

Deformation of press-fitted metallic resurfacing cups. Part 1: experimental simulation

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Abstract: The interference press fit of a metallic one-piece acetabular cup employed for metal-on-metal hip resurfacing procedures was investigated experimentally under laboratory conditions in the present study, in particular regarding the cup deformation. Tests were carried out in cadavers as well as polyurethane foams of various grades with different elastic moduli to represent different cancellous bone qualities. The cadaver test was used to establish the most suitable configuration of the foam model representing realistic support and geometrical conditions at the pelvis. It was found that a spherical cavity, with two identical areas relieved on opposite sides, was capable of creating a two-point pinching action of the ischeal and ilial columns on the cup as the worst-case scenario. Furthermore, the cup deformation produced from such a two-point loading model with a grade 30 foam was similar to that measured from the cadaver test. Therefore, such a protocol was employed in subsequent experimental tests. For a given size of the outside diameter of the cup of 60 mm, the cup deflection was shown to be dependent largely on the cup wall thickness and the diametral interference between cup and prepared cavity at implantation. For a relatively thin cup with a wall thickness between 2.3 mm (equator) and 4 mm (pole) and with a modest nominal diametral interference of 1 mm, which corresponds to an actual interference of approximately 0.5 mm, the maximum diametral cup deflection (at the rim) was around 60 μm , compared with a diametral clearance of 80–120 μm between the femoral head and the acetabular cup, generally required for fluid-film lubrication and tribological performances. Stiffening of the cup, by both thickening and lateralizing by 1 mm, reduced the cup deformation to between 30 and 50 μm with actual diametral interferences between 0.5 and 1 mm.

Keywords: metal-on-metal articulations, hip resurface, cup deformation

1 INTRODUCTION

The hip resurfacing procedure employing metal-on-metal (MOM) articulations has received significant attention recently, particularly in the UK as recommended by the National Institute for Clinical Excellence (19 June 2002) as one of the options for younger, more active, and more demanding patients. Hip resurfacing procedures have been considered to be attractive owing to perceived advantages such as bone conservation, more physiological loading, increased range of motion, and joint stability [1–4].

Poor results associated with early hip resurfacing procedures caused the concept to be largely abandoned during the 1970s and 1980s. These failures were primarily attributed to problems associated with the choice of the bearing material combination, rather than with the resurfacing concept itself [5, 6]. The material combinations for these early resurfacing designs were polyethylene cups articulating with metallic femoral heads. This caused problems of polyethylene-wear-induced osteolysis, significantly aggravated by the use of the large-sized component and the corresponding increased sliding distance [7].

The renewed interest in hip resurfacing procedures is in part due to the successful reintroduction of MOM articulations. The combination of intrinsically

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low-wear cobalt–chromium alloys and improved lubrication at the articulation have led to the fact that exceedingly low wear is found in conventional total hip replacements employing MOM material combinations [8]. Such an advantage in terms of wear reduction can be significantly enhanced with the large-sized hip resurfacing components, owing to the improved tribological conditions at the articulating surfaces, such as the increased sliding velocity, the reduced contact stress, and the increased contact area, all of which should promote fluid-film lubrication and hence reduce wear further.

Successful tribological performances of MOM total hip replacements also rely on the design and manufacturing parameters in ensuring low wear, including minimal clearance between the femoral head and acetabular cup, smooth bearing surfaces, and sphericity. However, the effects of different cobalt–chromium alloys on wear have been found to be negligible, provided that high-carbon materials are used [9]. The increased size associated with hip resurfacing prostheses means that the wall thickness of the acetabular cup must be reduced. From a tribological point of view, a thinner acetabular component is able to deform more, therefore spreading the load and potentially promoting fluid-film lubrication. From a bone conservation point of view, a thinner acetabular component is also desirable since less bone stock has to be removed. However, adverse problems potentially associated with thin cups must also be recognized. For example, excessive deformation of the cup is accompanied by an increase in the micromotion at the bone and cup interface, not only affecting bone ingrowth but also resulting in fretting wear. Significant deformation may lead to equatorial and edge contact, not only elevating stresses, but also blocking the lubricant entry and adversely affecting fluid-film lubrication. It is important to recognize that the metallic resurfacing cup is usually press fitted and inserted into the acetabulum as an interference fit. As a result, a significant deformation of thin cups may be possible, leading to changes in the bearing geometries such as the clearance and sphericity, which can adversely affect fluid-film lubrication and increase wear, and in extreme cases can eliminate the clearance and cause the joint to jam.

There have been a number of experimental and clinical studies on the press-fit mechanism of uncemented cups reported in the literature [10–15], but all with references to conventional total hip replacements. Difficulties in the study of press-fitted cups under clinical settings, such as supply of cadaveric pelvises and unrepresentative age and quality of

specimens, have led to the extensive development of laboratory models, mainly using polyurethane foams. The focus of both clinical and laboratory studies has been usually on the initial stability at the interface between the bone and cup, and the important parameters studied include the interfacial micromotion, contact area, and potential gaps remaining between the cup and the bone. Achieving adequate initial primary stability is important for long-term secondary stability and clinical success through biological (bone-ingrowth) fixation and osseointegration. Other studies have focused on impact biomechanics and pelvic deformation induced during the press-fit insertion of the cup.

