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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.

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

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71 penetration and decreased cement thickness increased the cement stresses, resulting in a

72 reduction in the cement mantle fatigue life. They also suggested that the mechanical factors

73 in the cement mantle make a major contribution to the failure mode of cemented polyethylene

74 cups. Lamvohee et al. investigated the stresses in the cement mantle considering the effect of

75 femoral implant size and bone quality. The FE results indicated that both good quality bone

76 and smaller sized femoral component led to decreased stresses in the cement mantle, resulting

77 in a higher survivorship for the cement [16]. Tong and colleagues conducted a series of FE

78 simulations and in vitro experiments to investigate the damage evolution and fatigue failure

79 of the cement mantle in cemented acetabular replacements under different loading conditions.

80 They demonstrated that the failure of the cement fixation initiated in the region where the

81 high stresses were identified from the FE studies [17-21]. All these studies have indicated

82 that high stresses developed in the cement mantle can lead to the damage of the cement

83 mantle and failure of the fixation, which could potentially lead to the loosening of the

84 components and failure of the prostheses.

85 It is generally believed that the performance of the cemented hip replacement and the

86 mechanical behaviour in the cement mantle near the bone-cement interface is related to many

87 factors such as the head diameter [16,22], penetration depth [15,23], cement thickness [15,16],

88 bone quality [16] and cup outer size [24] etc. Specifically, a clinical study has shown that

89 under similar conditions, the incidence of aseptic loosening for the acetabular cup with outer

90 diameter of 43 mm was smaller than that with outer diameter of 40 mm when penetration

91 depth increases. This was attributed to the lower friction torque with larger outer diameter of

92 the acetabular cup in this study [2]. However, whether other factors, such as the wear in the

93 polyethylene cup and the stresses developed in the cement mantle at the bone-cement

94 interface, will contribute to the different clinical performance of the two prosthesis designs is

95 not recognized. The synergistic effect of the cup outer diameter and cup penetration depth on

96   the contact mechanics of the bearing and stresses of the cement fixation for the cemented hip

97   replacement is not fully understood.

98   The aims of the present study were therefore to investigate how the cup outer diameters

99   influence the contact mechanics of the bearing and the mechanical behaviour of the cement

100 mantle in the cemented hip replacement with different penetration depths and under different

101 cup inclination angles, and by doing this, to explore whether the contact mechanics and

102 cement mechanical behaviour are the contributed factors causing the different performance of

103 the two sized arthroplasties (with cup outer diameters of 40 mm and 43 mm) observed

104 clinically.

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106 **2.   Materials and methods**

107   A typical Charnley hip system, consisting of a hemispherical ultra-high molecular weight

108 polyethylene (UHMWPE) cup and a stainless femoral head was analysed. The nominal

109 diameters of the femoral head and inner surface of the cup were 22.225 mm and 22.59 mm

110 respectively [25]. Two acetabular cups with outer diameters of 40 mm and 43 mm, which

111 represent two typical Charnley designs in clinical practice [2], were modelled. The thickness

112 of the polymethylmethacrylate (PMMA) bone cement was assumed to be 2 mm, as previous

113 studies suggested that the thickness of the cement mantle should be not less than 2 mm for

114 the 22.225 sized arthroplasty[16, 26]. The geometries of the acetabular cup and cement layer

115 were assumed to be hemispherical, as shown in Fig. 1 and Fig. 2. Different penetration depths

116 of 1 mm, 2 mm and 4 mm on the acetabular cup and different cup inclination angles of 45°,

117 55° and 65° were considered. Penetration was simulated by intersecting the cup using the

118 femoral head in the direction of resultant load. Firstly, the femoral head was offset in the

119 direction of the load by a distance of the desired penetration depth. The material of the cup

120   overlapped with the femoral head was then removed to get the worn cup. The cup inclination

121   angle was defined as the angle between the plane of the face of the acetabular cup and the

122   anatomical transverse plane. The reconstruction of the penetration and cup inclination angles

123   are illustrated in Fig. 1.

