interface was evaluated as a function of the percentage of surface contact at the interface, as well as the displacement of the cup due to loading. Insertion force was evaluated using the reaction force on the acetabular cup due to insertion by the imposed hammering distance. The results indicate that force needed to insert the acetabular cup increased as a function of increasing percentage of surface in contact before loading, as well as under-reaming (Figure 10).

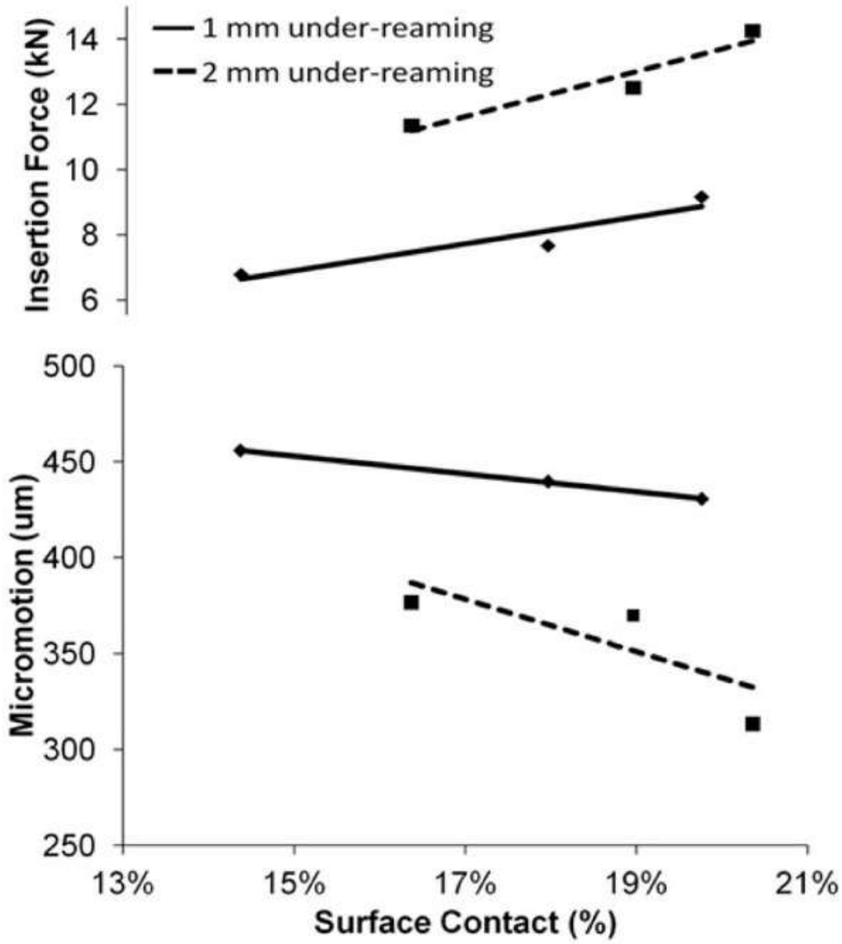

Fig. 10. Relation between insertion force, displacement, and percentage of surface contact at 1500 N of loading for 1 and 2 mm of under-reaming.

For the 1 mm of under-reaming 14.4% of contact was achieved with an insertion force increase of 19.7% going from 6.8 kN to 9.15 kN. On the other hand a 2mm under-reaming showed an increase from 16.3% to 20.1% with a change in insertion force from 11.3 to 14.3 kN. Conversely, displacement of the cup decreased as a function of increasing surface contact and a greater amount of under-reaming. While the movement of the cup remained relatively stable for 1 mm of under-reaming, ranging from 430 to 455 μm, a sharp decrease in displacement occurred from 378 to 313 μm for the case of 2 mm of under-reaming.

Discussion

Some of the factors influencing THR can be seen in a recent registry study of 35,140 THRs, where women had a 29% higher risk of implant failure than men in a 3-year follow-up period [46]. Researchers have found that, men device survival was significantly higher than in women, and women had a significantly higher risk of all-cause revision, aseptic revision, and septic revision. This gender disparity was postulated to be associated with differences in muscle mass and soft-tissue properties, which are related to genetic and hormonal differences between the sexes [46]. We speculate that the initial positioning of the cup and its subsequent loading conditions may lead to different stress conditions in men and women, which then determine THR outcome and the need for revision.

In this study, we determined the influence of cup insertion factors, particularly the amount of under-reaming and insertion force, on initial cup stability during THR. The roles of various parameters on the behavior and stability of the cup were varied and evaluated, and each had a strong influence on the stability and conformity of the cup-bone interface following cup insertion and loading.

The force used to insert the acetabular cup for various imposed hammering distances was determined for two different degrees of bone under-reaming. The average insertion force was approximately 10.4 kN±3.9kN, which is on the same order of magnitude as reported in the literature [47]. Considering that the differences in recorded insertion force can be attributed to differences in loading parameters, the forces needed to insert the cup that were predicted in this study fell within the same range of magnitude as those reported in literature.