The purpose of the present study was to investigate the effect of varying amounts of interference press-fit and cup geometry on cup deformation. The corresponding theoretical studies based on the finite element modelling are presented in Part 2 [16].

2 MATERIALS AND METHODS

2.1 Metallic resurfacing cups

A typical MOM hip resurfacing prosthesis design (ASR™, DePuy) shown in Fig. 1 was considered in the present study. The metallic acetabular cup of this couple, mainly with an outside diameter of 60 mm was investigated, as this is a midrange size. Other sizes will be considered in Part 2 [16]. One of the main design parameters of the metallic resurfacing cup is its wall thickness, and therefore an experimental prototype design with a reduced cup wall thickness was also investigated [Fig. 2(a)]. The wall



Fig. 1 A typical metal-on-metal hip resurfacing prosthesis consisting of a femoral resurfacing component (left) and a monobloc acetabular cup (right) (ASR™ DePuy)

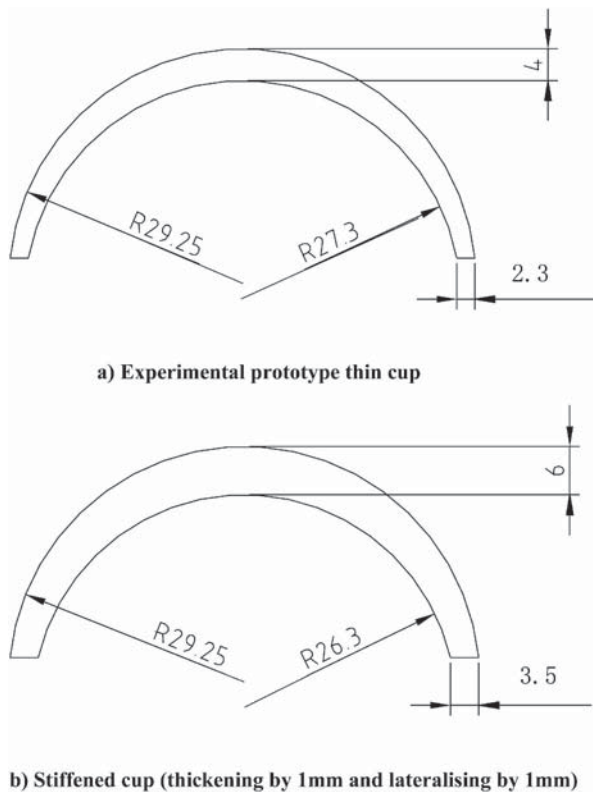


Fig. 2 Cross-sectional geometry (in millimetres, excluding the porous coating) of the three cups used in the present experimental study

thicknesses of the prototype varied between 2.3 mm (equator) and 4 mm (pole). Figure 2(b) represents a stiffened cup. Stiffening was achieved by both lateralising (moving the centre of the inner surface away from the cup by 1 mm, thus thickening the pole of the cup in preference to the edges) and by thickening of 1 mm. All the cups were made from cast high-carbon cobalt–chromium alloy, according to ASTM F75.

2.2 Test configurations

Both cadaver specimens and solid rigid polyurethane foams were used to implant the metallic resurfacing cups considered in the present study. The cadaver tests were carried out using fresh frozen hemipelvises. The age of the donors was in the 40–60 years range. Because of the limited supply of cadaver specimens, only a few of these tests were performed, while the majority of testing was performed on the polyurethane form (SAWBONES[®]), according to ASTM F-1839. Different foam grades (defined as density in pounds per cubic feet) of 15, 30, and 40 were employed to simulate different bone properties. The corresponding compressive elastic moduli, specified accordingly to ASTM, are 153, 553, and

943 MPa. Therefore, these foams should be able to represent different cancellous bone qualities [17].

Furthermore, different foam configurations were considered in order to simulate the geometry of the cadaver specimens. These included three foam cavity configurations as shown in Fig. 3: a simple block with an appropriate reamed spherical cavity, a thin walled cavity, and a spherical cavity with two areas relieved on opposite sides. The former two configurations were found to be unsatisfactory. It has been shown by Widmer *et al.* [15] that the outside of the cup under simulated one-leg stance was primarily loaded in three locations near the periphery of the acetabulum, mainly the cranial region (iliac bone, 55 per cent), the posterior-inferior region (ischial facet, 25 per cent), and the anterior region (pubic bone, 20 per cent). However, the present cadaver test of the metallic resurfacing cup showed that the cup deformation was primarily a result of squeezing between the ischeal and ilial columns. Such a deformation mode would represent the worst-case scenario and could be readily simulated by a spherical cavity, with