124   A three-dimensional FE model was developed to simulate the positions of both the femoral

125   head and acetabular cup implanted in a hemi-pelvic bone model (Fig. 2). The hemi-pelvic

126   bone model consisted of a cancellous bone region surrounded by a uniform cortical shell. The

127   acetabular subchondral bone was assumed to have been reamed completely prior to

128   implantation.

129   All the materials in the FE model were modelled as homogenous, isotropic and linearly

130   elastic except the UHMWPE cup which was modelled as a non-linear elastic-plastic material

131   with the plastic stress-strain constitutive relationship shown in Fig. 3 [27]. The other material

132   properties used in this study are given in Table 1. The femoral component was assumed to be

133   rigid because the elastic modulus of this metallic component is at least two orders of

134   magnitude greater than that of the UHMWPE material. The cortical shell and cancellous bone

135   in the pelvis were simulated using three-node shell elements and four-node tetrahedral

136   elements respectively while the acetabular cup and cement mantle were modelled using eight-

137   node brick elements and six-node wedge elements. An offset of 1.5 mm was applied for the

138   shell element, representing a thickness of 1.5 mm for the cortical bone of the pelvis [23].

139   Mesh convergence studies were carried out for the cup design with outer diameter of 40 mm

140   under cup inclination angle of 65° with no penetration in the cup and with penetration depth

141   of 4 mm. Nine models with different levels of mesh density for the pelvis bone (with element

142   numbers of 3032, 5608, 11304), acetabular cup and cement (with element numbers

143   combinations of 1292/3280, 2076/5184, 4416/10720) were tested for each condition. The

144   results showed convergence trends with respect to the peak contact pressure on the bearing

145   surfaces and the peak stresses of the cement mantle in terms of von Mises stress, maximum

146   principal stress and shear stress in the cement mantle at the bone-cement interface, with the

147   differences in the results between the two finest meshes being within 5%. Therefore, the

148   mesh density with approximately 5600, 2100 and 5200 elements for the pelvic bone,

149   acetabular cup and cement mantle respectively was selected for all FE models in the present

150   study.

151   A frictionless sliding contact formulation was applied to the articulating surface between

152   the head and the cup. The nodes situated at the sacroiliac joint and about the pubic symphysis

153   were fully constrained to simulate the sacral and pubic support of the pelvic bone. The

154   interfaces between the bone and the cement as well as between the cement and the prosthesis

155   were fully bonded, aiming to simulate a fully bone cement interlock and perfect fixation. A

156   fixed load of 2500 N with an angle of 10° medially was applied to the model through the

157   centre of the femoral head, simulating the mid-to-terminal stance loading of the gait cycle

158   [28]. The FE analysis was performed using ABAQUS software package (Version 6.9,

159   Abaqus Inc.).

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161   **3.   Results**

162   The peak contact pressure on the bearing surface for the two cup designs with outer

163   diameters of 40 mm and 43 mm were located at the superior region of the acetabular cup in

164   line with the load vector, and same pattern of the contact pressure was observed between the

165   two designs (Fig. 4).

166   For all cup inclination angles considered, an increase in the penetration depth in the

167   acetabular cup up to 4 mm led to a marked decrease of both the peak von Mises stress in the

168   acetabular cup and the peak contact pressure on the bearing surface by 20-32% and 41-50%

169 respectively (Fig. 5 a and b). At the same level of penetration depth, the peak von Mises

170 stress and peak contact pressure for the prosthesis with cup outer diameter of 40 mm were

171 observed to be higher than that with outer diameter of 43 mm. However, the discrepancies

172 were negligible (less than 5%) (Fig. 5 a and b).

173 The peak von Mises stress and peak maximum principal stress of the cement mantle at the

174 bone-cement interface were predicted at the superior region of the cement mantle. The

175 magnitudes for the 40 mm diameter cup design were higher than those for the 43 mm

176 diameter cup (Fig. 6 and Fig. 7).

177 For all cup inclination angles and two cup designs considered, a modest penetration in the

178 acetabular cup resulted in a decreased peak von Mises stress and peak shear stress, as well as

179 peak maximum principal stress in the cement mantle at the bone-cement interface. However,

180 when the penetration depth was increased to 4 mm, higher peak stresses were predicted (Fig.

