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The contact mechanics and occurrence of edge loading in modular metal-on-polyethylene total hip replacement during daily activities

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25 **Abstract:** The occurrence of edge loading in hip joint replacement has been associated with
26 many factors such as prosthetic design, component malposition and activities of daily living.
27 The present study aimed to quantify the occurrence of edge loading/contact at the articulating
28 surface and to evaluate the effect of cup angles and edge loading on the contact mechanics of
29 a modular metal-on-polyethylene (MoP) total hip replacement (THR) during different daily
30 activities. A three-dimensional finite element model was developed based on a modular MoP
31 bearing system. Different cup inclination and anteversion angles were modelled and six daily
32 activities were considered. The results showed that edge loading was predicted during normal
33 walking, ascending and descending stairs activities under steep cup inclination conditions
34 ($\geq 55^\circ$) while no edge loading was observed during standing up, sitting down and knee
35 bending activities. The duration of edge loading increased with increased cup inclination
36 angles and was affected by the cup anteversion angles. Edge loading caused elevated contact
37 pressure at the articulating surface and substantially increased equivalent plastic strain of the
38 polyethylene liner. The present study suggested that correct positioning the component to
39 avoid edge loading that may occur during daily activities is important for MoP THR in
40 clinical practice.

41

42 **Keywords:** edge loading, activities, metal-on-polyethylene, contact mechanics, cup angles

43

44 **1 Introduction**

45 Despite the successful outcomes and encouraging long-term clinical performance of hip joint
46 replacement, the clinical complications and unexpected failure of the prostheses linked to
47 edge loading are causing concerns [1-5]. The edge loading, usually described as the contact
48 of the femoral head on the edge of the acetabular component, was observed in many retrieval
49 components and usually identified as the condition under which the maximum depth of
50 penetration of the wear scar occurs at the rim of the cup or the wear scar has a distinct
51 boundary in retrieval studies [6-8]. In numerical studies, true edge loading was specified and
52 defined as the condition where the contact patch between the acetabular and femoral
53 components extends over the rim of the cup [9, 10].

54 Edge loading can reduce the tribological performance and may cause unexpected clinical
55 problems [3,6,11-14]. In metal-on-metal (MoM) hip replacement, edge loading can produce
56 accelerated local and overall articulation wear [15, 16] and lead to metallosis, adverse peri-

57 prosthetic tissue reactions such as pseudotumours [2,6,17]. In ceramic-on-ceramic (CoC)
58 articulations, edge loading has been associated with accelerated articulation wear, stripe wear
59 on either the femoral or acetabular component, and in some situation, squeaking and fracture
60 of components [11,18-20]. For metal-on-polyethylene (MoP) and ceramic-on-polyethylene
61 (CoP) combinations, although in vitro experimental studies indicated that edge loading
62 induced by steep cup inclination and lateral microseparation did not increase the wear of
63 prostheses compared to that without edge loading [21,22], finite element (FE) studies have
64 shown that substantial increase in the stresses and plastic strain of polyethylene component
65 were predicted for the hip prosthesis under edge loading conditions [13], which may
66 contribute to subsequent fatigue and fracture. Therefore, persistent and sustained efforts to
67 reduce or prevent edge loading should be still made for hard-on-soft articulations.

68 It has been recognized that the occurrence of edge loading on the hip joint replacement is
69 related to many factors such as prosthetic design [10,23], malposition of components
70 [9,14,16], impingement and dislocation [24,25], and patient activities [17,26]. Particularly,
71 the malposition of the components has been recognized as an important factor causing the
72 poor outcome of hip joint replacement. Although a golden “safe zone” with cup inclination of
73 $40^{\circ}\pm 10^{\circ}$ and anteversion of $15^{\circ}\pm 10^{\circ}$ was recommended and accepted by most surgeons [27], a
74 large variation in the cup orientation was observed in clinical practice [28, 29]. The adverse
75 effect of malposition of acetabular component on the performance and outcome of the hip
76 joint replacement was also reported [29,30]. Schmalzried et al. conducted a study to
77 investigate the relationship between the design, position and wear of acetabular component
78 and the development of pelvic osteolysis [30]. They demonstrated that the osteolysis of the
79 ilium was associated with a lateral opening of the acetabular component of more than 50
80 degrees. Kennedy et al. reviewed two groups of total hip arthroplasties with mean inclination
81 angles of 61.9° and 49.7° and concluded that although the postoperative Mayo clinical hip
82 score was similar for the two groups, the group with a mean inclination of 61.9° had higher
83 rate of recurrent dislocation, osteolysis, wear asymmetry and acetabular component
84 migration, compared to the group with a mean inclination of 49.7° [29]. Therefore, the
85 malposition of components on edge loading and performance of hip joint replacement should
86 be examined.

87 The important contribution of daily activity patterns on the occurrence of edge loading has
88 been demonstrated in a number of previous studies [17,26,31]. Mellon et al. investigated the
89 effect of function activities (i.e. level walking and stair descent) and cup orientation on the

90 edge loading and contact stress of MoM hip resurfacing using FE method and a combination
91 of the computed tomography (CT) and three-dimensional lower limb motion capture data
92 [26]. They suggested that steep cup inclination can cause edge loading and that individual's
93 activity patten can compensate or even override the influence of steep cup inclination and
94 prevent edge loading. Using the same method, Kwon et al. quantified the duration and
95 magnitude of in vivo edge loading during functional activities (i.e. level walking, stair
96 climbing and rising from a chair) in MoM hip resurfacing arthroplasty with and without
97 pseudotumours [17]. They indicated that edge loading in MoM hip resurfacing with
98 pseudotumours (which was associated with higher inclination and anteversion angles)
99 occurred with significantly longer duration and greater magnitude of force compared to that
100 without pseudotumours during daily activities. A study conducted by von Arkel et al. showed
101 that the prevalence of posterior edge loading can be reduced by introducing abduction to
102 activities that require deep flexion such as rising from a chair and stooping [31]. These
103 studies have demonstrated the important contribution of patient's daily activities on the edge
104 loading in total hip replacement (THR). However, these studies were based on in vivo
105 evaluation and therefore the edge loading was roughly evaluated by using either the distance
106 or angle between the hip contact force vector and acetabular cup edge vector. In this case, the
107 magnitude of loading and deformation of the component were not considered in these studies.
108 The aims of the present study were, firstly, to determine whether edge loading occurred, the
109 duration of edge loading occurrence and the specific instances over which edge loading
110 occurred during different daily activities under different cup orientation conditions, and
111 secondly, to investigate the effect of cup orientations and edge loading on the contact
112 mechanics of a modular MoP THR during different daily activities using FE method.