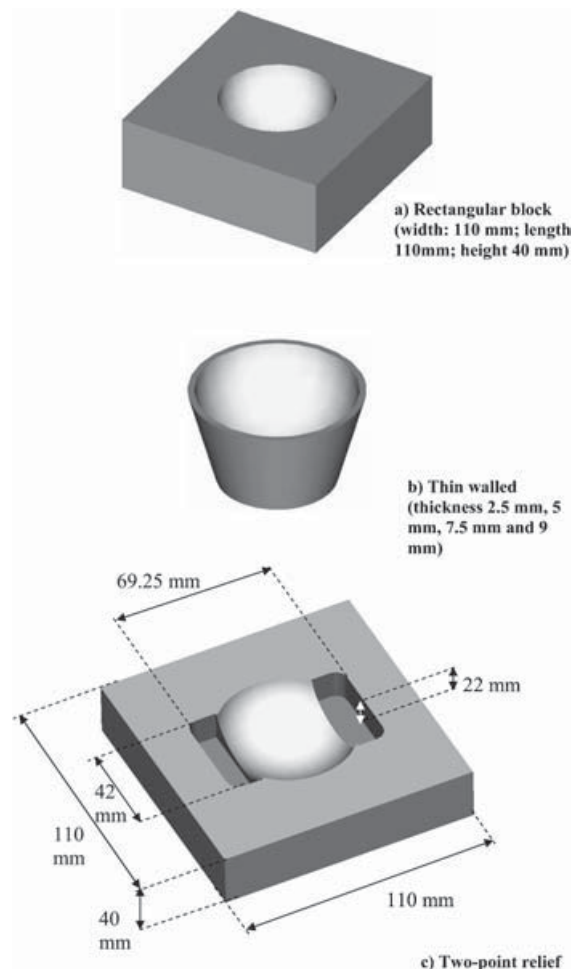


Fig. 3 Various cavity configurations for press-fit testing

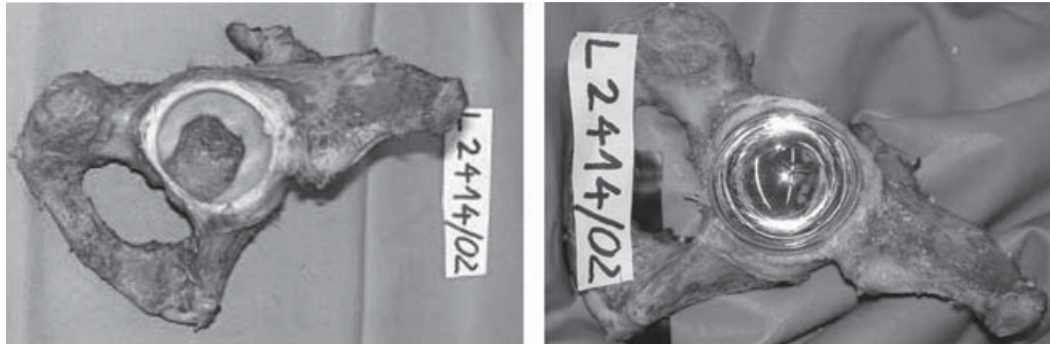


Fig. 4 Cadaveric hemipelvis (a) before implantation and (b) after implantation

two areas relieved on the opposite sides, producing a two-point pinching loading on the cup [Fig. 3(c)].

2.3 Press-fit simulation and measurement of cup deformation

For a chosen diametral interference and a fixed outside diameter of the cup, the spherical cavity was under-reamed accordingly either by hand or by machine (for foam specimens only). For the cadaver testing, the diameter of the acetabulum was measured and then prepared using standard surgical hemispherical reamers. The reamers were used sequentially, increasing in diameter until all cartilage and part of the subchondral bone plate was removed. A cup 1 mm less in diameter (1 mm of nominal interference) was then implanted as shown in Fig. 4. In some cases, the cups were then removed from the pelvis and the acetabuli further reamed to assess the effect of the degree of bone preparation and interference on cup deformation. The actual dimensions of the cavity were measured with a Mitutoyu coordinate-measuring machine (CMM) (BHN 305) as

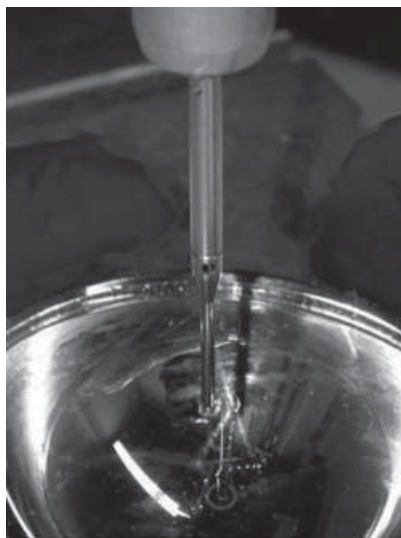


Fig. 5 Set up for the CMM

shown in Fig. 5, close to the equator in 10° (for the cadaver testing) and 45° (for the foam testing) intervals circumferentially. The cup was then impacted repeatedly into the cavity until fully seated. The cup's inner surface was mapped with the CMM before and after implantation and the radii of the impacted cup and the cup deformation were determined. The cup deformation, particularly around the equator, was analysed, and the maximum diametral cup deflection, defined as the maximum diametral deviation (out of round) from the corresponding nominal circle was calculated and compared.

Furthermore, immediately following implantation, component deformation was also assessed by placing the matching femoral head, smeared with a dye (engineers' blue), into the cup which was then rotated to see whether it articulated normally, thus allowing the area of contact between head and cup to be determined visually.