181 8 a, b and c).

182 At a given penetration depth and cup inclination angle, the peak von Mises stress, peak

183 shear stress and peak maximum principal stress of the cement mantle at the bone-cement

184 interface for the model with 40 mm cup outer diameter were observed to be higher compared

185 to those with 43 mm cup outer diameter. The discrepancies were predicted to be 15%-19%,

186 15%-22% and 18%-20% respectively. It is also interesting to note that for the cup design with

187 outer diameter of 43 mm, the peak stresses were less influenced by the penetration depths

188 compared to that with outer diameter of 40 mm (Fig. 8 a, b and c).

189

190 **4.   Discussion**

191 The principal objectives of the present study were to determine the effect of cup outer

192 diameter, cup penetration depth and cup inclination angle on the contact mechanics of the

193 bearing surface and mechanical behaviour of the cement mantle for a typical cemented metal-

194 on-UHMWPE THR, and to explore whether the contact mechanics of the bearing and

195 mechanical behaviour of the cement fixation should be responsible for the different clinical

196 performance of two prosthesis designs. The mechanical behaviour in terms of von Mises

197 stress, shear stress and maximum principal stress in the cement mantle at the bone-cement

198 interface were examined, due to the fact that the von Mises stress and maximum principal

199 stress are directly associated with the fatigue failure and tensile damage of the cement mantle,

200 and the shear stress could be an important contributor to the shear damage at the bone-cement

201 interface, all of which can consequently lead to the loosening and failure of cemented

202 acetabular components [9,10,29,30]. The bone-cement interface was examined in detail, since

203 the failure of the cement fixation is likely to be initiated at this interface [18-21], and the

204 stress variation across the thickness of the cement mantle was found to be within 10%. The

205 validation of the present study was conducted by comparing the present predictions of contact

206 area and contact pressure with the experimental measurement and FE predictions in a

207 previous study carried out by Jin et al [25], for the same prosthesis design and under the same

208 conditions. Excellent agreements were obtained between the present predictions and the

209 previous results, with a maximum difference of 5%.

210 The FE predictions from the present study showed that under the same cup inclination

211 angle condition, similar tribological characteristics in terms of contact pressure on the bearing

212 surface and von Mises stresses in the acetabular cup were observed between the hip

213 prostheses with cup outer diameter of 43 mm and 40 mm at a given penetration depth. This

214 can be explained from the consideration of the cup thickness and conformity. Due to the

215 sufficient thickness of the acetabular cup, for the 40 mm prostheses, the cup thickness is

216 approximately 8.7 mm, an increased diameter of 43 mm results in an increased cup thickness

217 to around 10.2 mm. However, such an increase in the cup thickness is unlikely to cause large

218   changes in the contact mechanics at the articulating surfaces [31]. Even though the severe

219   penetration contributes to the decrease of the cup thickness, the improved conformity could

220   compensate such a loss. Furthermore, the results indicated that wear would not be influenced

221   by the cup outer diameter considered in the present study, since neither the contact area,

222   contact pressure nor the motion between the head and cup were altered markedly by the

223   increased cup outer diameter.

224   It is interesting to note that the peak von Mises stress and peak maximum principal stress

225   of the cement mantle at the bone-cement interface occurred in the superior quadrant of the

226   cement mantle, which was consistent with the region where the initial failure of the cement

227   fixation was observed in vitro [19-21]. The peak von Mises stress, peak maximum principal

228   stress and peak shear stress of the cement mantle at the bone-cement interface for the hip

229   prosthesis with cup outer diameter of 40 mm were predicted to be higher compared to those

230   for the 43 mm prosthesis for all inclination and penetration conditions. This observation was

231   supported by the previous studies conducted by Lamvohee et al [16,32], who reported that

232   both the maximum tensile stress and shear stress in the cement mantle decreased with an

233   increasing acetabular component size. This is presumably due to the fact that for a given

234   penetration depth, a larger cup outer diameter implies an increase in the thickness of the

235   acetabular cup which helped to distribute the stresses better in the acetabular component itself

236   rather than transferring the compressive loading to the cement mantle directly.

