The effect of cup outer sizes on the contact mechanics and cement fixation of cemented total hip replacements

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Abstract: One important loosening mechanism of the cemented total hip arthroplasty is the mechanical overload at the bone-cement interface and consequent failure of the cement fixation. Clinical studies have revealed that the outer diameter of the acetabular component is a key factor in influencing aseptic loosening of the hip arthroplasty. The aim of the present study was to investigate the influence of the cup outer diameter on the contact mechanics and cement fixation of a cemented total hip replacement (THR) with different wear penetration depths and under different cup inclination angles using finite element (FE) method. A three-dimensional FE model was developed based on a typical Charnley hip prosthesis. Two acetabular cup designs with outer diameters of 40 mm and 43 mm were modelled and the effect of cup outer diameter, penetration depth and cup inclination angle on the contact mechanics and cement fixation stresses in the cemented THR were studied. The results showed that for all penetration depths and cup inclination angles considered, the contact mechanics in terms of peak von Mises stress in the acetabular cup and peak contact pressure at the bearing surface for the two cup designs were similar (within 5%). However, the peak von Mises stress, the peak maximum principal stress and peak shear stress in the cement mantle at the bone-cement interface for the 43 mm diameter cup design were predicted to be lower compared to those for the 40 mm diameter cup design. The differences were predicted to be 15%-19%, 15%-22% and 18%-20% respectively for different cup penetration depths and inclination angles, which compares to the clinical difference of aseptic loosening incidence of about 20% between the two cup designs.

Key words: Contact mechanics; Cement fixation; Cup outer diameter; Inclination angle; Penetration; Total hip replacement

1. Introduction

The Charnley total hip replacement (THR) has been widely used in clinical practice since 1962. The success of this prosthesis has been attributed mainly to its low frictional torque [1,2]. Follow-up studies of Charnley hip replacements have generally shown the arthroplasty to have excellent long-term functional outcome and survivorship. However, like all the other types of artificial hip joints using a metal-on-polyethylene articulation, aseptic loosening of the components, particularly on the acetabular side, has caused majority of the revision and failure of the prostheses [3-6].

The etiology of aseptic loosening of the hip replacement is multifactorial. Osteolytic bone resorption due to the wear particles mainly generated at the articulating surfaces is widely accepted as the main cause [7]. Other mechanisms have also been proposed, including cement damage, bone adaptation, micromotion and high fluid pressure etc. [8]. Particularly, the damage of the cement mantle and subsequent failure of the fixation has been identified as one of possible mechanisms that initiates the loosening and eventual failure of the hip prosthesis [9,10]. The damage of the cement mantle can further produce cement particles, which can invade the articulating surfaces and cause more severe third-body articulating wear. The damage can also provide a pathway for the particulate debris to access the bone-cement interface directly, facilitating the propagation of inflammatory and eventual osteolytic events [11,12].

Evidences from finite element (FE) studies and in vitro experiments indicate that the damage of the cement mantle and failure of the fixation is closely associated with the mechanical behaviour within the cement mantle and at the bone-cement interface [13,14]. Coultrup et al. developed a computational cement damage accumulation method to investigate the effect of polyethylene wear rate, cement mantle thickness and porosity on the mechanical failure of the cemented hip replacement [15]. They demonstrated that both cup penetration and decreased cement thickness increased the cement stresses, resulting in a reduction in the cement mantle fatigue life. They also suggested that the mechanical factors in the cement mantle make a major contribution to the failure mode of cemented polyethylene cups. Lamvohee et al. investigated the stresses in the cement mantle considering the effect of femoral implant size and bone quality. The FE results indicated that both good quality bone and smaller sized femoral component led to decreased stresses in the cement mantle, resulting in a higher survivorship for the cement [16]. Tong and colleagues conducted a series of FE simulations and in vitro experiments to investigate the damage evolution and fatigue failure of the cement mantle in cemented acetabular replacements under different loading conditions. They demonstrated that the failure of the cement fixation initiated in the region where the high stresses were identified from the FE studies [17-21]. All these studies have indicated that high stresses developed in the cement mantle can lead to the damage of the cement mantle and failure of the fixation, which could potentially lead to the loosening of the components and failure of the prostheses.

It is generally believed that the performance of the cemented hip replacement and the mechanical behaviour in the cement mantle near the bone-cement interface is related to many factors such as the head diameter [16,22], penetration depth [15,23], cement thickness [15,16], bone quality [16] and cup outer size [24] etc. Specifically, a clinical study has shown that under similar conditions, the incidence of aseptic loosening for the acetabular cup with outer diameter of 43 mm was smaller than that with outer diameter of 40 mm when penetration depth increases. This was attributed to the lower friction torque with larger outer diameter of the acetabular cup in this study [2]. However, whether other factors, such as the wear in the polyethylene cup and the stresses developed in the cement mantle at the bone-cement interface, will contribute to the different clinical performance of the two prosthesis designs is not recognized. The synergistic effect of the cup outer diameter and cup penetration depth on the contact mechanics of the bearing and stresses of the cement fixation for the cemented hip replacement is not fully understood.

The aims of the present study were therefore to investigate how the cup outer diameters influence the contact mechanics of the bearing and the mechanical behaviour of the cement mantle in the cemented hip replacement with different penetration depths and under different cup inclination angles, and by doing this, to explore whether the contact mechanics and cement mechanical behaviour are the contributed factors causing the different performance of the two sized arthroplasties (with cup outer diameters of 40 mm and 43 mm) observed clinically.