2.4 Testing

2.4.1 Pilot testing

The thin prototype cup [Fig. 2(a)] was tested initially in cadavers and the results obtained were mainly used to develop a laboratory sawbones model using the three different configurations discussed in section 2.3. Six hand-reamed foam samples were tested for a particular cup design and a given nominal diametral interference. Both the hand-reamed and the machined foam cavities were considered.

2.4.2 Final testing

Following the pilot testing, the two-point relief configuration in the sawbones model was used subsequently for the comprehensive testing of the stiffened cup design [Fig. 2(b)]. Only machined foam cavities were considered and one sample was used for each case over a wide range of nominal diametral interferences. The results were then finally validated using cadaveric tests as described in section 2.2.

3 RESULTS

3.1 Pilot testing

Implantation of the thin-walled cup [Fig. 2(a)] in cadaveric bone produced a pinching effect caused by the squeezing between the ischeal and ilial columns (Fig. 6); the unimplanted device showed a deviation of $6.8\ \mu\text{m}$ from spherical which grew to $63.4\ \mu\text{m}$ owing to implantation. The results of the initial testing using the thin components are given in Table 1. These showed that with 1 mm of nominal under-reaming (as generally recommended for this style of acetabular cup) the deformation varied between 25 and $103\ \mu\text{m}$. At the medium ($61\ \mu\text{m}$) level of deformation a head with a nominal mismatch of $100\ \mu\text{m}$ articulated freely without jamming in the cup and the dispersion of the dye indicated dome contact. At the higher ($103\ \mu\text{m}$) level of deformation the cup did not articulate properly and there was no dome contact. The intermediate ($75\ \mu\text{m}$) level of deformation articulated normally, but there was no apparent contact at the dome, indicating that this was close to the maximum permissible level of deformation for normal articulation in this bearing system.

The tests using the synthetic material on rectangular blocks [Fig. 3(a)] made from polyurethane foam grade 15 with the thin cup [Fig. 2(a)] with a 2 mm nominal diametral interference yielded a maximum diametral cup deflection of $80\ \mu\text{m}$ after implantation from the original maximum diametral derivation of $21\ \mu\text{m}$ before implantation [Fig. 7(b)]. Figure 7(a) shows the coordinates close to the equator of the reamed foam cavity. The diameter of the impacted cup shown in Fig. 7(b) was found to be the same as the original cup, $54.59\ \text{mm}$. The results of the tests on grade 40 foam are shown in Fig. 8. The diameter of the impacted cup decreased from 54.59 to

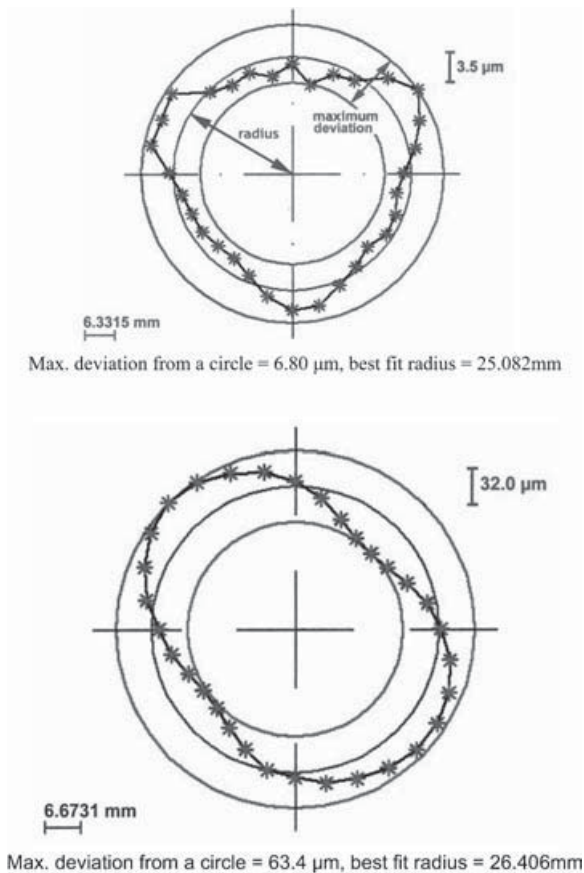


Fig. 6 Mitutoyo measurement plot of the circumferential geometry of the thin cup measured about 4 mm below the rim (a) before implantation and (b) after implantation in the cadaver (1 mm nominal undersize). The scales in each plot are adapted to the measurements. Each plot has two different scales: one scale is based on the best-fit radius, and the other scale on the maximum deviation. The deviation scale is greatly enlarged in order to highlight the deviations from a circle

Table 1 Deformation of the thin cup design measured in cadaveric bone with different amounts of nominal under-reaming

Sample	Cup outside diameter (mm)	Cup deformation (μm) for following nominal diametral under-reams		Comment
		1 mm	0 mm	
1	60	61		Blue at dome; articulated
2	58	103		No blue at dome; did not articulate
2	60		56	Blue at dome; articulated
3	56	75		No blue at dome; articulated
3	58	*		Blue at dome; articulated
4	60	25		Articulated [†]
5	60	63		Articulated

* Unique deformation signature.

