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1 Patient Characteristics Affect Hip 2 Contact Forces during Gait 3

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22

23 Running Headline: **Patient characteristics affect HCF**

24 **Abstract**

25

26 **Objective:** To examine hip contact force (HCF), calculated through multibody modelling, in a large
27 total hip replacement (THR) cohort stratified by patient characteristics such as BMI, age and
28 function.

29 **Method:** 132 THR patients undertook one motion capture session of gait analysis at a self-selected
30 walking speed. HCFs were then calculated using the AnyBody Modelling System. Patients were
31 stratified into three BMI groups, five age groups, and finally three functional groups determined by
32 their self-selected gait speed. Independent 1-dimensional linear regression **analyses** were performed
33 to separately evaluate the influence of age, BMI and functionality on HCF, by means of statistical
34 parametric mapping (SPM).

35 **Results:** The mean predicted HCF were comparable to HCFs measured with an instrumented
36 prosthesis reported in the literature. The regression analyses revealed a **statistically** significant
37 positive **linear correlation** between BMI and HCF, indicating that obese patients **are more likely to**
38 experience higher HCF during most of the stance phase, while a **statistically** significant **linear**
39 **correlation** with age was found only during the late swing-phase. Patients with higher functional
40 ability exhibited significantly increased peak contact forces, while patients with lower functional
41 ability displayed a pathological flattening of the typical double hump force profile.

42 **Conclusion:** HCFs experienced at the bearing surface are highly dependent on patient
43 characteristics. BMI and functional ability were determined to have the biggest influence on contact
44 force. Current preclinical testing standards do not reflect this.

45 **Keywords:** Total hip replacement, Hip contact force, Stratification, Biomechanics, Gait

46

47

48

49 **Introduction**

50

51 Total hip replacement (THR) surgery is commonly regarded as one of the most successful elective
52 orthopaedic surgeries of the 20th century ¹. It alleviates pain in patients suffering from debilitating
53 hip osteoarthritis and improves function. However there is some lifetime risk of implants requiring
54 revision, the rates of which are currently 4.4% at 10 years and 15% at 20 years². Epidemiological
55 studies have provided evidence to suggest that patient characteristics, such as age, BMI and gender
56 are important factors in the survivorship of hip implants^{2,3}. One in three patients undergoing THR at
57 < 50 years of age are expected to require revision surgery during their lifetime, with risks of one in
58 five for patients 50 to 59 years, one in ten for patients 60 to 69 years, and one in 20 for patients ≥ 70
59 years ⁴. The revision risk for younger patients is consistently higher than for older patients at all
60 time-points i.e. 5, 10, 15 and 20 years and gender also seems to affect risk ². Men aged younger than
61 70 years old have an increased revision risk compared to female patients, and at the age of 50 years
62 females have a 15% lower chance of revision compared to their male counterparts. BMI also
63 contributes to lifetime revision risk, with obese patients having twice the risk of revision at 10 years
64 compared to healthy weight and overweight patients, and it has been suggested by Culliford *et al.* ⁵
65 that for every unit increase of BMI, there is a 2% increased risk of revision of a THR.

66 The precise reason for these differences in revision rates between patient sub-groups is not clear,
67 however the variations in revision rates suggest that the demands placed on the implant likely differ
68 between patient groups. Due to the relatively small sample sizes typically employed in
69 biomechanical studies of THR cases, few studies have explored how patient characteristics can
70 differentially influence function post THR, and ultimately how those characteristics might affect
71 what demand is placed on the implant.

72

73 In these few studies age and BMI have been shown to influence function in THR patients. In one
74 analysis of a larger sample of patients from multiple retrospective studies, Foucher *et al.*⁶ found that
75 older patients had limited hip sagittal ROM and hip power generation compared to younger patients
76 who recovered better post-operatively. When stratifying gait function by age in a large cohort
77 (n=134) of THR patients, Bennett *et al.*^{7, 8} reported that gait kinematics and kinetics were not
78 influenced by age, except for a reduced ROM exhibited in an 80 years and over age group, a finding
79 also consistently observed in healthy control patients of a similar age range⁹. Foucher *et al.*^{6 10}
80 reported that BMI plays a role in recovery, with higher BMI patients having a reduced hip range of
81 motion (ROM) and hip abductor moment compared to healthy control participants. Furthermore,
82 lower BMI was associated with higher postoperative values of sagittal ROM, adduction moments,
83 and external rotation moments compared to THR patients with a higher BMI.