2. Materials and methods

A typical Charnley hip system, consisting of a hemispherical ultra-high molecular weight polyethylene (UHMWPE) cup and a stainless femoral head was analysed. The nominal diameters of the femoral head and inner surface of the cup were 22.225 mm and 22.59 mm respectively [25]. Two acetabular cups with outer diameters of 40 mm and 43 mm, which represent two typical Charnley designs in clinical practice [2], were modelled. The thickness of the polymethylmethacrylate (PMMA) bone cement was assumed to be 2 mm, as previous studies suggested that the thickness of the cement mantle should be not less than 2 mm for the 22.225 sized arthroplasty[16, 26]. The geometries of the acetabular cup and cement layer were assumed to be hemispherical, as shown in Fig. 1 and Fig. 2. Different penetration depths of 1 mm, 2 mm and 4 mm on the acetabular cup and different cup inclination angles of 45°, 55° and 65° were considered. Penetration was simulated by intersecting the cup using the femoral head in the direction of resultant load. Firstly, the femoral head was offset in the direction of the load by a distance of the desired penetration depth. The material of the cup overlapped with the femoral head was then removed to get the worn cup. The cup inclination angle was defined as the angle between the plane of the face of the acetabular cup and the anatomical transverse plane. The reconstruction of the penetration and cup inclination angles are illustrated in Fig. 1.

A three-dimensional FE model was developed to simulate the positions of both the femoral head and acetabular cup implanted in a hemi-pelvic bone model (Fig. 2). The hemi-pelvic bone model consisted of a cancellous bone region surrounded by a uniform cortical shell. The acetabular subchondral bone was assumed to have been reamed completely prior to implantation.

All the materials in the FE model were modelled as homogenous, isotropic and linearly elastic except the UHMWPE cup which was modelled as a non-linear elastic-plastic material with the plastic stress-strain constitutive relationship shown in Fig. 3 [27]. The other material properties used in this study are given in Table 1. The femoral component was assumed to be rigid because the elastic modulus of this metallic component is at least two orders of magnitude greater than that of the UHMWPE material. The cortical shell and cancellous bone in the pelvis were simulated using three-node shell elements and four-node tetrahedral elements respectively while the acetabular cup and cement mantle were modelled using eight-node brick elements and six-node wedge elements. An offset of 1.5 mm was applied for the shell element, representing a thickness of 1.5 mm for the cortical bone of the pelvis [23]. Mesh convergence studies were carried out for the cup design with outer diameter of 40 mm under cup inclination angle of 65° with no penetration in the cup and with penetration depth of 4 mm. Nine models with different levels of mesh density for the pelvis bone (with element numbers of 3032, 5608, 11304), acetabular cup and cement (with element numbers combinations of 1292/3280, 2076/5184, 4416/10720) were tested for each condition. The results showed convergence trends with respect to the peak contact pressure on the bearing surfaces and the peak stresses of the cement mantle in terms of von Mises stress, maximum principal stress and shear stress in the cement mantle at the bone-cement interface, with the differences in the results between the two finest meshes being within 5%. Therefore, the mesh density with approximately 5600, 2100 and 5200 elements for the pelvic bone, acetabular cup and cement mantle respectively was selected for all FE models in the present study.

A frictionless sliding contact formulation was applied to the articulating surface between the head and the cup. The nodes situated at the sacroiliac joint and about the pubic symphysis were fully constrained to simulate the sacral and pubic support of the pelvic bone. The interfaces between the bone and the cement as well as between the cement and the prosthesis were fully bonded, aiming to simulate a fully bone cement interlock and perfect fixation. A fixed load of 2500 N with an angle of 10° medially was applied to the model through the centre of the femoral head, simulating the mid-to-terminal stance loading of the gait cycle [28]. The FE analysis was performed using ABAQUS software package (Version 6.9, Abaqus Inc.).

3. Results

The peak contact pressure on the bearing surface for the two cup designs with outer diameters of 40 mm and 43 mm were located at the superior region of the acetabular cup in line with the load vector, and same pattern of the contact pressure was observed between the two designs (Fig. 4).

For all cup inclination angles considered, an increase in the penetration depth in the acetabular cup up to 4 mm led to a marked decrease of both the peak von Mises stress in the acetabular cup and the peak contact pressure on the bearing surface by 20-32% and 41-50%

respectively (Fig. 5 a and b). At the same level of penetration depth, the peak von Mises stress and peak contact pressure for the prosthesis with cup outer diameter of 40 mm were observed to be higher than that with outer diameter of 43 mm. However, the discrepancies were negligible (less than 5%) (Fig. 5 a and b).

The peak von Mises stress and peak maximum principal stress of the cement mantle at the bone-cement interface were predicted at the superior region of the cement mantle. The magnitudes for the 40 mm diameter cup design were higher than those for the 43 mm diameter cup (Fig. 6 and Fig. 7).

For all cup inclination angles and two cup designs considered, a modest penetration in the acetabular cup resulted in a decreased peak von Mises stress and peak shear stress, as well as peak maximum principal stress in the cement mantle at the bone-cement interface. However, when the penetration depth was increased to 4 mm, higher peak stresses were predicted (Fig. 8 a, b and c).

At a given penetration depth and cup inclination angle, the peak von Mises stress, peak shear stress and peak maximum principal stress of the cement mantle at the bone-cement interface for the model with 40 mm cup outer diameter were observed to be higher compared to those with 43 mm cup outer diameter. The discrepancies were predicted to be 15%-19%, 15%-22% and 18%-20% respectively. It is also interesting to note that for the cup design with outer diameter of 43 mm, the peak stresses were less influenced by the penetration depths compared to that with outer diameter of 40 mm (Fig. 8 a, b and c).