[†] No measurements taken.

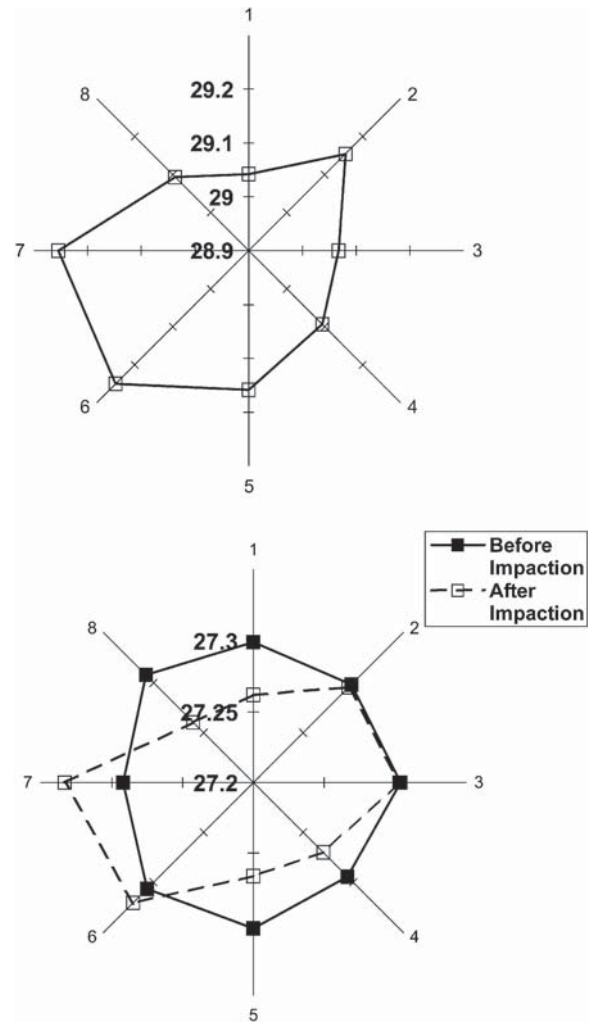
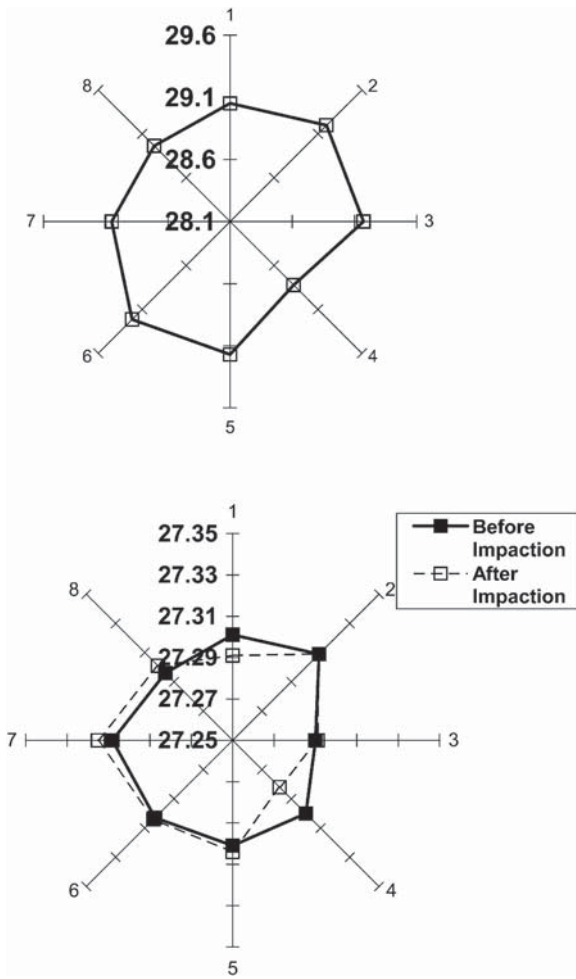


Fig. 7 Cavity preparation and deformation results for grade 15 rectangular foam blocks with a nominal diametral interference of 2 mm: (a) the radii (in millimetres) close to the equator of the reamed cavity; (b) the radii (in millimetres) close to the equator of the thin cup before and after impact

Fig 8 Cavity preparation and deformation results for grade 40 rectangular foam blocks with a nominal diametral interference of 2 mm: (a) the radii (in millimetres) close to the equator of the reamed cavity; (b) the radii (in millimetres) close to the equator of the thin cup before and after impact

54.54 mm, while the maximum diametral cup deflection was increased to 96 μ m.

The thin-wall (2.5mm) cavity model [Fig. 3(b)] was tested with a cup of the same diameter and the same diametral interference. This failed for grade 15 and 40 foam, as the cavity wall was split as shown in Fig. 9. Therefore this test configuration was abandoned.