237   A clinical study has shown that under similar conditions, the cup with outer diameter of 43

238   mm had a smaller chance of aseptic loosening with increasing penetration depths compared

239   to that with outer diameter of 40 mm. This was attributed to the lower friction torque with

240   larger outer diameter of the acetabular cup [2]. The present study, however, provided another

241   explanation. It has been suggested that wear would not be influenced by the cup outer

242   diameter for the two cup designs and therefore is not the major contribution factor to the

243   difference of aseptic loosening incidence observed clinically. However, it is interesting to

244   note that the peak von Mises stress, peak maximum principal stress and peak shear stress of

245   the cement mantle at the bone-cement interface for the 43 mm cup outer diameter hip

246   prosthesis were predicted to be lower compared to those for the 40 mm prosthesis, with the

247   differences of 15%-19%, 15%-22% and 18%-20% respectively. Such discrepancies were

248   found to compare to the difference of aseptic loosening incidence of about 20% between the

249   two cup designs reported clinically [2]. Therefore, it is proposed that in addition to the

250   friction torque, the difference of stresses amplification in the cement mantle at the bone-

251   cement interface between the two cup designs could also be responsible for the different

252   incidence of aseptic loosening observed clinically.

253   There were, however, a number of limitations with the present computational simulation.

254   The main limitation was that the cement-implant interface was fully boned in the present

255   study to simulate a perfect cement fixation for purpose of simplifying the FE simulations,

256   which, however, may not conform to the real clinical practice. Therefore, additional

257   simulations, considering a standard contact formulation with friction efficient of 0.16 for the

258   cement-implant interface [33], were conducted for the prosthesis with cup outer diameter of

259   40 mm and for the penetration depths of 0 mm, 1 mm, 2 mm and 4 mm under cup inclination

260   angle of 45°. The simulation results showed that for all these conditions considered, the

261   assumption of considering the cement-implant fixation as being bonded has little effect on the

262   simulation results compared to the case considering a contact formulation for the interface,

263   with differences within 2% and 3% for the peak von Mises stress and contact pressure of the

264   cup respectively, and within 4%, 5% and 3% for the peak von Mises stress, maximum

265   principal stress and shear stress of the cement mantle respectively. This suggested that the

266   assumptions made in the present study were considered to be justified. The geometrical

267   characterization of the penetration in the acetabular cup was simplified by intersecting the

268   cup using the femoral head. Therefore, the local clearance between the femoral head and

269   worn region of the cup was assumed to be zero and the wear direction was assumed towards

270   the direction of the resultant load. However, it is interesting to note in retrieval studies that

271   there were clearances between the worn area of the cup and the femoral head, and the

272   direction of the wear in the cup was generally observed to be lateral with respect to the cup

273   position in the human body [34,35]. Therefore, the specific clearance and wear direction need

274   to be further studied. Additionally, a static constant loading with fixed direction was

275   considered in the present study, representing the maximum contact force on the joint during

276   the normal walking gait. However, both the magnitude and the direction of the contact forces

277   vary during gait which may affect how the contact pressure distributes on the articulating

278   surfaces and potentially the stresses in the cement mantle. Therefore, whilst the case

279   considered in the present study is likely to cause the highest stress in the cement, further

280   studies analysing the whole gait cycle with damage accumulation would provide a better

281   indication of how the location and magnitude of the maximum stress varies and how damage

282   would build up over time [15]. The pelvic bone was assumed as homogenous material in the

283   present study. However, previous studies have shown that the real pelvic bone has a non-

284   homogenous, anisotropic property and the material properties of the bone are site-dependent

285   and density-dependent, this assumption may have some effect on the simulated results. A

286   heterogeneous anisotropic material for the bone should be considered in future studies [37,

287   38]. More adverse conditions such as edge loading and microseparation conditions as well as

288   potential impingement with higher cup inclination and anteversion angles should also be

289   investigated to further understand the clinical observations and failure mechanism of hip

290   replacements seen across a real patient cohort [39-42].