4. Discussion

The principal objectives of the present study were to determine the effect of cup outer diameter, cup penetration depth and cup inclination angle on the contact mechanics of the bearing surface and mechanical behaviour of the cement mantle for a typical cemented metal-on-UHMWPE THR, and to explore whether the contact mechanics of the bearing and mechanical behaviour of the cement fixation should be responsible for the different clinical performance of two prosthesis designs. The mechanical behaviour in terms of von Mises stress, shear stress and maximum principal stress in the cement mantle at the bone-cement interface were examined, due to the fact that the von Mises stress and maximum principal stress are directly associated with the fatigue failure and tensile damage of the cement mantle, and the shear stress could be an important contributor to the shear damage at the bone-cement interface, all of which can consequently lead to the loosening and failure of cemented acetabular components [9,10,29,30]. The bone-cement interface was examined in detail, since the failure of the cement fixation is likely to be initiated at this interface [18-21], and the stress variation across the thickness of the cement mantle was found to be within 10%. The validation of the present study was conducted by comparing the present predictions of contact area and contact pressure with the experimental measurement and FE predictions in a previous study carried out by Jin et al [25], for the same prosthesis design and under the same conditions. Excellent agreements were obtained between the present predictions and the previous results, with a maximum difference of 5%.

The FE predictions from the present study showed that under the same cup inclination angle condition, similar tribological characteristics in terms of contact pressure on the bearing surface and von Mises stresses in the acetabular cup were observed between the hip prostheses with cup outer diameter of 43 mm and 40 mm at a given penetration depth. This can be explained from the consideration of the cup thickness and conformity. Due to the sufficient thickness of the acetabular cup, for the 40 mm prostheses, the cup thickness is approximately 8.7 mm, an increased diameter of 43 mm results in an increased cup thickness to around 10.2 mm. However, such an increase in the cup thickness is unlikely to cause large changes in the contact mechanics at the articulating surfaces [31]. Even though the severe penetration contributes to the decrease of the cup thickness, the improved conformity could compensate such a loss. Furthermore, the results indicated that wear would not be influenced by the cup outer diameter considered in the present study, since neither the contact area, contact pressure nor the motion between the head and cup were altered markedly by the increased cup outer diameter.

It is interesting to note that the peak von Mises stress and peak maximum principal stress of the cement mantle at the bone-cement interface occurred in the superior quadrant of the cement mantle, which was consistent with the region where the initial failure of the cement fixation was observed in vitro [19-21]. The peak von Mises stress, peak maximum principal stress and peak shear stress of the cement mantle at the bone-cement interface for the hip prosthesis with cup outer diameter of 40 mm were predicted to be higher compared to those for the 43 mm prosthesis for all inclination and penetration conditions. This observation was supported by the previous studies conducted by Lamvohee et al [16,32], who reported that both the maximum tensile stress and shear stress in the cement mantle decreased with an increasing acetabular component size. This is presumably due to the fact that for a given penetration depth, a larger cup outer diameter implies an increase in the thickness of the acetabular cup which helped to distribute the stresses better in the acetabular component itself rather than transferring the compressive loading to the cement mantle directly.

A clinical study has shown that under similar conditions, the cup with outer diameter of 43 mm had a smaller chance of aseptic loosening with increasing penetration depths compared to that with outer diameter of 40 mm. This was attributed to the lower friction torque with larger outer diameter of the acetabular cup [2]. The present study, however, provided another explanation. It has been suggested that wear would not be influenced by the cup outer diameter for the two cup designs and therefore is not the major contribution factor to the difference of aseptic loosening incidence observed clinically. However, it is interesting to note that the peak von Mises stress, peak maximum principal stress and peak shear stress of the cement mantle at the bone-cement interface for the 43 mm cup outer diameter hip prosthesis were predicted to be lower compared to those for the 40 mm prosthesis, with the differences of 15%-19%, 15%-22% and 18%-20% respectively. Such discrepancies were found to compare to the difference of aseptic loosening incidence of about 20% between the two cup designs reported clinically [2]. Therefore, it is proposed that in addition to the friction torque, the difference of stresses amplification in the cement mantle at the bone-cement interface between the two cup designs could also be responsible for the different incidence of aseptic loosening observed clinically.

There were, however, a number of limitations with the present computational simulation. The main limitation was that the cement-implant interface was fully boned in the present study to simulate a perfect cement fixation for purpose of simplifying the FE simulations, which, however, may not conform to the real clinical practice. Therefore, additional simulations, considering a standard contact formulation with friction efficient of 0.16 for the cement-implant interface [33], were conducted for the prosthesis with cup outer diameter of 40 mm and for the penetration depths of 0 mm, 1 mm, 2 mm and 4 mm under cup inclination angle of 45º. The simulation results showed that for all these conditions considered, the assumption of considering the cement-implant fixation as being bonded has little effect on the simulation results compared to the case considering a contact formulation for the interface, with differences within 2% and 3% for the peak von Mises stress and contact pressure of the cup respectively, and within 4%, 5% and 3% for the peak von Mises stress, maximum principal stress and shear stress of the cement mantle respectively. This suggested that the assumptions made in the present study were considered to be justified. The geometrical characterization of the penetration in the acetabular cup was simplified by intersecting the cup using the femoral head. Therefore, the local clearance between the femoral head and worn region of the cup was assumed to be zero and the wear direction was assumed towards the direction of the resultant load. However, it is interesting to note in retrieval studies that there were clearances between the worn area of the cup and the femoral head, and the direction of the wear in the cup was generally observed to be lateral with respect to the cup position in the human body [34,35]. Therefore, the specific clearance and wear direction need to be further studied. Additionally, a static constant loading with fixed direction was considered in the present study, representing the maximum contact force on the joint during the normal walking gait. However, both the magnitude and the direction of the contact forces vary during gait which may affect how the contact pressure distributes on the articulating surfaces and potentially the stresses in the cement mantle. Therefore, whilst the case considered in the present study is likely to cause the highest stress in the cement, further studies analysing the whole gait cycle with damage accumulation would provide a better indication of how the location and magnitude of the maximum stress varies and how damage would build up over time [15]. The pelvic bone was assumed as homogenous material in the present study. However, previous studies have shown that the real pelvic bone has a non-homogenous, anisotropic property and the material properties of the bone are site-dependent and density-dependent, this assumption may have some effect on the simulated results. A heterogeneous anisotropic material for the bone should be considered in future studies [37, 38]. More adverse conditions such as edge loading and microseparation conditions as well as potential impingement with higher cup inclination and anteversion angles should also be investigated to further understand the clinical observations and failure mechanism of hip replacements seen across a real patient cohort [39-42].