113

114 **2 Materials and methods**

115 A typical modular MoP total hip system, consisting of metallic acetabular shell, polyethylene
116 liner and metallic femoral head, was analysed. The inside of the acetabular shell is comprised
117 two distinct regions: the central dome region and the locking mechanism. The central dome
118 region covers approximately 140 degrees of the interior of the shell, providing backside
119 support to the liner. Peripheral to the dome is the locking mechanism, which extends to the
120 face of the acetabular shell. The polyethylene liner is mechanically locked with the acetabular

121 shell via the locking mechanism, forming two areas between the acetabular shell and
122 polyethylene liner: the dome spherical region and equatorial region, as shown in Fig. 1.

123 The nominal diameters of the femoral head and inner surface of polyethylene liner were 36
124 mm and 36.6 mm respectively, giving a radial clearance of 0.3 mm between the femoral head
125 and polyethylene liner. The radii of the central dome region of the acetabular shell and outer
126 surface of the polyethylene liner were 24.14 mm and 24 mm respectively, giving a gap of
127 0.14 mm between the acetabular shell and polyethylene liner at the central dome region
128 (dome spherical region). The outer diameter of the acetabular shell was 56 mm. A polar
129 fenestration with radius of 10 mm was considered in the central dome region of the
130 acetabular shell.

131 A three-dimensional FE model was developed to simulate the implantation of the modular
132 MoP total hip system into a hemi-pelvic bone model (Fig. 1). The hemi-pelvic bone model
133 consisted of a cancellous bone region surrounded by a uniform cortical shell with thickness of
134 1.5 mm [32]. The acetabular subchondral bone was assumed to have been reamed completely
135 prior to implantation.

136 All the materials in the FE model were modelled as homogenous, isotropic and linear elastic
137 except the polyethylene liner which was modelled as non-linear elastic-plastic behaviour with
138 the plastic stress-stain constitutive relationship showing in Fig. 2 [33,34]. The femoral head
139 was modelled as a rigid body as the elastic modulus of the metallic femoral head is about 200
140 times that for polyethylene liner. The mechanical properties for the materials are presented in
141 Table 1. The FE model comprised approximately 92,000 elements, including triangular shell
142 elements for the cortical bone with element sizes less than 3 mm, tetrahedral elements for the
143 cancellous bone with element sizes less than 3 mm, hexahedral and wedge elements for the
144 prosthetic components with element sizes less than 0.8 mm and 0.3 mm respectively. Mesh
145 converge studies were conducted for the FE model under normal walking activity under cup
146 inclination angle of 75° and anteversion angle of 0° , an assumed extreme condition under
147 which the polyethylene liner was assumed to have the worst mechanical behaviour with
148 respect to the contact pressures, von Mises stresses and plastic strain. The results showed that
149 when the element size was reduced by half, the change in any of the parameters of interest
150 was within 5%.

151 A sliding contact formulation was applied both on the articulating surface between the
152 femoral head and polyethylene liner and at the interface between the acetabular shell and

153 polyethylene liner, with friction coefficients of 0.083 and 0.15 respectively [35,36]. The
154 nodes situated at the sacroiliac joint and about the pubic symphysis were fully constrained.
155 All relative movements were prevented between the pelvic bone and the acetabular shell,
156 simulating a situation where the porous sintered coating and in-grown bone were well
157 bonded. The centre of the femoral head was constrained in rotational degrees of freedom and
158 allowed to move freely along the translational free degrees of freedom to allow self-
159 alignment. The validation of the FE model was presented in a previous study, which
160 demonstrated that good agreements of contact areas at the articulating surface were obtained
161 between the FE predictions and experimental measurements using Leeds Prosim hip joint
162 simulator [34].

163 The physiological loadings of six different human activities, which were measured in vivo
164 previously using an instrumented total hip prosthesis [37], were applied to the FE model.
165 These activities were as follows: normal walking (NW), ascending stairs (AS), descending
166 stairs (DS), standing up (SU), sitting down (SD) and knee bending (KB). In order to consider
167 the specific direction and orientation of the forces, the three components of the resultant hip
168 joint forces relative to the pelvis coordinate system in the in vivo study [37] were exported
169 and discretized into 22 or 23 steps, which were then applied directly to the centre of the
170 femoral head in the FE model in a quasi-static manner, as shown in Fig 3. At this case, the
171 global coordinate system in the FE model was assumed to be aligned with the pelvis
172 coordinate system in the in vivo study [37]. A total of 20 orientations of cup angles were
173 considered, with inclination angles varying between 35° and 75° and anteversion angles
174 varying between 0° and 30°, both in 10° increments. The FE analysis was performed using
175 ABAQUS software package (Version 6.9; Dassault Systèmes Simulia Corp., Providence, RI,
176 United States). Edge loading at the articulating surface was detected and evaluated at each
177 instance during the whole cycle of these activities. In the present study, edge loading was
178 defined to occur when the contact patch between the femoral head and polyethylene liner
179 extends over the rim of the liner, as shown in Fig. 4.

180

181 **3 Results**

182 **Contact pressures distribution during gait**

183 Fig. 5 shows the distribution and peak value of contact pressures on the articulating surface of
184 the polyethylene liner with different cup inclination and anteversion angles at instance of
185 17% gait of normal walking activity.

186 Generally, the areas of the contact patch were located about the superior region of the liner
187 and shifted toward the superior edge as inclination angle increased. The peak contact pressure
188 was located at the dome spherical region at low cup inclination conditions (i.e. 35° and 45°)
189 and moved to the equatorial region when the inclination angle was increased to 75°. Edge
190 loading started to occur when the cup inclination angle increased to 65°.

191 **Edge loading**

192 The duration of edge loading and specific instances of cycle at which edge loading occurred
193 during different activities as a function of cup angles are shown in Fig. 6.

194 Edge loading was predicted at some instances of cycle during normal walking, ascending and
195 descending stairs activities under steep cup inclination angle conditions ($\geq 55^\circ$). No edge
196 loading was predicted for standing up, sitting down and knee bending cases for all cup angles
197 considered. For normal walking and ascending stair cases, the combination of steep cup
198 inclination and low anteversion was more likely to cause edge loading. For example, for
199 normal walking activity, the proportion of gait cycle when edge loading occurred increased
200 from 5% (at specific instances of 50-55% of gait cycle) to 50% (at specific instances of 10-
201 60% of gait cycle) as cup inclination angles increased from 55° to 75° with anteversion of 0°.
202 With cup inclination of 65°, the proportion of gait cycle when edge loading occurred
203 decreased from 40% to 13% when the cup anteversion angles increased from 0° to 30°. In
204 contrast, for descending stair activity, the combination of steep cup inclination and high
205 anteversion tended to induce edge loading.