291

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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.

Conflict of interest

JF is a consultant to Depuy.

Ethical approval

Not required.

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408   arthroplasties. Clin Orthop Relat Res 2012; 470(11):3077-82.

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411   3303-9.

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UHMWPE cup

Penetration depth

Worn surface

Original surface

Femoral head

45°

10° Loading direction

Fig. 1

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Fully constrained at
sacroiliac joint

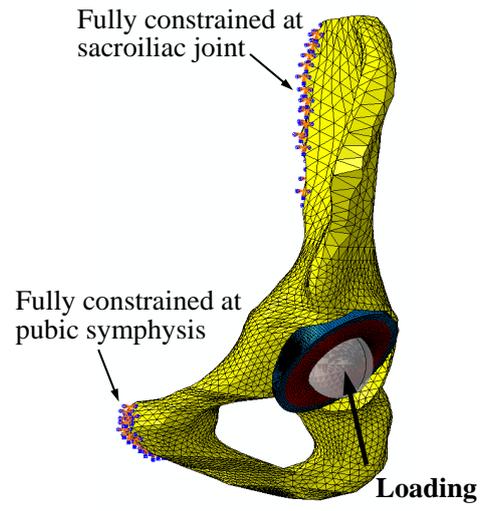Fully constrained at
pubic symphysis

Loading

Fig. 2

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Fig. 3

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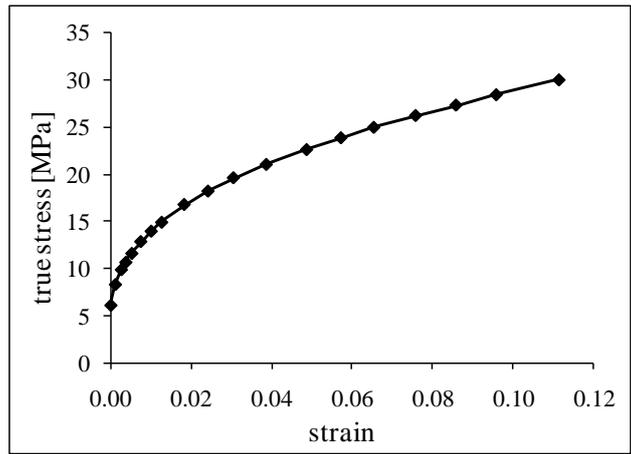
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491   (a)   (b)

492   **Fig. 4**
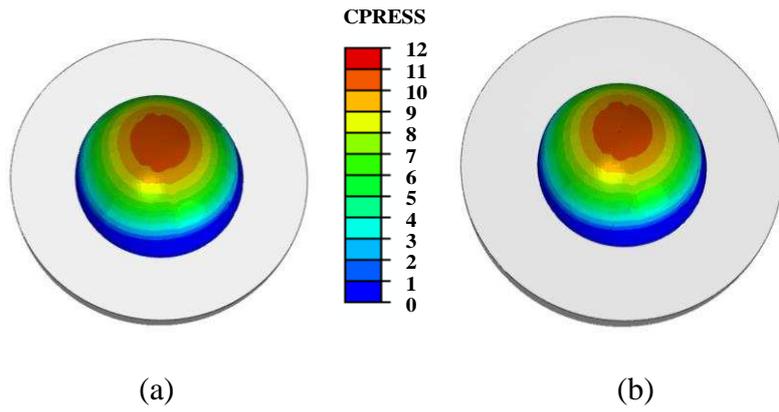
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Fig. 5