Table 2 shows the detailed comparison of the measured maximum diametral cup deflection for different grade foams and different nominal hole diameters and consequently nominal diametral interferences. It is clear from Table 2 that significant cup deformation had occurred for the thin cup and the corresponding maximum diametral cup deflection became similar to or exceeded the diametral clearances between 80 and 120 μ m generally re-

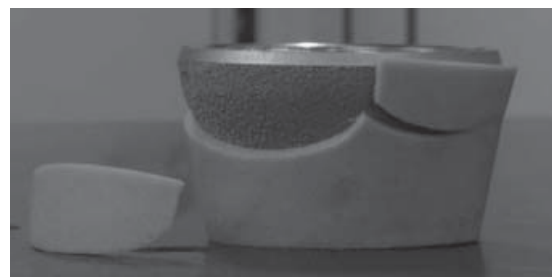


Fig. 9 Fracture of grade 40 foam (2.5 mm thick) after impact

quired for adequate tribological performances of MOM hip implants [18, 19]. Rim contact was found when the matched femoral head was brought to contact with the deformed cup. Therefore it was neces-

Table 2 Results from two-point load testing for the prototype thin cup

Foam grade	Nominal diametral under-ream (mm)	Maximum diametral cup deflection (μm)
15	2	56
15	1	22
15	1	16
30	2	Not available (foam split)
30	2	108
30	2	123
30	1	109
30	1	21
30	1	26
30	1	102
30	1	95
30	1	86
30	1	87
30	1	93
30	1	80
30	1	90

sary to stiffen the cup by thickening and lateralizing (as defined in section 2.1), as shown in Fig. 2(b). Furthermore, the measured maximum diametral cup deflection from the grade 30 foam was compared with that tested in the cadaver as shown in Table 1 and a good quantitative agreement was found. Consequently, all the subsequent experiments were carried out using only grade 30 foam and the stiffened cups.

3.2 Final testing

Figure 10 shows the maximum diametral cup deflection for various stiffened cups with a given outside diameter of 60 mm as a function of the cavity diameter, i.e. interference. The results for the original thin cup were also superimposed for the purpose of comparison. A relatively large scatter of the experimental measurements was noted in Fig. 10, particularly for

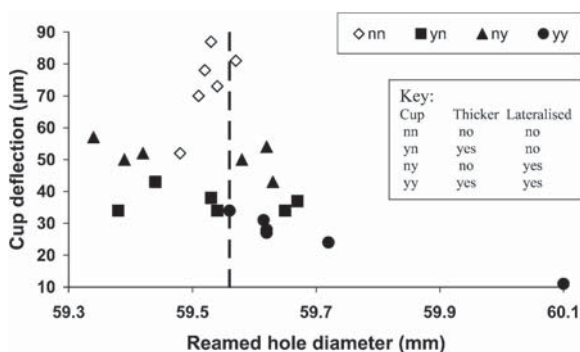


Fig. 10 Effect of the diameter of the cavity reamed by hand on the maximum diametral cup deflection for the thin and various thickened and lateralized cups tested in a two-point loading cavity model with grade 30 foam

the thin cup as shown in Fig. 2(a), although a fixed nominal cavity diameter was specified. This was mainly due to the errors resulting from hand reaming, which showed not only a large scatter in the diameter produced but also a systematic increase in the reamed diameter compared with the nominal size of the reamer. The average diameter of the reamed cavity was 59.56 mm (standard deviation, 0.15 mm) for a nominal reamer size of 59 mm. Therefore, further tests were carried out on the machined cavities and the results are shown in Figs 11(a) and (b) for the thin cup and the stiffened cup respectively.

Finally, the maximum diametral cup deflections are compared in Fig. 12 between the thin and the stiffened cups for different diametral interferences, assessed with the machined cavities.

Final validation of the effect of stiffening was carried out by implanting the lateralized and thickened cup in cadaveric material using the same protocol as the initial pilot testing. These results are given in Table 3. This shows that, in two tests (samples 4 and 5), under similar test conditions to the initial series, the increase in thickness reduced the deformation to 21 μm and 22 μm respectively. To check

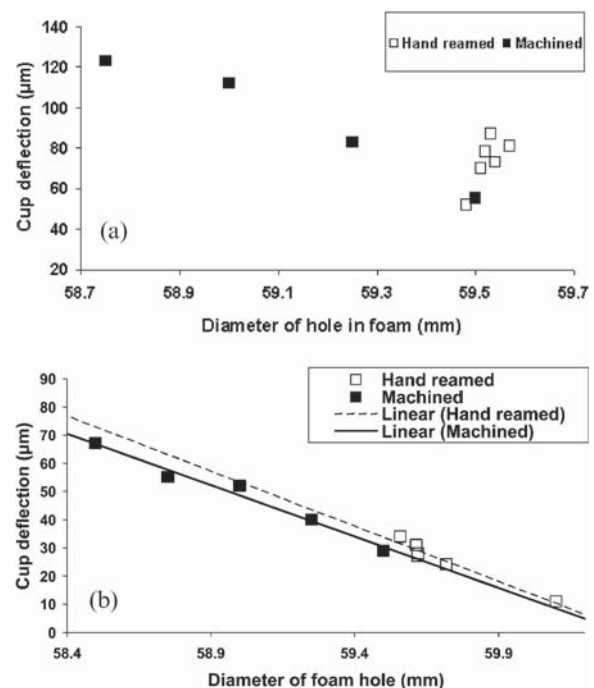


Fig. 11 Comparison of the maximum diametral cup deflection between hand-reamed and machined holes for (a) the thin cup tested in a two-point loading cavity model with grade 30 foam and (b) the stiffened cup tested in a two-point loading cavity model with grade 30 foam.