206 **Effect of activities, cup angles and edge loading on contact mechanics**

207 The activities and cup angles were found to have a synergistic effect on the peak contact
208 pressure at the articulating surface and equivalent plastic strain of the liner (Fig. 7 and 8).
209 Edge loading caused elevated peak contact pressure at the articulating surface and marked
210 increase of peak equivalent plastic strain of the polyethylene liner (Fig. 7, 8 and 9). For
211 normal walking, ascending and descending stairs activities, the cup inclination angles had
212 marked effect on the peak contact pressure and equivalent plastic strain while the cup
213 anteversion angles had minor effect. Considering the cup anteversion, the peak contact
214 pressure over the whole cycle firstly decreased by approximately 7%-12%, 5%-9% and 7%-

215 14% for normal walking, ascending stair and descending stair activities respectively when the
216 cup inclination angle increased from 35° to 55°, and then increased by about 18%-26%, 22%-
217 28% and 27%-33% respectively for the three activities when the cup inclination angle
218 increased to 75°, where edge loading occurred (Fig. 7). Correspondingly, the peak equivalent
219 plastic strain over the whole cycle firstly decreased by approximately 31%-53%, 13%-21%
220 and 15%-28% when the cup inclination increased from 35° to 45° and then increased by
221 about 234%-306%, 179%-231% and 178%-213% when the cup inclination increased to 75°
222 for the three activities respectively.

223 In contrast, for standing up, sitting down and knee bending activities, the cup anteversion
224 angles were found to have dominated effect on the peak contact pressure and equivalent
225 plastic strain. Considering the cup inclination, the peak contact pressure and equivalent
226 plastic strain over the whole cycle increased by approximately 14%-24% and 88%-164%,
227 2%-21% and 57%-148%, 4%-12% and 56%-138% for standing up, sitting down and knee
228 bending activities respectively when the cup anteversion increased from 0° to 30°.

229

230 **4 Discussion**

231 Edge loading as an adverse condition that could cause unexpected clinical problems has
232 attracted more and more attentions in biomechanics fields [38,39]. The factors that may lead
233 to edge loading have been recognized and were generally associated with the component
234 positions (i.e. cup angles, head offset/microlateralisation), prosthetic design (i.e. radial
235 clearance, cup coverage), impingement and activities. The contribution and effect of
236 component malposition, prosthetic design, impingement and dislocation on the edge loading
237 of hip replacement have been investigated in a number of previous studies [9,10,23-25,40-
238 42]. The primary purposes of the present study were therefore to investigate the effect of cup
239 orientations and daily activities on the contact mechanics and occurrence of edge loading for
240 a modular MoP THR. The duration of edge loading and instances of cycle at which edge
241 loading occurred during six daily activities were evaluated. To the authors' acknowledge, this
242 was the first to quantify the duration and period of time of true edge loading in THRs during
243 different daily activities, by considering the deformation of pelvic bone and components.

244 The FE simulations showed that an individual's activity patterns played an important role on
245 the occurrence of edge loading in MoP THR. For the THR considered in the present study,
246 edge loading occurred at some instances during normal walking, ascending and descending

247 stairs activities under steep cup inclination conditions. With increased cup inclination angles,
248 the duration and period of time over which the hip experienced edge loading increased. These
249 were supported by an in vivo study to evaluate edge loading in MoM hip resurfacing patients
250 with and without pseudotumours which showed that edge loading in patients with well-
251 functioning MoM hip resurfacing arthroplasty was observed during functional activities and
252 that edge loading in the hips with pseudotumours (which was associated with higher cup
253 inclination) occurred for a significantly longer period of time compared to that without
254 pseudotumours [17]. The present study also showed that the duration and period of time of
255 edge loading was activity-dependent, with the longest duration of edge loading being
256 observed for normal walking activity. No edge loading was predicted for standing up, sitting
257 down and knee bending activities. These observations, however, were found to be different
258 from the previous in vivo study which indicated that edge loading also occurred for rising
259 from or sitting down to chair activity [17]. A retrieval study conducted by Esposito et al also
260 demonstrated both anterior and posterior edge loading in retrieval ceramic components and
261 they assumed that posterior edge loading may occur during activities such as climbing stairs
262 or rising from a chair [43]. The different conclusions between the present study and the in
263 vivo and retrieval studies may be due to several reasons. Firstly, in vivo study, edge loading
264 was defined to occur when the locus of the force vector intersection with the acetabular
265 component was located within the areas where the distance to the edge of the component was
266 no larger than 10% of the component radius, while in the present study, edge loading was
267 defined as the case when the contact patch extends over the rim of the component. The
268 limitation of the in vivo study was that although the force vector for the rising up/sitting down
269 activities was located in the edge loading zone defined in the in vivo study for a longer period
270 of time, the force magnitude was smaller compared to that in normal walking, ascending and
271 descending stairs activities, leading to a smaller contact patch at the bearing surface of the
272 component. Therefore, if the radius of the contact patch was smaller than 10% of the
273 component radius, edge loading would not occur. However, at this case, edge loading was
274 assumed to still occur in the in vivo study. Secondly, the different design of prosthesis
275 considered in the present study (MoP) and the in vivo (MoM) and retrieval (CoC) studies may
276 be an important factor causing the different conclusions. In the present study, the radial
277 clearance between femoral head and polyethylene liner was 0.3 mm. If a smaller clearance is
278 considered, the contact stresses will be decreased and the contact areas will be increased. At
279 this case, the contact patch will potentially extend over the rim of the polyethylene liner,
280 causing posterior edge loading for rising up/sitting down activities. In fact, in the present

281 simulation, for most instances of rising up/sitting down activities, the contact patch was prone
282 to locating at the posterior area of the bearing surface, having the potential to cause posterior
283 edge loading. Therefore, the effect of prosthetic design such as radial clearances and cup
284 coverages on the occurrence of edge loading will be examined in future studies. Thirdly, the
285 posterior edge loading observed in the retrieval study may be caused by some adverse
286 conditions such as impingement of the components, which has been reported to be common
287 for MoP THR in retrieval studies [28,44]. However, the adverse condition of impingement
288 was not considered in the present study.