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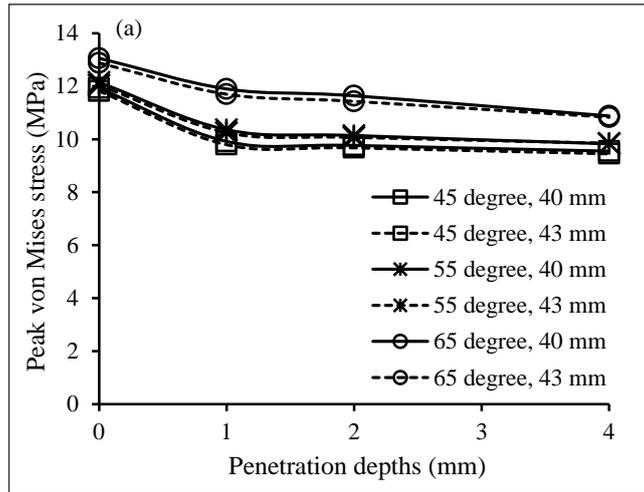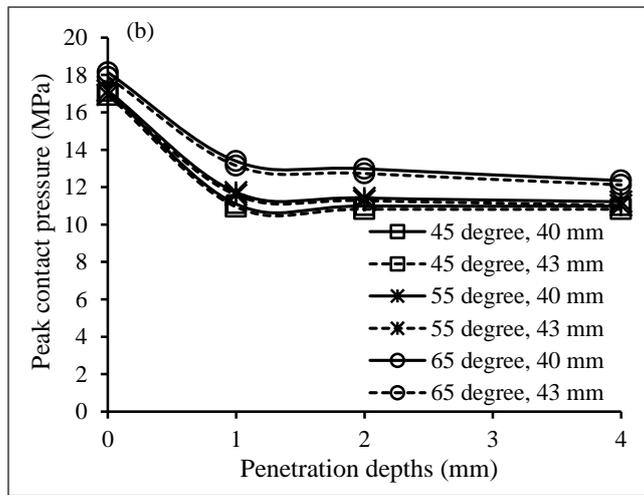
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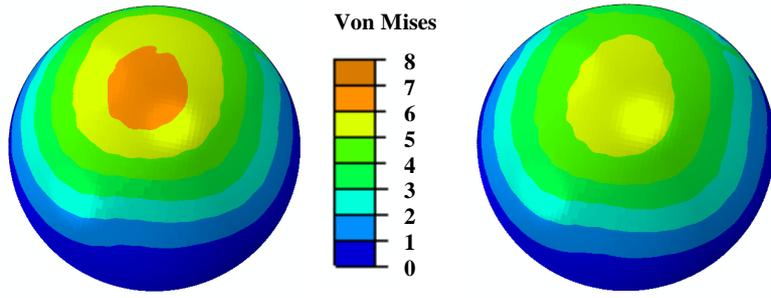
514   (a)   (b)

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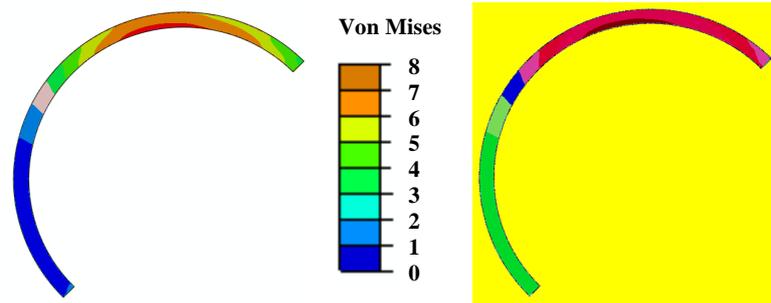
516   (c)   (d)