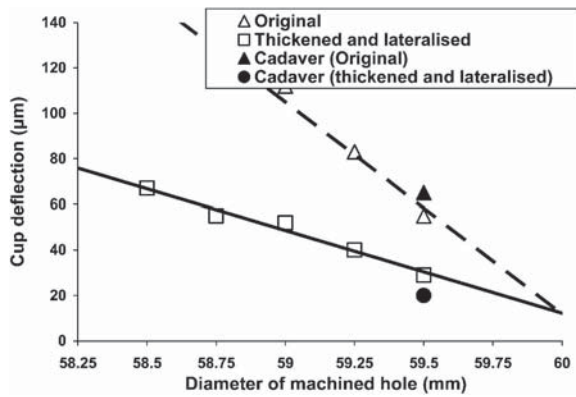


Fig. 12 Comparison of the maximum diametral cup deflection between the thin and stiffened cups tested in a machined two-point loading cavity model implanted in grade 30 foam and cadaveric bone

Table 3 Deformation of different sized stiffened cups measured in cadaveric pelvises with different amounts of nominal under-reaming

Samples	Cup outside diameter (mm)	Cup deformation (μm) for following nominal diametral under-ream			Comment
		4 mm	2 mm	1 mm	
4	60	–	–	21	Articulated
5	60	–	–	22	Articulated
6	60	–	32	–	Articulated
6	60	–	–	16	Articulated*
7	58	–	10	–	Articulated
8	58	–	8	–	Articulated
8	60	–	7	–	Articulated
8	62	8	–	–	Articulated

*Medial wall perforated.

that the reduction in deformation was real and not related to the quality of bone, the components were removed and replaced by original thin components. In one case (sample 4), the deformation was unaltered and the deformation trace did not have the characteristic two-lobed appearance, suggesting that the prepared cavity had been disrupted by the removal process. However, in the second case (sample 5), the deformation increased to $63 \mu\text{m}$, a level consistent with that reported in the initial testing. Two specimens (7 and 8) were then prepared with 2 mm nominal interference; sample 8 was re-prepared and a larger size implanted. These all produced very low levels of deformation ($7\text{--}10 \mu\text{m}$). The cup was removed from sample 8 and replaced with a cup having a 4 mm nominal diametral mismatch. This also produced a very low level of deformation ($8 \mu\text{m}$). The results from sample 6 were discarded because the medial wall of the acetabulum was perforated during the reaming process. The

overall results are summarized in Fig. 11 which consists of the foam (relieved-cavity) and cadaveric data for both thin and stiffened cups. The cadaveric data points have assumed an average actual diametral interference of approximately 0.5 mm instead of the nominal 1 mm, a value determined from measurements of the hand-reamed foam cavities.

4 DISCUSSION

The nominal diametral mismatch between the femoral and acetabular components considered in the present study was $100 \mu\text{m}$. The experimental observation of the contact between the head and the implanted thin cup in cadavers, i.e. freely and no jamming when the cup was deformed by 61 and $75 \mu\text{m}$, indicates that the maximum deformation for an effective bearing system must be about $25 \mu\text{m}$ less than the nominal clearance. This tentative conclusion must be tempered by the fact that it is based on a small number of pelvises used, that human bone is very variable, that there is one level of interference, and that one size of acetabular cup was used. One measurement was made with a cup implanted after reaming size for size (0 mm nominal interference); this surprisingly also produced significant deformation of $56 \mu\text{m}$, which might be an indication for a poorly hand-reamed non-spherical hole. Furthermore, it should be pointed out that further deformation of the cup may occur under physiological loading and the effective clearance may be further reduced. However, such an effect was not found to be significant for MOM McKee–Farrar hip implants [20].

When implanting the cups into foam the phenomenon of ‘bounce-back’ was noted and usually a number of impacts (about three) were required to seat the cup. These observations have also been reported from other experimental studies of press-fitted metal-backed cups [10, 14]. Difficulties in seating the cup were most marked for stiff foams such as grade 40 and larger interferences above 2 mm. Upon removal, the cup and foam cavity were measured again and no significant plastic deformation was observed. The occurrence of ‘bounce-back’ and associated effects such as polar gaps were beyond the scope of the present study and should be addressed in future investigations [21].

The nominally spherical socket reamed by hand when measured was not spherical and resulted in significant cup deformation. For example, using a cup of 60 mm diameter with a nominal diametral interference of 2 mm, the nominal diameter of the

cavity should be 58 mm. However, the deviations were between $-20\ \mu\text{m}$ and $+40\ \mu\text{m}$ for grade 15 foam as shown in Fig. 7(a) and between $+50\ \mu\text{m}$ and $+250\ \mu\text{m}$ for grade 40 foam in Fig. 8(a). As a result, the cup deformation was non-uniform and irregular as shown in Figs 7(b) and 8(b). The largest radial cup deformation [about $30\ \mu\text{m}$ in Fig 7(a) and about $50\ \mu\text{m}$ in Fig. 8(a)] appeared to correspond to the smallest dimension of the reamed cavity. Changing the foam from grade 15 to grade 40 (increasing the elastic modulus) increased the cup deformation; however, the reaming errors could also affect the cup deformation. The maximum diametral cup deflection of $80\text{--}96\ \mu\text{m}$ was similar to a nominal diametral clearance between 80 and $120\ \mu\text{m}$ generally required for MOM hip implants to give a satisfactory tribological performance. However, such a large cup deformation could partially result from the unrealistic rectangular block foam configuration. Therefore, thin-walled cavity models were tried, but no meaningful results were obtained because the cavity wall split and fractured.