289 Previous studies have shown that the cup inclination of no larger than 45° is best for
290 achieving stability and preventing wear [45,46]. The present study supported this conclusion
291 that no edge loading occurred when the cup inclination angle was no larger than 45° for all
292 the activities and cup anteversion angles considered. In addition, the cup anteversion was
293 found to have a crucial effect on the duration and occurrence of edge loading as well. For
294 example, under a steep cup inclination angle of 65° , the duration of occurrence of edge
295 loading during normal walking was over 40% gait cycle under anteversion angle of 0° , which
296 reduced to less than 15% gait cycle under anteversion angle of 30° . Edge loading was most
297 likely to occur at the instances between 45-55%, 15-20% and 90-95% cycle time for normal
298 walking, ascending and descending stairs activities respectively. This was a result of the
299 synergistic effect between the force vector and magnitude. Indeed, in a paper to investigate
300 the effect of motion patterns on edge-loading of MoM hip resurfacing, Mellon et al.
301 suggested that the force vector at the instance of 60% gait cycle was closer to the edge of
302 component than any other time during the stance phase of gait [26].

303 The analysis of the effect of cup angles on the contact pressures at the articulating surface
304 showed that mild increase of the cup inclination angle resulted in decreased peak contact
305 pressure at the articulating surface of the modular MoP THR for normal walking, ascending
306 and descending stairs activities, which was found to be different from the non-modular THR
307 [33,45]. This was probably due to the factor that at lower cup inclination condition (i.e. 35°),
308 the contact area was mainly located in the dome spherical region of the polyethylene liner in
309 modular MoP THR. When the cup inclination angles increased (i.e. 45° , 55°), the contact area
310 moved to the transition area between the dome spherical region and equatorial region. The
311 different deformation of the polyethylene liner due to the different stiffness of support behind
312 the liner would cause enlarged contact areas at this transition region, leading to decreased
313 contact pressures [47,48]. When the cup inclination angle increased further (i.e. 75°), edge

314 loading would occur and the contact pressures increased. For all cup angles conditions and
315 activities considered, plastic deformation of the polyethylene liner was predicted. Similarly,
316 the equivalent plastic strain of the polyethylene liner was first increased and then decreased
317 with increased cup inclination angles.

318 It is well known that the cup inclination angles had a marked effect on the contact mechanics
319 and stability of hip joint replacement under both normal and adverse conditions
320 [13,33,45,47]. The present study demonstrated that for normal walking, ascending and
321 descending stairs activities, the cup inclination angles had a leading effect on the contact
322 pressures at the articulating surface and equivalent plastic strain of the polyethylene liner,
323 while for standing up, sitting down and knee bending activities, the cup anteversion had
324 dominated impact. Therefore, it is suggested that the importance of cup anteversion should be
325 considered and recognized during the positioning of cup component in clinical practice.

326 The FE analysis also showed that edge loading caused elevated contact pressures at the
327 articulating surface and equivalent plastic strain in the components, which was consistent
328 with previous studies [13,14]. In particular, there was a substantial increase in the equivalent
329 plastic strain when the cup inclination increased from 55° to 65° and from 65° to 75° for
330 normal walking, ascending and descending stairs activities, where edge loading occurred.
331 This indicated that obvious plastic deformation would occur under these conditions, as
332 observed in previous in vitro study [21]. The amplified plastic deformation could potentially
333 induce creep and fatigue of the liner [49,50], and also pitting and delamination of the surface
334 at this area, leading to fatigue damage and fracture of the component [51]. Therefore, it is
335 indicated that the positioning of the component is important clinically to avoid severe plastic
336 deformation of the component and that lower cup inclination angle remains a
337 recommendation for implant positioning of the modular THR.

338 There are several limitations to the present study. First, the muscle and ligament surrounding
339 the hip were not considered in the present study, which was proved to play an important role
340 in the stability of hip replacements [52]. Previous study has shown that the muscles inserted
341 into the distal femur, patella or tibia can contribute to edge loading of well-positional cup
342 [31]. Therefore, a large-scale computational model that integrate the FE model and
343 musculoskeletal dynamic model could be developed for getting a better understanding of
344 edge loading during different daily activities in future studies. Second, only six activities
345 were considered in the present study whereas a broad variety of challenge maneuver ensue in
346 activities of daily living which would cause adverse complications such as impingement and

347 dislocation [53]. However, the activities considered in the present study did represent the
348 most frequent activities for human daily living [37]. Third, homogeneous, isotropic and linear
349 material properties for the bone and uniform thickness of cortical bone were assumed in the
350 present study. However, a real bone should have a non-homogenous, anisotropic property
351 [54], and previous studies have shown that the thickness of the cortical bone layer and the
352 material properties of the bone were site-dependent and bone density-dependent [55,56]. The
353 effect of bone properties on the results should be evaluated and addressed in the future
354 studies. Moreover, lubrication may play an important role in the occurrence of edge loading
355 which was not considered in the present study. However, a recent study to investigate the
356 contact mechanics and lubrication of ceramic-on-metal total hip replacements demonstrated
357 that the profiles and magnitude of the film pressures calculated using elastohydrodynamic
358 lubrication (EHL) theory was closely similar to those of the dry contact pressures calculated
359 using FE modelling [57]. Finally, the femoral head was assumed to be located perfectly
360 within the liner during all activities in the FE simulation. However, in deep flexion activities
361 such as standing up or sitting down activities, there is possibilities that impingement of the
362 components occurs, causing a posterior subluxation of the femoral head and posterior edge
363 loading in the acetabular liner. These were not simulated in the present study.

364 Despite these limitation listed above, the present study suggested that edge loading would
365 occur during some of the functional daily activities such as normal walking,
366 ascending/descending stairs under steep cup inclination conditions. Edge loading induced by
367 these daily activities and steep cup inclination can result in elevated contact pressures at the
368 articulating surface and equivalent plastic strain in the component for the modular MoP THR.
369 Therefore, it is suggested that clinically it is important to optimise the orientation of the
370 components in hip joint replacements to avoid edge loading that may occur during activities
371 of daily living.

372

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379 **Conflict of interest**

380 John Fisher is a consultant to DePuy Synthes Joint Reconstruction.

381

382 **Ethical approval**

383 Not required.

384

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534 **List of figure captions:**

535 **Fig. 1** The FE modelling and boundary conditions, and cross-section of the modular MoP
536 THR showing the detailed structure and features.