517   **Fig. 6**

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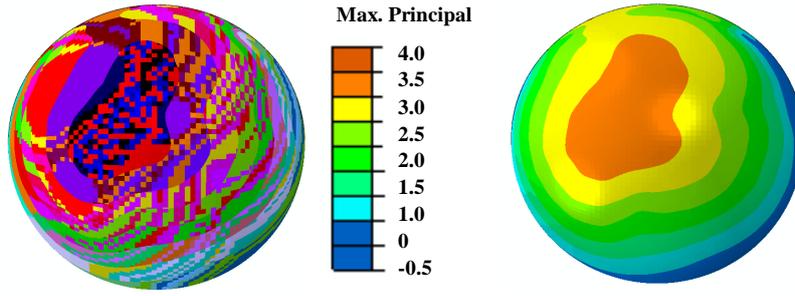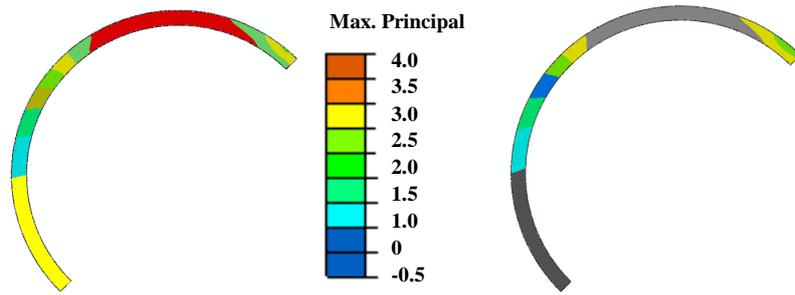527   (a)   (b)

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529   (c)   (d)
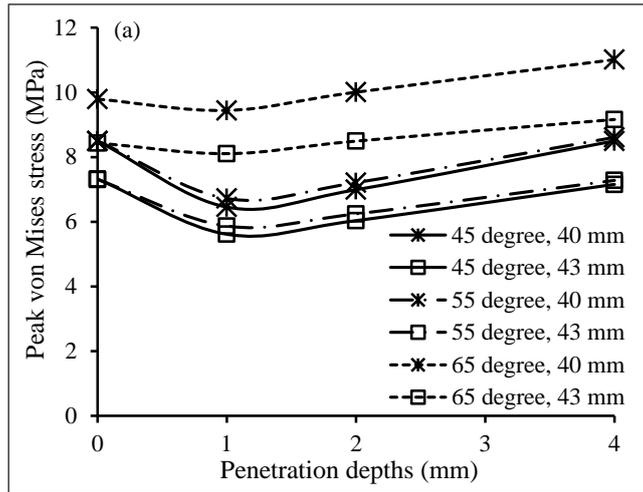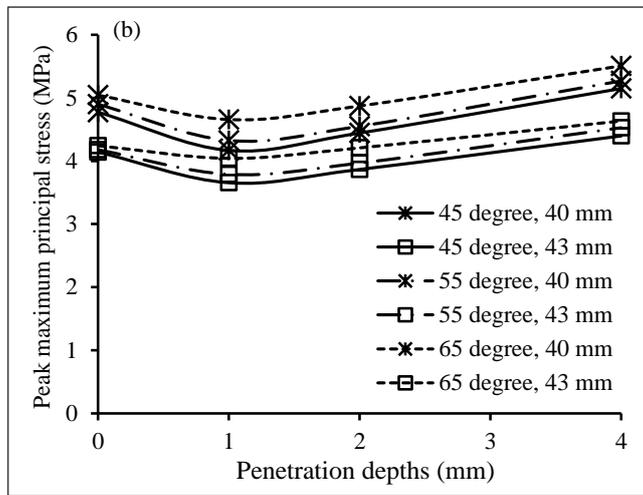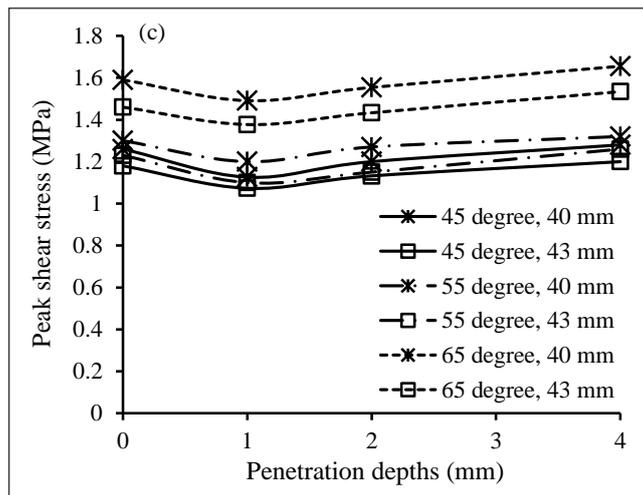
530   Fig. 7

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Fig. 8

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543   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

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