It is clearly noted from Table 2 that the deformation of the prototype thin cup shown in Fig. 2(a), as a result of the interference press fit, is excessive, compared with the clearance required for optimal tribological conditions. Therefore, it was necessary to stiffen the cup, by both thickening and lateralizing, and perhaps to reduce the amount of interference in line with measured cavity sizes. The resultant cup deformation was significantly reduced, as shown in Fig. 10. For example, for a given cavity diameter of $59.5\ \text{mm}$, the maximum diametral cup deflection was reduced from about $80\ \mu\text{m}$ for the thin cup to $30\ \mu\text{m}$ for the stiffened cup.

The cup deformation varied significantly for a nominal diametral interference particularly for the thin cup, as shown in Fig. 10. Although a number of repeats were carried out, this was mainly associated with the reaming errors, which was not the focus of the present study. The maximum diametral deflections in thin cups were tested in hand-reamed and machined holes [Fig. 11(a)]. It is clear that the scatter of the measurement is substantially reduced for the machined cavities and an approximately linear variation in the maximum diametral cup deflection against the cavity diameter was observed. The similar comparison of the maximum diametral cup deflection for the stiffened cup is shown in Fig. 11(b). The scatter of the experimental measurements from the hand-reamed cavities is reduced for the stiffened cup when compared with the thin cup. It is also interesting to note that the cup deformation is almost linear with the diametral interference. This probably

justifies the use of a single specimen for each case in the final testing of the stiffened cups.

The diametral cup deflection when cups were implanted into the machined two-point loading models is shown in Fig. 12 and compares the thin and the stiffened cups. It is clear that, for a given diametral interference, the cup deflection for the stiffened cup was reduced by a factor of almost 2 when compared with the thin cup. Furthermore, the maximum diametral cup deflection for a given diametral interference between 0.5 and $1\ \text{mm}$ was between 30 and $50\ \mu\text{m}$, compared with the diametral clearances of between 80 and $120\ \mu\text{m}$ generally specified for MOM bearings. The deformation noted in the cadaveric study for the stiffened cup was $21\text{--}22\ \mu\text{m}$. The nominal interference was $1\ \text{mm}$ which was shown in Fig. 10 to correspond to an actual effective interference of $0.5\ \text{mm}$ in the hand-reamed foam model. The deformation in the foam model and the cadaver can therefore be seen to be very similar (about $30\ \mu\text{m}$ and about $20\ \mu\text{m}$ respectively).

There are a number of limitations of the present study. First of all, neither the polyurethane foam used to represent bone nor the cadaver bone exhibits significant viscoelastic features. Therefore, the effect of viscoelastic characteristics generally associated with both cortical and cancellous bone on cup deformation and corresponding stresses was not taken into consideration in the present study. It is generally expected that the viscoelastic relaxation of living bone would lessen the cup deformation in a longer term and reduce the stresses. However, it is still important to limit and control the short-term cup deformation, since this is directly related to the post-surgical rehabilitation and recovery. Large errors from hand-reamed cavities were found and the effective interference was only about half the nominal value. Although it was possible to reduce the reaming error and its effect on the cup deformation by using machined cavities, this would be very difficult to achieve under clinical settings [22]. All these should be investigated in future studies. Furthermore, only one size of $60\ \text{mm}$ was mainly considered and different sizes will be investigated in Part 2 [16].

5 SUMMARY

The effects of the interference press fit of a metallic one-piece acetabular cup employed for MOM hip resurfacing prosthesis was investigated under laboratory conditions using both cadaver and polyurethane

foams. The diametral cup deformation as a result of the press fit was measured on different foam cavity models and grades. The deformation measured from a two-point pinching cavity model using grade 30 foam represented the worst-case scenario and was consistent with the cadaver test. The maximum diametral cup deflection of an experimental prototype thin cup with wall thicknesses between 2.3 mm (equator) and 4 mm (pole) and an outside diameter of 60 mm, was between 60 and 100 μm with actual diametral interferences of between 0.5 and 1 mm respectively. This would correspond to nominal interferences under clinical conditions of 1 and 1.5 mm respectively. This was considered to be excessive when compared with diametral clearances of between 80 and 120 μm generally specified for MOM hip implants from a tribological point of view. Stiffening of the cup by both thickening and lateralizing by 1 mm reduced the maximum diametral cup deformation to below 50 μm with an actual interference of 1 mm and to below 30 μm for an actual interference of approximately 0.5 mm, the latter corresponding to the generally recommended surgical technique of 1 mm of under-reaming.

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