The degree of under-reaming and insertion force used for cup placement plays a significant role in the amount of surface contact and micromotion of the cup at the interface. Increased surface contact, measured by evaluating the number of nodes in contact at the interface, showed a positive correlation with increasing levels of under-reaming, as well as increasing force needed for cup insertion. This phenomenon was visible throughout loading, and was more pronounced with 2 mm of under-reaming, which showed a sharper decrease in micromotion with increasing insertion force. However, it is interesting to note that approximately the same percent of surface contact can be achieved with 1 mm of under-reaming and less insertion force. For example, under-reaming by 2 mm requires an insertion force 22.5% greater than the force needed to seat a cup with 1 mm under-reaming, but achieves only a 13% reduction in micromotion with nearly the same surface contact.

Conclusion

Our analysis of the polar gap observed following cup insertion has important clinical relevance. The occurrence of such gaps is difficult to observe in practice, but can be easily investigated with the tools of finite element analysis. While a greater insertion force used during cup insertion resulted in a slightly greater amount of surface conformity at the interface, a polar gap was still present for both cases of under-reaming despite the insertion force used. Furthermore, because a very high insertion force would be needed to achieve full surface conformity, the stresses on the bone are likely to increase as well, which may result in fracture of the acetabular wall.

The data show a significant correlation between the degree of under-reaming and the resulting total stress at the cup-bone interface, as well as the distribution of von Mises stress throughout the entire bone. However, while increased under-reaming results in higher stresses, especially at the interface, it is clear that greater under-reaming also results in a more rigid fixation between the cup and bone. The results suggest that because greater than 90% of elements predicted a stress ranging from 0 to 55 MPa, with just a small region of higher predicted stress exceeding the elastic range of bone, fracture of the acetabular wall will not likely occur during cup insertion or loading. It is important to note that because the stresses seen for 2 mm of under-reaming approached the yield strength for bone in some regions, caution should be taken during cup insertion to avoid fracture of the acetabular wall, especially for higher amounts of under-reaming.

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FIGURE LEGENDS

Fig. 1. Dislocation of the THA due to a vertically placed acetabulum component B. Acetabular shell maintained converted into a constrained liner.

Fig. 2. Dislocation of the hip with failure of constrained liner (broken locking ring) B. Final revision acetabular shell placed in anatomic position with unconstrained liner (the constrained liner didn’t address the pathology of the initial placed vertical cup).

Fig. 3. Reference System adopted with Origin in the cup center, Y axis defined as the projection on the plane of the cup of the line joining the ASIS and the Ischial tuberosity and XY plane containing the cup edge.

Fig. 4. Positioning and hammering steps in THA: a) Rendering of a Cup-bone interface and cross section of pelvis wall; b) Initial setting before hammering; c) Hammering and bottoming of the cup; d) Bouncing back at equilibrium position.

Fig. 5. Finite element model of the hemi-pelvis with constraints at the pubic symphysis and the sacro-iliac joint. Boundary constraints of the hemi-pelvis with the ilium and a portion of the ischium fully constrained and the gap distance of cup-bone interface before and after cup placement shown.

Fig.6. THA Experiment Setup: a) execution; b) Layout of the sensors adopted c) Comparison of predicted and experimental values from 0 to 1500 N.

Fig. 7. Percent of contact surface at cup-bone interface (z) before and after cup insertion and Insertion force (z’) as function of x (hammering distance expressed as percentage of the initial polar gap distance) and y (value of under-reaming).

Fig. 8. Total stress (MPa) on the acetabular wall at the cup-bone interface for 1 and 2 mm of under-reaming during (a) 80% (b)100%, and (c) 120% of initial gap distance.

Fig. 9. Cross section view of the acetabular cup with 2mm of undereaming during the three loading phases for three imposed hammering distances a,b and c of 80,100 and 120% of intial gap distance.

Fig. 10. Relation between insertion force, displacement, and percentage of surface contact at 1500 N of loading for 1 and 2 mm of under-reaming

TABLES

Table 1. Details of the specimens investigated

age	sex	Diameters [mm]			Distance PSIS-ASIS [mm]
		head of femur	acetabulum	cup	
85	male	47.8	53.0	56	163.9
73	male	54.4	57.9	64	177.6
85	female	38.6	45.5	50	160.3
69	male	46.9	53.7	60	164.4
85	female	46.1	52.4	56	164.7
na	na	40.3	46.9	50	154.3
averages		45.7	51.6	56	162.9

Table 2. Percentage of elements predicting a range of Total stress values at the cup-bone interface for 1 and 2 mm of under-reaming during cup insertion.

	Imposed Target Hammering Distance					
Total Stress (MPa)	80% of Initial Gap	Initial Gap	120% of Initial Gap	80% of Initial Gap	Initial Gap	120% of Initial Gap
0-55	94.94%	94.43%	93.89%	89.66%	89.06%	88.48%
56-110	3.35%	3.57%	3.84%	5.88%	6.26%	6.49%
111-165	0.95%	1.19%	1.38%	2.07%	2.18%	2.26%
166-220	0.34%	0.32%	0.34%	1.20%	1.24%	1.42%
220+	0.42%	0.49%	0.55%	1.19%	1.26%	1.35%
	1 mm under reaming			2 mm under reaming		