84

85 As described above, real-world patient function¹⁰ and survivorship of the hip implant² is affected by
86 the characteristics of the patient, although this is not currently reflected in preclinical wear testing
87 standards such as ISO 14242. Current preclinical testing protocols use a stylised waveform vaguely
88 representing a 'standard' THR patient's walking cycle to test the wear properties of the implant. A
89 recent study found that post-operative patient function accounts for 42% to 60% of wear, compared
90 to surgical factors which account for 10% to 33% of wear¹¹, emphasising the importance of
91 understanding how gait varies between different patient groups. No previous studies have tried to
92 understand how patient characteristics affect the absolute forces at the bearing surface, forces
93 which arguably will have the most influence on *in vivo* wear rates. Instrumented implants have been
94 used to calculate contact force at the bearing surface^{12, 13}, however the data available from these
95 implants is limited to small numbers of patients and extrapolating these data to the wider patient
96 population is not appropriate. Modern computational models of the musculoskeletal system can be
97 used to calculate joint contact forces and are becoming increasingly more clinically applicable¹⁴.
98 These models have the capability to calculate accurate joint contact forces in THR patients¹⁵, and

99 can be used to predict and compare contact forces in stratified samples derived from a large patient
100 cohort ¹⁶. The primary aim of this study therefore, was to examine hip contact force (HCF),
101 calculated through multibody modelling, in a large THR cohort when stratified by patient
102 characteristics such as BMI, age and function.

103 **Method**

104

105 ***Patients***

106

107 132 THR patients were recruited into the study through a clinical database of surgical cases.
108 Inclusion criteria for the hip replacement group were; between 1-5 years THR post-surgery, older
109 than 18 years of age, no lower limb joint replaced other than hip joint(s), fully pain free and not
110 suffering from any other orthopaedic or neurological problem which may compromise gait. Ethical
111 approval was obtained via the UK national NHS ethics (IRAS) system and all participants provided
112 informed, written consent.

113

114 ***Data Capture***

115

116 Lower limb kinematics and kinetics were collected using a ten camera Vicon system (Vicon MX,
117 Oxford Metrics, UK) sampling at 100Hz, integrated with two force plates (AMTI, Watertown, MA,
118 USA) capturing at 1000Hz in a 10m walkway. The operated limb (or most recently operated limb, in
119 bilateral cases) was used for analysis. All patients were allowed a familiarisation period prior to
120 completing 3-5 successful trials of each walking condition. A successful trial was defined as a clean
121 foot strike within the boundary of the force plate. The CAST marker set was used to track lower limb
122 segments kinematics in six degrees of freedom, with four non-orthogonal marker clusters positioned
123 over the lateral thighs, lateral shanks and sacrum as described comprehensively elsewhere ^{17, 18}. Six

124 retroreflective markers were positioned on the first, second and fifth metatarsophalangeal joints as
125 well as the malleoli and calcanei. Participants wore a pair of tight-fitting shorts and a vest onto which
126 reflective markers were affixed using double-sided tape at bony anatomical landmarks to determine
127 anatomical joint centres. Before walking trials commenced, a static trial was collected in an
128 anatomical reference position.

129

130 ***Data Processing***

131

132 All markers were labelled and gap-filled using the spline fill function in Vicon Nexus 2.5 (Vicon MX,
133 Oxford Metrics, UK), before the labelled marker coordinates and kinetic data were exported to
134 Visual 3D modelling software (C-Motion, Rockville, USA) for further analysis. Kinematic data were
135 filtered using a low-pass (6Hz) Butterworth filter. Ground reaction force (GRF) data were filtered
136 using a low-pass Butterworth filter (25Hz) and heel strike and toe-off were determined using
137 thresholds (>20N for heel strike and <20N for toe off) from the GRF.

138

139 ***Musculoskeletal modelling***

140

141 Musculoskeletal simulations were performed using commercially available software (AnyBody
142 Modeling System, Version 7.1, Aalborg, Denmark). A recently validated generic musculoskeletal
143 model ¹⁹ was scaled to match the anthropometrics of each patient. The scaling of the model
144 segments was based on the marker data collected during a static trial ²⁰. Marker trajectories and GRF
145 data from each gait trial served as input to an inverse dynamics analysis, based on a 3rd order
146 polynomial muscle recruitment criterion, to calculate muscle forces and HCFs. A total of 494 gait
147 trials were processed and analyzed through the toolkit AnyPyTools ([https://github.com/AnyBody-
148 Research-Group/AnyPyTools](https://github.com/AnyBody-Research-Group/AnyPyTools)).

149 The different components of HCFs, defined in a common femur-based reference frame ¹² were
150 computed for the operated limb over a gait cycle. The data were time-normalized from heel-strike
151 (0%), through toe-off (60%), to heel strike (100%) and interpolated to 1% steps (101 points). An
152 average per patient was then calculated based on the 3-5 trials collected.

153

154 ***Stratification by patient characteristics***

155

156 Patients were stratified by into three groups based on their BMI. BMI scores were calculated as
157 measured weight divided by measured height squared (kg/m^2). The three groups were; healthy
158 weight ($\text{BMI} \leq 25 \text{ kg}/\text{m}^2$); overweight ($\text{BMI} > 25 \text{ kg}/\text{m}^2$ to $\leq 30 \text{ kg}/\text{m}^2$) and obese ($\text{BMI} > 30 \text{ kg}/\text{m}^2$)²¹.
159 Patients were also stratified by age into five groups; 1) age 54 to 64 years, 2) 65 to 69 years, 3) 70 to
160 74 years, 4) 75 to 79, and 5) 80 years and over.