5. Conclusions

FE analyses of the present study showed that for a given penetration depth and cup inclination angle, the contact mechanics features at the bearing surface between the hip replacements with cup outer diameter of 43 mm and 40 mm were similar. However, the peak von Mises stress, maximum principal stress and shear stress of the cement mantle at the bone-cement interface for the hip arthroplasty with a cup outer diameter of 43 mm were predicted to be lower compared to those for 40 mm arthroplasty, and the differences were found to be comparable to the difference of aseptic loosening incidence between the two cup designs observed clinically. Therefore, the present study suggests that in addition to the friction torque, the difference of stresses developed in the cement mantle between the two cup designs is also responsible for the different incidence of aseptic loosening for the two cup designs observed clinically.

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Conflict of interest

JF is a consultant to Depuy.

Ethical approval

Not required.

References

[1] Charnley J. Arthroplasty of the hip. A new operation. Lancet 1961; 1(7187):1129-32.

[2] Wroblewski BM, Siney PD, Fleming PA. The principle of low frictional torque in the Charnley total hip replacement. J Bone Joint Surg Br 2009; 91(7):855-8.

[3] Garellick G, Herberts P, Strömberg C, Malchau H. Long-term results of Charnley arthroplasty. A 12-16-year follow-up study. J Arthroplasty 1994; 9(4):333-40.

[4] Callaghan JJ, Bracha P, Liu SS, Piyaworakhun S, Goetz DD, Johnston RC. Survivorship of a Charnley total hip arthroplasty. a concise follow-up: at a minimum of thirty-five years, of previous reports. J Bone Joint Surg Am 2009; 91(11):2617–21.

[5] Wroblewski BM, Siney PD, Fleming PA. Charnley low-frictional torque arthroplasty: follow-up for 30 to 40 years. J Bone Joint Surg Br 2009; 91(4):447-50.

[6] Caton J, Prudhon JL. Over 25 years survival after Charnley's total hip arthroplasty. Int Orthop 2011; 35(2):185-8.

[7] Ingham E, Fisher J. The role of macrophages in osteolysis of total joint replacement. Biomaterials 2005; 26(11):1271-86.

[8] Sundfeldt M, Carlsson LV, Johansson CB, Thomsen P, Gretzer C. Aseptic loosening, not only a question of wear: a review of different theories. Acta Orthop 2006; 77(2):177-97.

[9] Kuehn KD, Ege W, Gopp U. Acrylic bone cements: mechanical and physical properties. Orthop Clin North Am 2005; 36(1):29-39.

[10] McCormack BA, Prendergast PJ. Microdamage accumulation in the cement layer of hip replacements under flexural loading. J Biomech 1999; 32(5):467-75.

[11] Schmalzried TP, Kwong LM, Jasty M, Sedlacek RC, Haire TC, O'Connor DO, et al. The mechanism of loosening of cemented acetabular components in total hip arthroplasty. Analysis of specimens retrieved at autopsy. Clin Orthop Relat Res 1992; (274):60-78.

[12] Hirakawa K, Jacobs JJ, Urban R, Saito T. Mechanisms of failure of total hip replacements: lessons learned from retrieval studies. Clin Orthop Relat Res 2004; (420):10-7.

[13] Jasty M, Maloney WJ, Bragdon CR, O'Connor DO, Haire T, Harris WH. The initiation of failure in cemented femoral components of hip arthroplasties. J Bone Joint Surg Br 1991; 73(4):551-8.

[14] Huiskes R. Failed innovation in total hip replacement. Diagnosis and proposals for a cure. Acta Orthop Scand 1993; 64(6):699-716.

[15] Coultrup OJ, Hunt C, Wroblewski BM, Taylor M. Computational assessment of the effect of polyethylene wear rate, mantle thickness, and porosity on the mechanical failure of the acetabular cement mantle. J Orthop Res 2010; 28(5):565-70.

[16] Lamvohee JM, Mootanah R, Ingle P, Cheah K, Dowell JK. Stresses in cement mantles of hip replacements: effect of femoral implant sizes, body mass index and bone quality. Comput Methods Biomech Biomed Engin 2009; 12(5):501-10.

[17] Heaton-Adegbile P, Zant NP, Tong J. In vitro fatigue behaviour of a cemented acetabular reconstruction. J Biomech 2006; 39(15):2882-6.

[18] Zant NP, Wong CKY, Tong J. Fatigue failure in the cement mantle of a simplified acetabular replacement model. Int J Fatigue 2007; 29(7):1245-52.

[19] Tong J, Zant NP, Wang JY, Heaton-Adegbile P, Hussell JG. Fatigue in cemented acetabular replacements. Int J Fatigue 2008; 30(8):1366-75.