537 **Fig. 2** The plastic stress-strain relation for the polyethylene liner [33,34].

538 **Fig. 3** Resultant hip joint forces during normal walking. The resultant force was converted to
539 three components (F_X , F_Y , F_Z) and computed as $F = \sqrt{F_X^2 + F_Y^2 + F_Z^2}$. During the simulation
540 process, the resultant hip joint force was discretized into 23 steps.

541 **Fig. 4** The definition of edge loading in MoP THR in the present study. Left: edge loading
542 did not occur as the contact patch was within the inner surface of the liner; right: edge
543 loading occurred as the contact patch extended over the rim of the liner.

544 **Fig. 5** The distribution and peak value of the contact pressures (MPa) on the articulating
545 surface of the polyethylene liner as a function of cup inclination and anteversion angles at
546 17% gait cycle during normal walking activity.

547 **Fig. 6** The duration of edge loading and specific instances at which edge loading occurred on
548 the articulating surface of the liner as a function of cup inclination and anteversion angles
549 during different activities (NW: normal walking, AS: ascending stairs, DS: descending
550 stairs). No edge loading was predicted for standing up, sitting down and knee bending
551 activities.

552 **Fig. 7** The peak contact pressure (MPa) at the articulating surface over the whole cycle as a
553 function of cup inclination and anteversion angles during different activities ((NW: normal
554 walking, AS: ascending stairs, DS: descending stairs, SU: standing up, SD: sitting down, KB:
555 knee bending).

556 **Fig. 8** The peak equivalent plastic strain in the polyethylene liner over the whole cycle as a
557 function of cup inclination and anteversion angles during different activities ((NW: normal
558 walking, AS: ascending stairs, DS: descending stairs, SU: standing up, SD: sitting down, KB:
559 knee bending).

560 **Fig. 9** The maximum contact pressure at the articulating surface of liner during normal
561 walking for different cup inclination angles with cup anteversion angle of 10°. The bold red
562 lines represent the instances when edge loading occurred.

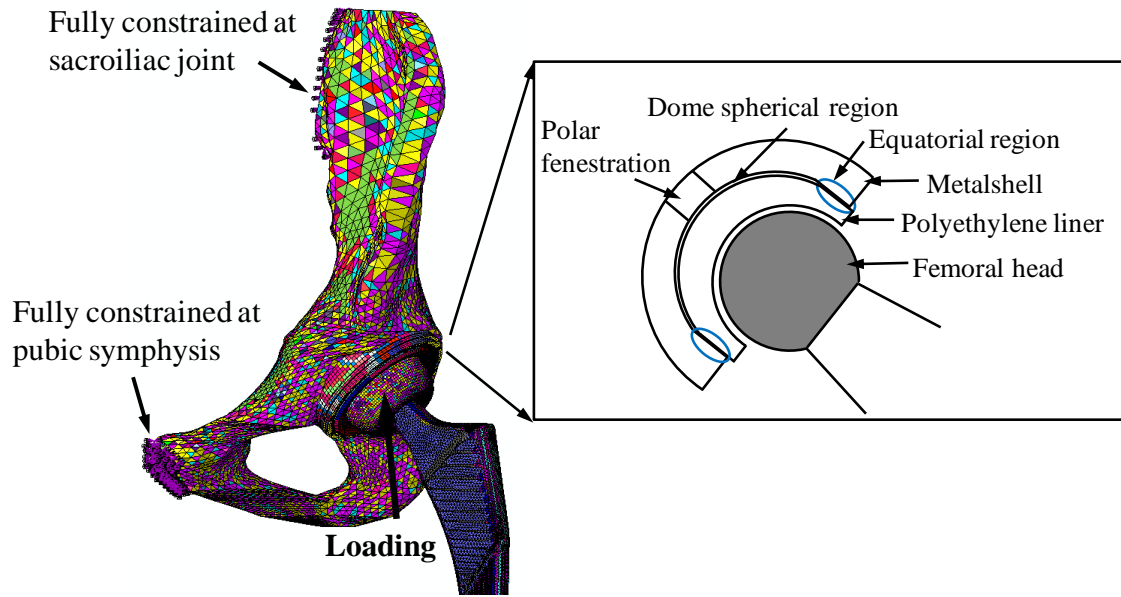
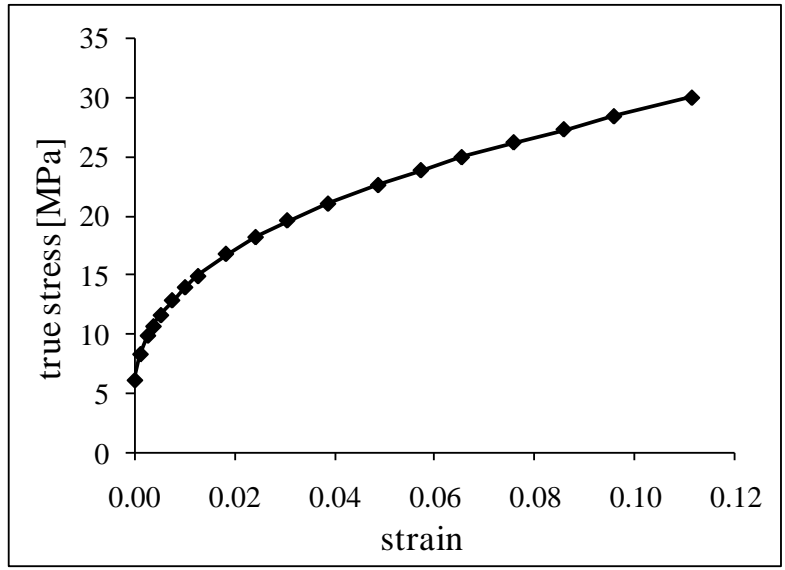