[20] Zant NP, Heaton-Adegbile P, Hussell JG, Tong J. In vitro fatigue failure of cemented acetabular replacements: a hip simulator study. J Biomech Eng 2008; 130(2):021019.

[21] Wang JY, Heaton-Adegbile P, New A, Hussell JG, Tong J. Damage evolution in acetabular replacements under long-term physiological loading conditions. J Biomech 2009; 42(8):1061-8.

[22] Morrey BF, Ilstrup D. Size of the femoral head and acetabular revision in total hip-replacement arthroplasty. J Bone Joint Surg Am 1989; 71(1):50-5.

[23] Hua X, Wroblewski BM, Jin Z, Wang L. The effect of cup inclination and wear on the contact mechanics and cement fixation for ultra high molecular weight polyethylene total hip replacements. Med Eng Phys 2012; 34(3):318-25.

[24] Phillips AT, Pankaj, Usmani AS, Howie CR. The effect of acetabular cup size on the short-term stability of revision hip arthroplasty: a finite element investigation. Proc Inst Mech Eng H 2004; 218(4):239-49.

[25] Jin ZM, Heng SM, Ng HW, Auger DD. An axisymmetric contact model of ultra high molecular weight polyethylene cups against metallic femoral heads for artificial hip joint replacements. Proc Inst Mech Eng H 1999; 213(4):317-27.

[26] Lankester, BJA, Stoney, J, Gheduzzi, S, Miles, AW, Bannister, GC. An in-vitro evaluation of optimal acetabular cement mantle thickness. J Bone Joint Surg Br 2004; 86-B (1):75-76.

[27] Liu F. Contact mechanics and elastohydrodynamic lubrication analysis of metal-on-metal hip implant with a sandwich acetabular cup under transient walking condition. PhD Thesis, 2005; University of Bradford.

[28] Bergmann G, Deuretzbacher G, Heller M, Graichen F, Rohlmann A, Strauss J, Duda GN. Hip contact forces and gait patterns from routine activities. J Biomech 2001; 34(7): 859-71.

[29] Kim DG, Miller MA, Mann KA. A fatigue damage model for the cement-bone interface. J Biomech 2004; 37(10):1505-12.

[30] Arola D, Stoffel KA, Yang DT. Fatigue of the cement/bone interface: the surface texture of bone and loosening. J Biomed Mater Res B Appl Biomater. 2006; 76(2):287-97.

[31] Bartel DL, Bicknell VL, Wright TM. The effect of conformity, thickness, and material on stresses in ultra-high molecular weight components for total joint replacement. J Bone Joint Surg Am 1986; 68(7):1041-51.

[32] Lamvohee JMS, Mootanah R., Dowell, Ingle P. Patients' bone morphology and bone quality affect the performance of fixation techniques in cemented total hip replacements. J Biomech 2007; 40(Suppl. 2):228.

[33] Kurtz SM, Edidin AA, Bartel DL. The role of backside polishing, cup angle, and polyethylene thickness on the contact stresses in metal-backed acetabular components. J Biomech 1997; 30(6):639-42.

[34] Wroblewski BM. Direction and rate of socket wear in Charnley low-friction arthroplasty. J Bone Joint Surg Br 1985; 67(5):757-61.

[35] Hall RM, Siney P, Wroblewski BM, Unsworth A. Observations on the direction of wear in Charnley sockets retrieved at revision. J Bone Joint Surg Br 1998; 80(6):1067-72.

[36] Udofia I, Liu F, Jin Z, Roberts P, Grigoris P. The initial stability and contact mechanics of a press-fit resurfacing arthroplasty of the hip. J Bone Joint Surg Br 2007; 89(4):549-56.

[37] Dalstra M, Huiskes R, Odgaard A, van Erning L. Mechanical and textural properties of pelvic trabecular bone. J Biomech 1993; 26(4-5): 523-35.

[38] Dalstra M, Huiskes R, van Erning L. Development and validation of a 3-dimensional finite-element model of the pelvic bone. J Biomech Eng 1995; 117(3): 272-8.

[39] Wroblewski BM, Siney PD, Fleming PA. Charnley low-friction arthroplasty: survival patterns to 38 years. J Bone Joint Surg Br 2007; 89(8):1015-8.

[40] Fisher J. Bioengineering reasons for the failure of metal-on-metal hip prostheses: an engineer's perspective. J Bone Joint Surg Br 2011; 93(8):1001-4.

[41] Harris WH. Edge loading has a paradoxical effect on wear in metal-on-polyethylene total hip arthroplasties. Clin Orthop Relat Res 2012; 470(11):3077-82.

[42] Hua X, Li J, Wang L, Jin Z, Wilcox R, Fisher J. Contact mechanics of modular metal-on-polyethylene total hip replacement under adverse edge loading conditions. J Biomech 2014; 47(13): 3303-9.

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Fig. 1 A schematic cross-section showing the generation of the wear penetration in the acetabular cup. Firstly, the femoral head was offset in the direction of the load application by a distance of the penetration depth simulated. The material of the cup overlapped with the femoral head was then removed to get the desired penetration depth.

Fig. 2 The boundary conditions and loading conditions for the three-dimensional FE model. The load was applied to the centre of the femoral head with a direction of 10° medially.

Fig. 3 The plastic stress-strain relationship for the UHMWPE [27].

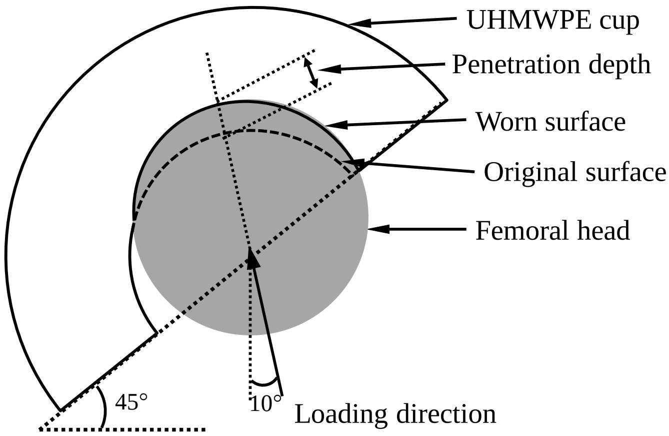
Fig. 4 Contour plots of the predicted contact pressures (MPa) on the bearing surface at cup inclination angle of 45º and for penetration depth of 1 mm with cup outer diameters of (a) 40 mm and (b) 43 mm.

Fig. 5 The predicted peak stresses for the acetabular cup as a function of penetration depth with different cup inclination angles and cup outer diameters: (a) peak von Mises stress in the acetabular cup, (b) peak contact pressure on the bearing surface.

Fig. 6 Comparison of the predicted von Mises stresses (MPa) of the cement mantle at 45º cup inclination angle and for 1 mm penetration depth for different cup outer diameters. The images show the stresses (MPa) at the bone-cement interface for cup outer diameters of (a) 40 mm and (b) 43 mm; and the stresses within the cement mantle for cup outer diameters of (c) 40 mm and (d) 43 mm.

Fig. 7 Comparison of the predicted maximum principal stresses (MPa) of the cement mantle at 45º cup inclination angle and for 1 mm penetration depth for different cup outer diameters. The images show the stresses at the bone-cement interface for cup outer diameters of (a) 40 mm and (b) 43 mm, and the stresses within the cement mantle for cup outer diameters of (c) 40 mm and (b) 43 mm.

Fig. 8 The predicted peak stresses of the cement mantle as a function of penetration depths with different cup inclination angles and cup outer diameters: (a) peak von Mises stress, (b) peak maximum principal stress and (c) peak shear stress at the bone-cement interface.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Penetration depth
Worn surface
Original surface
Femoral head
45°
10°
Loading direction

Fig. 1

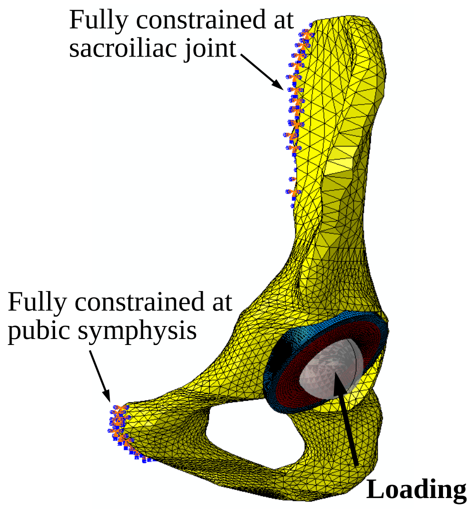
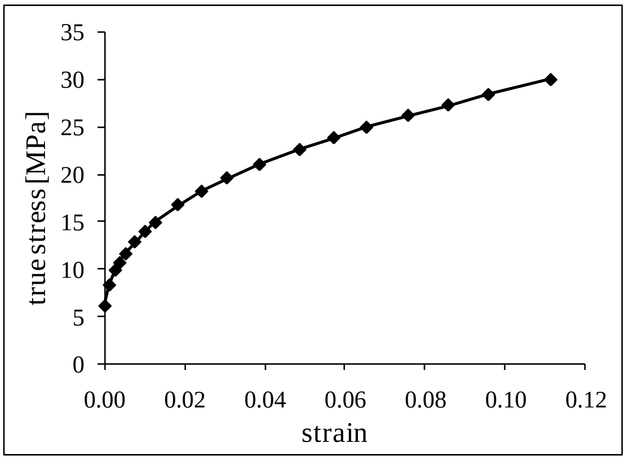

Fully constrained at
sacroiliac joint
Fully constrained at
pubic symphysis
Loading

Fig. 2


true stress [MPa]
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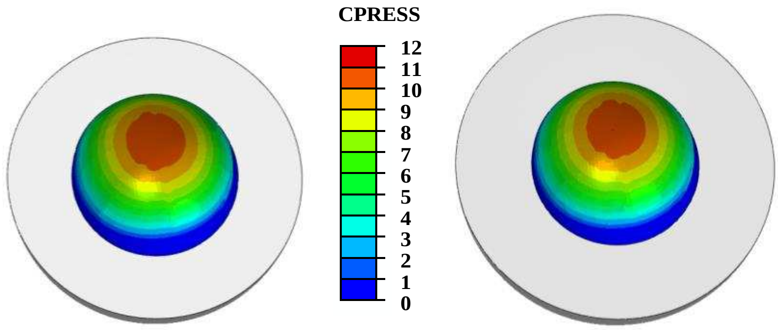
strain

Fig. 3



(a) (b)