Fig. 1

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Fig. 2

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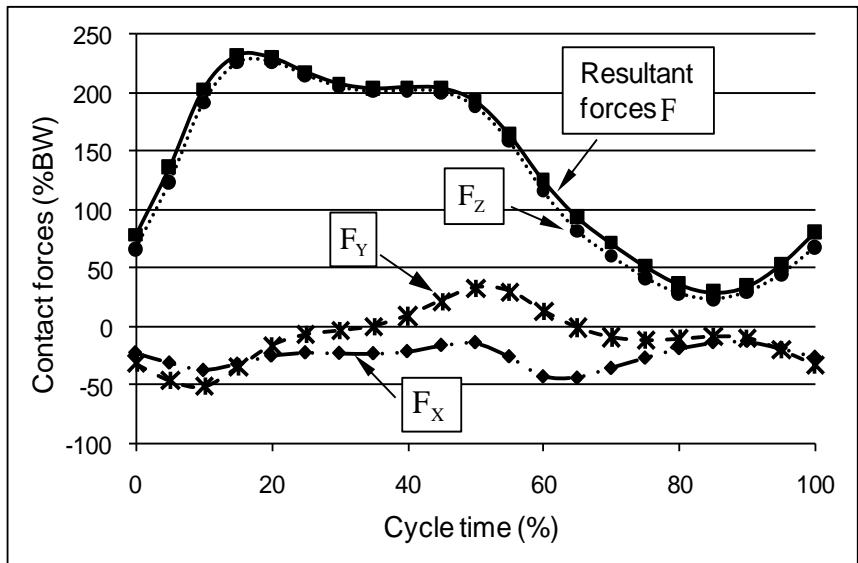


Fig. 3

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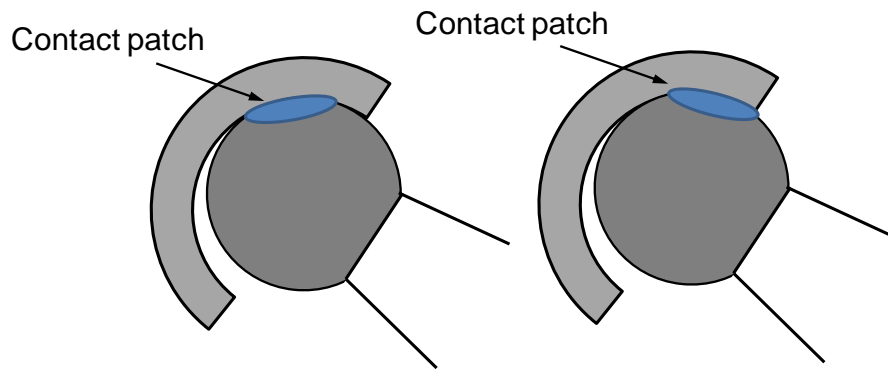
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Fig. 4