Fig. 4

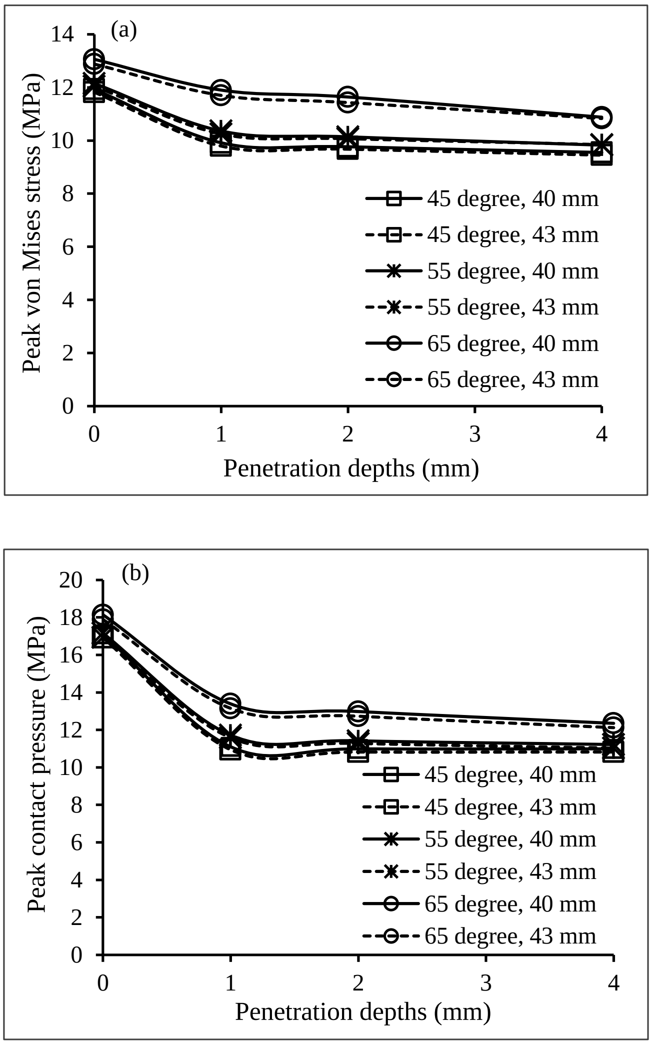

(a)
Peak von Mises stress (MPa)
Penetration depths (mm)
45 degree, 40 mm
45 degree, 43 mm
55 degree, 40 mm
55 degree, 43 mm
65 degree, 40 mm
65 degree, 43 mm
(b)
Peak contact pressure (MPa)
Penetration depths (mm)
45 degree, 40 mm
45 degree, 43 mm
55 degree, 40 mm
55 degree, 43 mm
65 degree, 40 mm
65 degree, 43 mm

Fig. 5

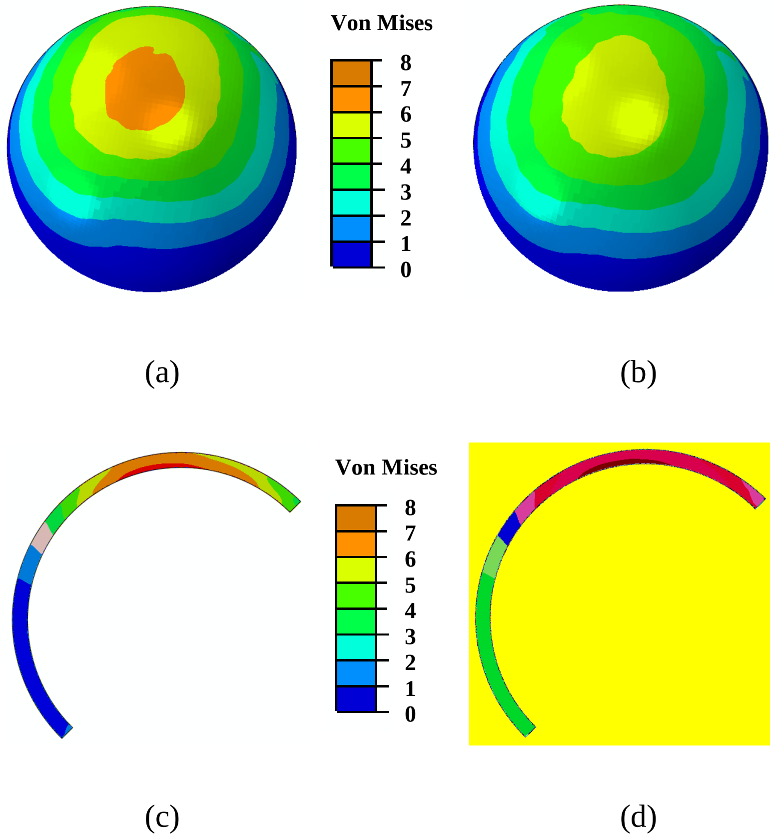


Fig. 6

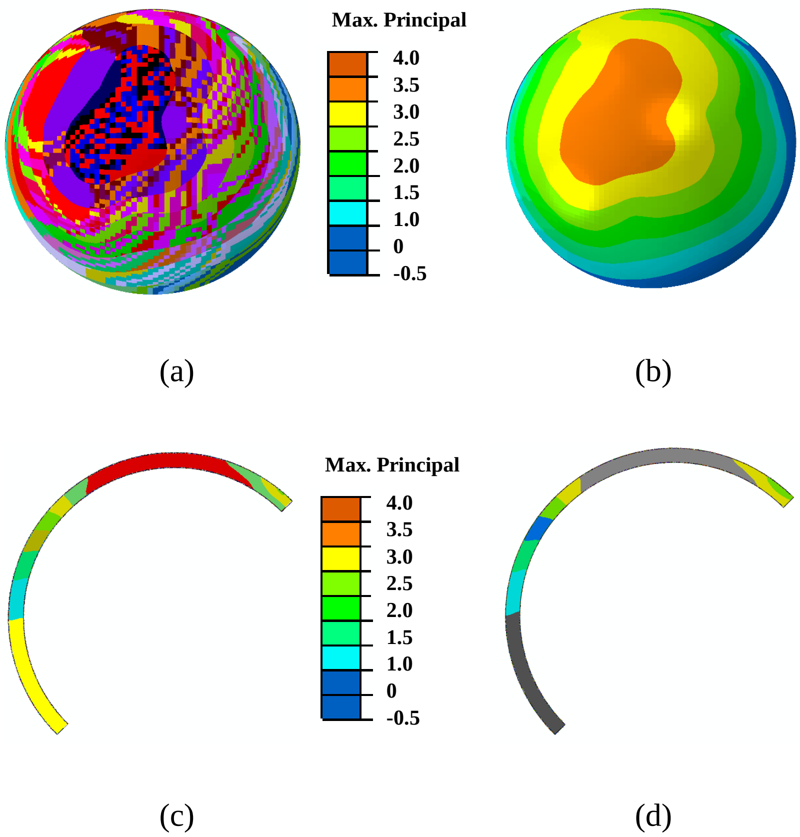

Max. Principal
4.0
3.5
3.0
2.5
2.0
1.5
1.0
0
-0.5
(a)
(b)
Max. Principal
4.0
3.5
3.0
2.5
2.0
1.5
1.0
0
-0.5
(c)
(d)

Fig. 7

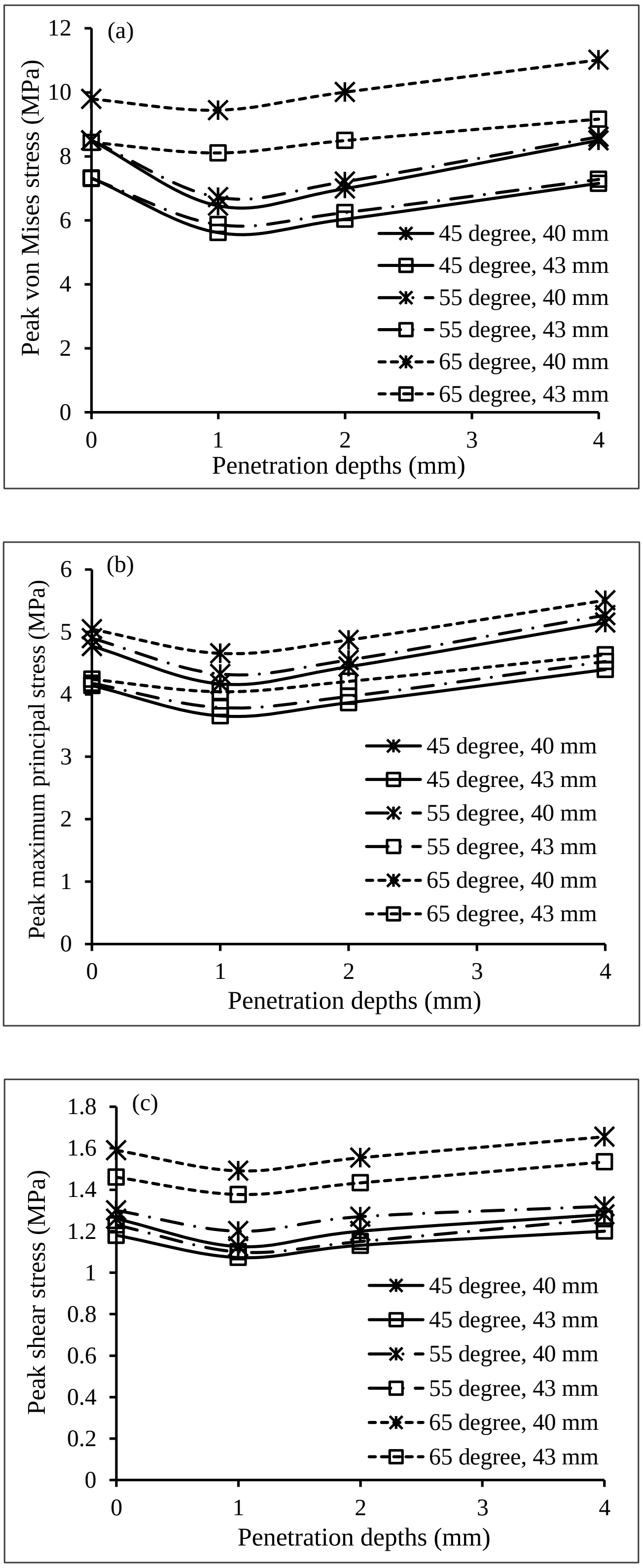

(a)
Peak von Mises stress (MPa)
Penetration depths (mm)
45 degree, 40 mm
45 degree, 43 mm
55 degree, 40 mm
55 degree, 43 mm
65 degree, 40 mm
65 degree, 43 mm
(b)
Peak maximum principal stress (MPa)
Penetration depths (mm)
45 degree, 40 mm
45 degree, 43 mm
55 degree, 40 mm
55 degree, 43 mm
65 degree, 40 mm
65 degree, 43 mm
(c)
Peak shear stress (MPa)
Penetration depths (mm)
45 degree, 40 mm
45 degree, 43 mm
55 degree, 40 mm
55 degree, 43 mm
65 degree, 40 mm
65 degree, 43 mm

Fig. 8

Table 1. The material properties for the components in the present model [25,36].

Components	Materials	Young’s modulus (GPa)	Poisson’s ratio
UHMWPE cup	UHMWPE	1	0.4
Bone cement	PMMA	2.5	0.254
Cortical bone	Cortical bone	17	0.3
Cancellous bone	Cancellous bone	0.8	0.2