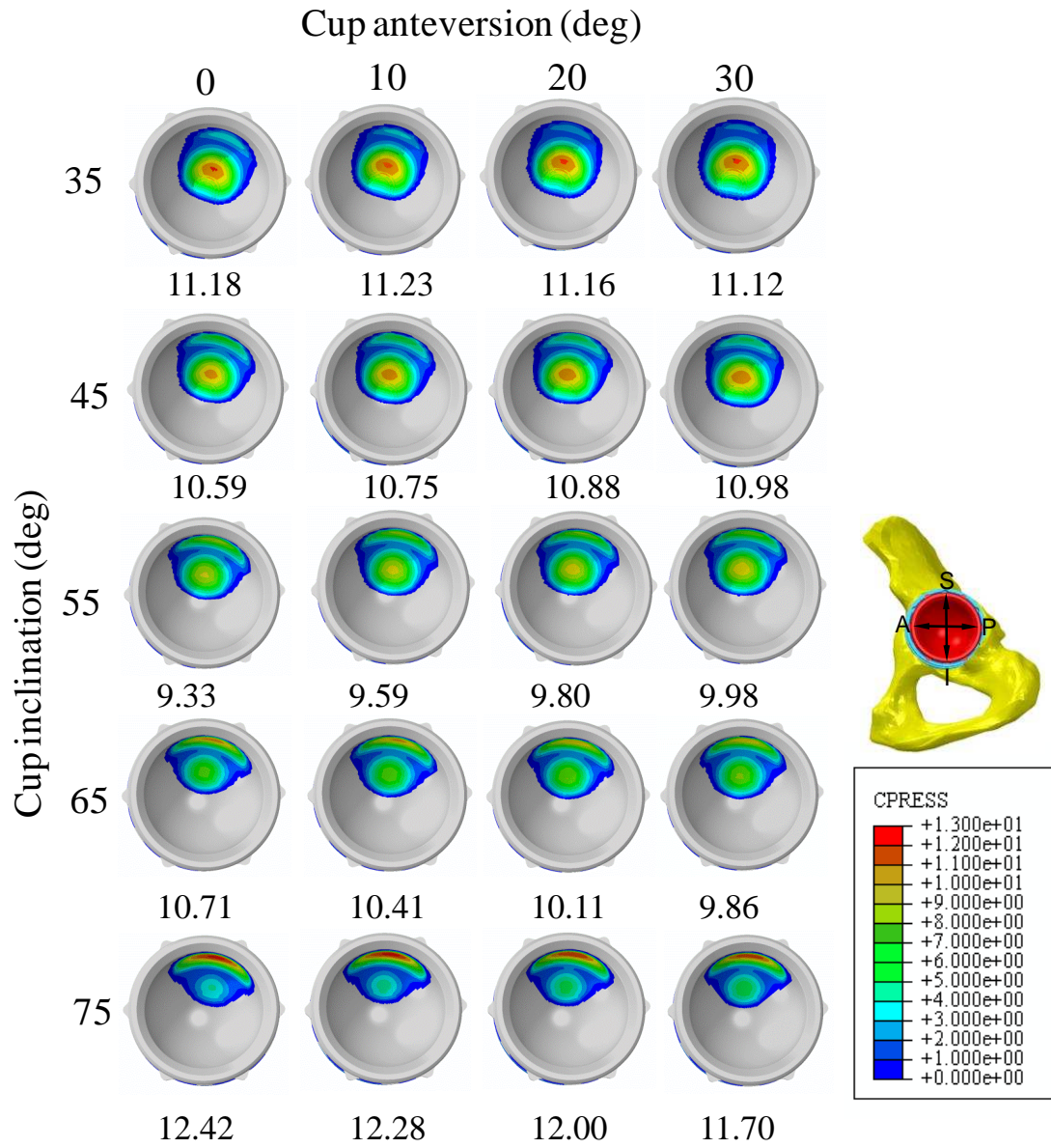
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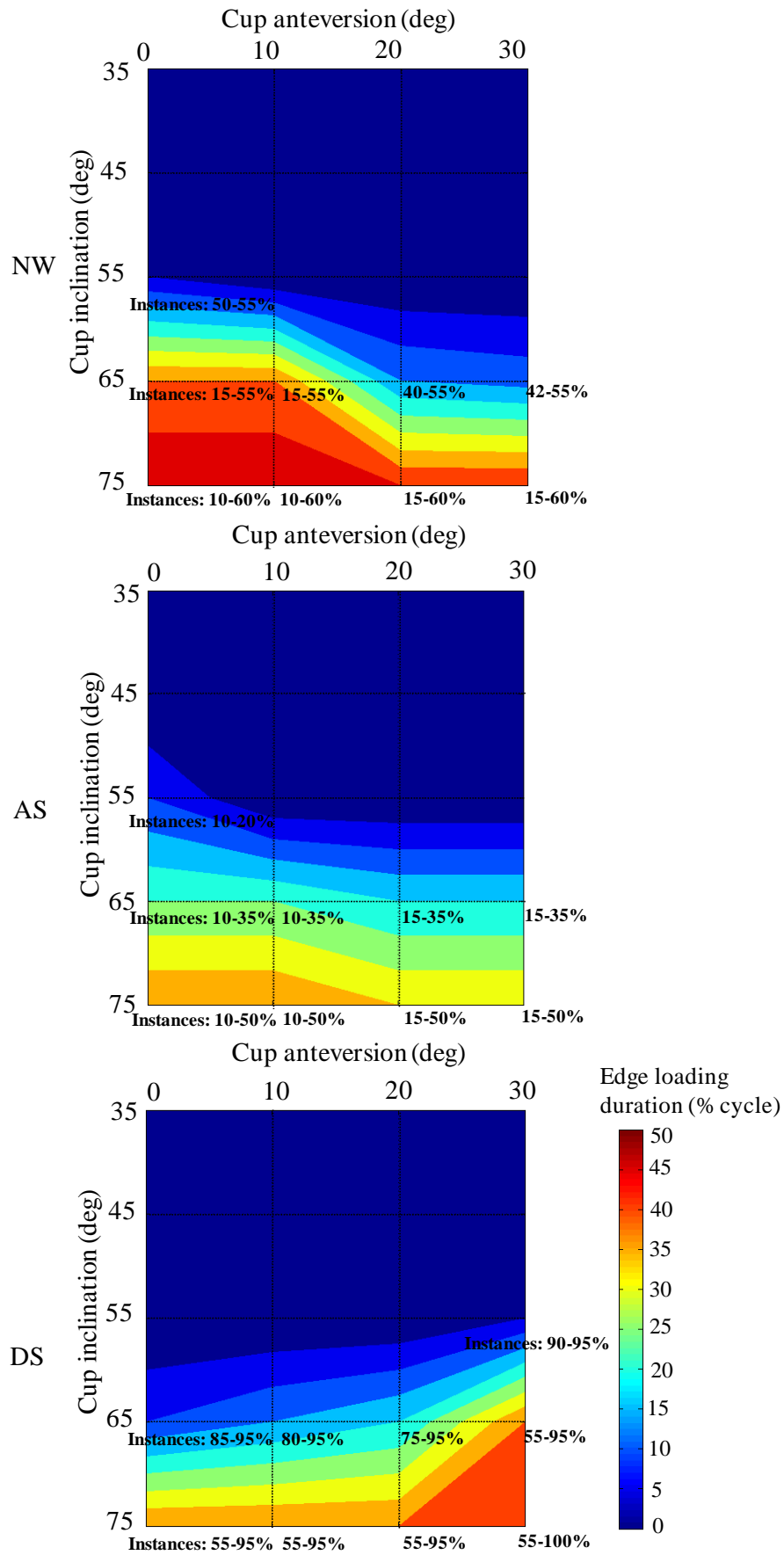
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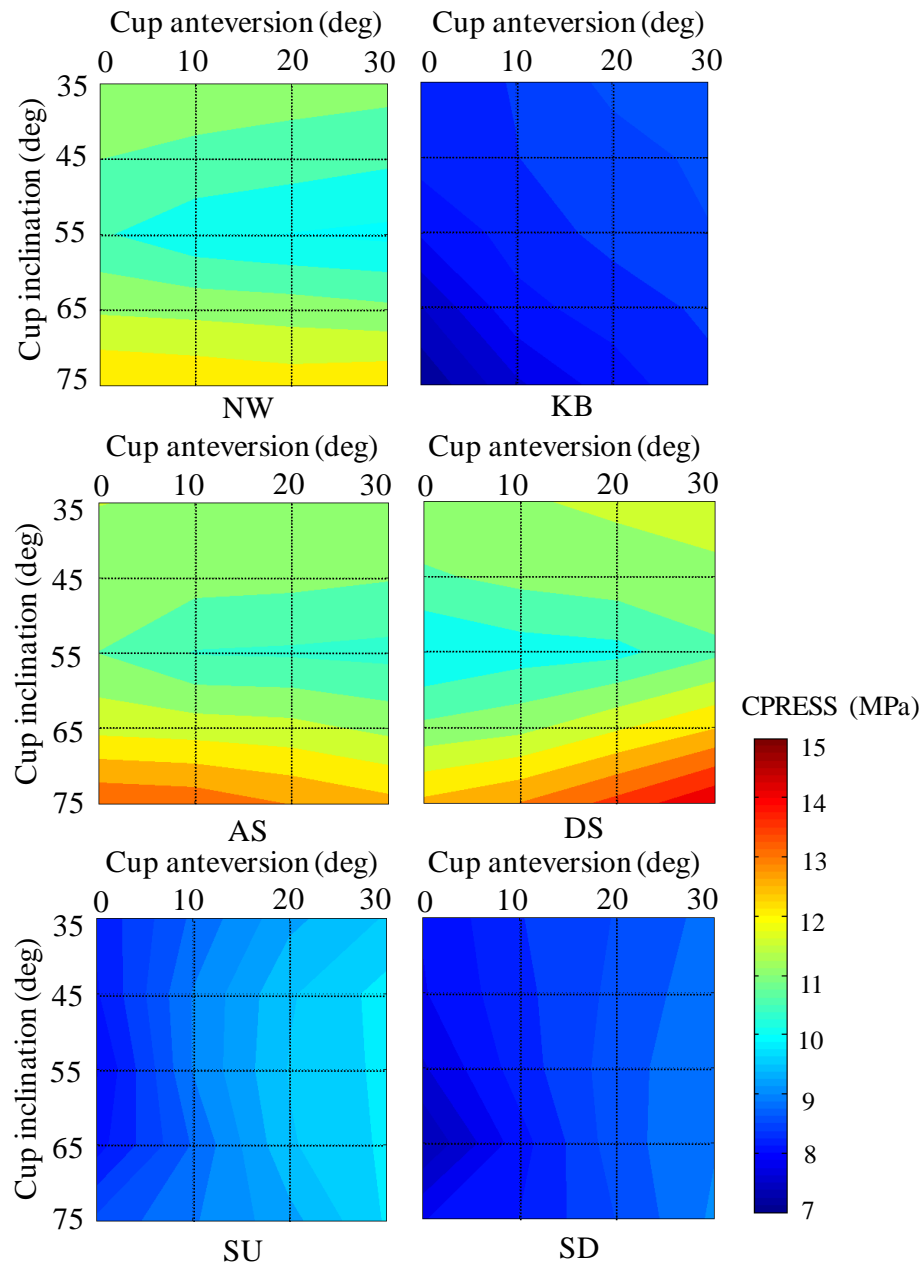
Fig. 5



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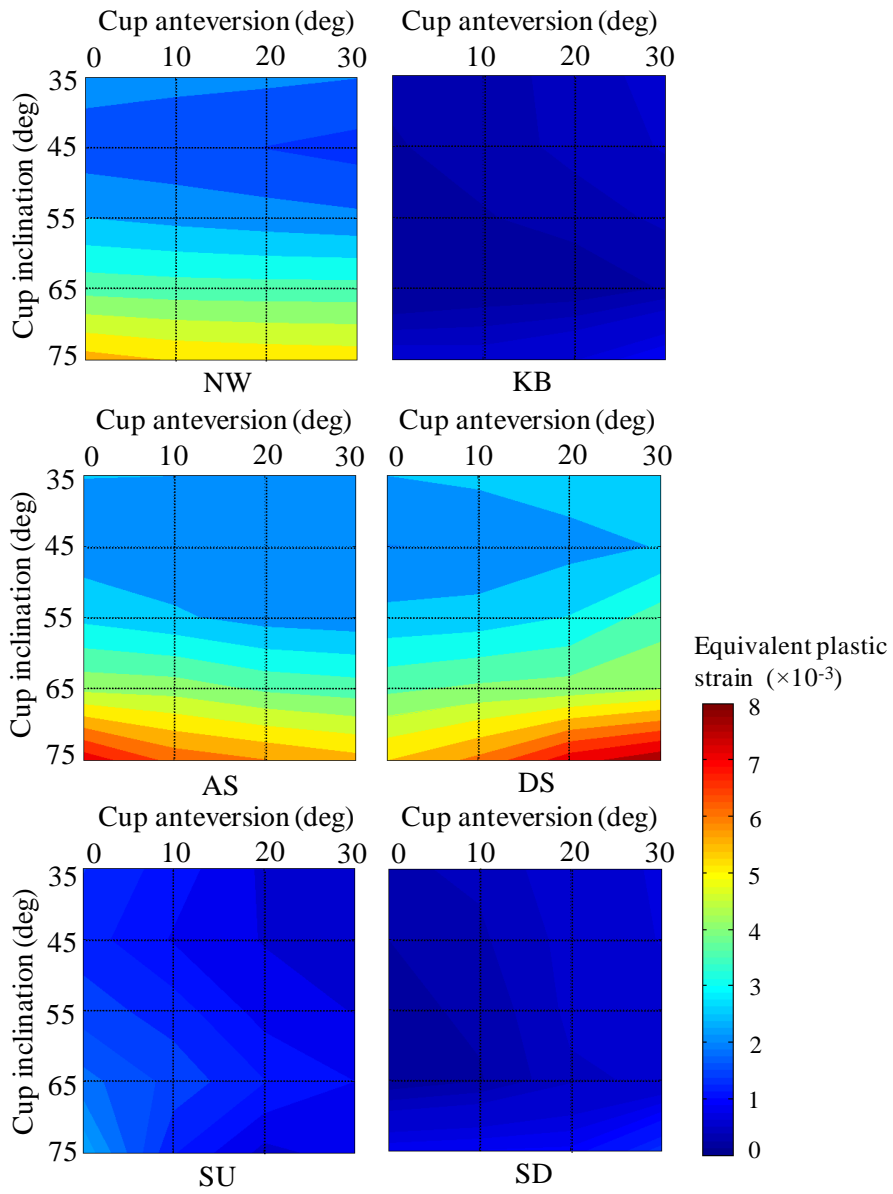


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Fig. 7



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Fig. 8

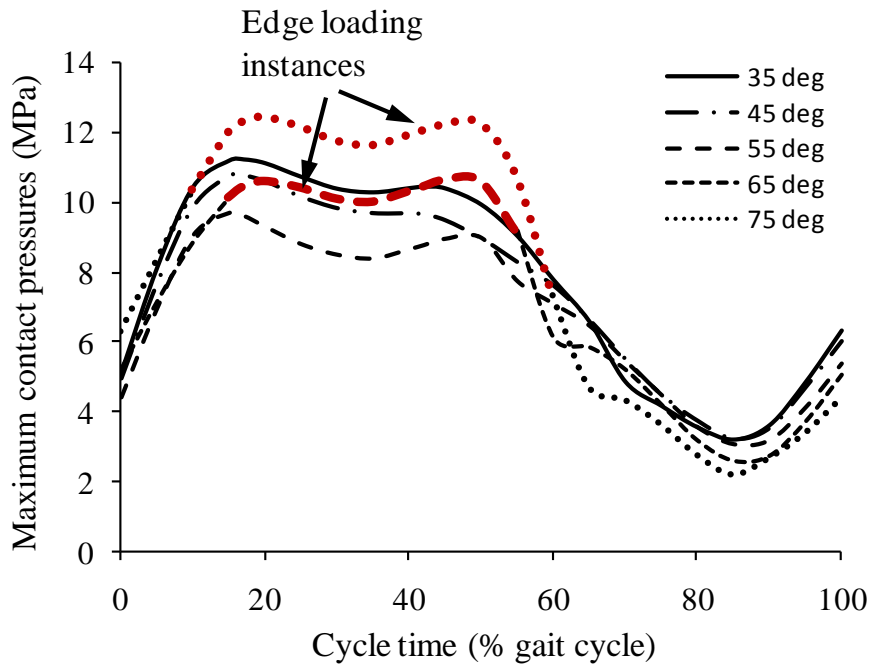


Fig. 9

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640 **Table 1** Material properties for the components used in the present study [32,33,34].

Components	Materials	Young's modulus (GPa)	Poisson's ratio
Polyethylene liner	UHMWPE	1	0.4
Metal shell	Titanium	116	0.25
Cortical shell	Cortical bone	17	0.3
Cancellous bone	Cancellous bone	0.8	0.2

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