161

162 ***Stratification by functional ability***

163

164 A widely used alternative measure of overall functional ability is gait speed ^{22, 23}. There is some
165 negative overall correlation between chronological age and gait speed ²⁴, although age has been
166 shown to only explain 30% of the variance in gait speed ²⁵, suggesting that gait speed itself might be
167 a unique differential indicator of function compared to age. **Furthermore in a recent study ²⁶**
168 **suggested that patients walking at a higher gait speed is representative of the high functioning**
169 **patients compared to slower patients who would represent the low functioning patients. Therefore,**
170 in the main analysis, in addition to the stratification by age, patients were also stratified into three
171 functional strata determined by their self-selected gait speed. To define the functional strata, the
172 mean and standard deviations (SD) of the gait speeds for the whole cohort were determined. All
173 patients with a gait speed falling within 1SD of the mean were defined as normally functioning (NF).

174 Patients with a gait speed greater than 1SD above the mean were defined as high functioning (HF),
175 and those with a gait speed more than 1SD below the mean were defined as low functioning (LF).

176

177 ***Data Analysis***

178

179 Comparisons were made initially between the HCFs derived from the AnyBody model and the
180 measured HCFs from the Bergmann Orthoload literature¹². This was to compare absolute values and
181 ranges between the two populations and to test the validity of the computational model outputs.
182 Stratified mean peak values and 95% confidence intervals for the resultant force and the three force
183 components are also reported.

184 ***Statistical Parametric Mapping (SPM) analysis***

185

186 The computed HCFs were analysed using Statistical Parametric Mapping²⁷ (SPM, www.spm1D.org,
187 v0.4, in the Python programming language, www.python.org). Independent linear regression
188 analyses were performed to evaluate the influence of function, age, and BMI on the magnitude of
189 the HCFs, as well as on the individual force components. For each linear regression analysis, the t
190 statistic was computed at each point in the time series, thereby forming the test statistic continuum
191 $SPM\{t\}$, technical details are provided elsewhere²⁸⁻³⁰. Significance level was set at $\alpha=0.01$, and the
192 corresponding t^* critical threshold was calculated based on the temporal smoothness of the input
193 data through Random Field Theory. Finally, the probability that similar supra-threshold regions
194 would have occurred from equally smooth random waveforms was calculated. This analysis is based
195 on the assumptions of random sampling and homology of data³⁰, as well as normality in the data
196 distribution. Adherence to the latter assumption was tested by comparing the above-mentioned
197 parametric linear regression analyses with their non-parametric counterparts³⁰. The good

198 agreement between the two types of analysis, in terms of number, temporal extent, and size of the
199 supra-threshold clusters, supports the validity of the assumption of data normality.

200 The results of the three independent, 1-dimensional linear regression analyses from SPM were
201 further verified by means of 0-dimensional multiple regression analyses. The additional analyses
202 were run in SPSS (IBM SPSS Statistics for Windows, Armonk, NY, USA) at specific time points during
203 the gait cycle, corresponding with the peak loads during stance and the local minimum during mid-
204 stance (15, 32, and 48% of the gait cycle). The force values for the 132 patients at each of these time
205 points, as well as the investigated predictor variables (BMI, age, and gait speed) were normally
206 distributed. Variance inflation factor (VIF) and Tolerance statistics revealed no multi-collinearity in
207 the data, while Durbin-Watson statistics confirmed no autocorrelation between residuals. The
208 assumptions of homoscedasticity and normal distributions of the residuals were also met.

209 **Results**

210

211 *Patient Demographics*

212

213 132 patients took part in the study and the demographics can be found in Table 1.

214 - **Insert Table 1 here** -

215 *Musculoskeletal Model Simulations*

216

217 The predicted contact forces showed comparable trends and values with measured hip contact force
218 data. The mean values were comparable with those in the Orthoload published data and the ranges
219 were generally wider as might be expected from a larger dataset¹² (Figure1 and Table 2). Hip joint
220 angles and joint moments are available as supplementary data (Supplementary File 1)

221

222 - Insert Table 2 and Figure 1 here -

223

224 **Peak Hip Contact Forces**

225

226 Stratified mean peak values for the resultant force and the three force components are reported in
227 full as supplementary data (Supplementary File Table 1).

228 **Statistical Parametric Mapping**

229

230 The results of the comparator multiple linear regression analyses were in agreement with the
231 outcome of the SPM analysis, confirming a statistically significant positive linear correlation for both
232 BMI and gait speed with HCF during both the 1st peak and 2nd peak of the stance phase, and a
233 statistically significant positive linear correlation for BMI and a negative one for gait speed during
234 the mid-stance valley. For the SPM analysis, only differences which were statistically significant for
235 more than 2% of the gait cycle are discussed.

236

237 ***BMI***

238

239 There was a statistically significant linear correlation between BMI and the magnitude of the total
240 HCF (Figure 2a). Obese patients demonstrated significantly increased HCF throughout the loaded
241 stance phase (8.8 – 53.8%), mid-swing (74.6 – 79.3%), and terminal swing (88.7 – 100%). All the
242 supra-thresholds clusters exceeded the critical threshold $t^*=3.676$ with associated p-values <0.001,
243 0.003, and <0.001 respectively.

244 The same trends were observed for the proximo-distal component (Figure 2b), for which the test
245 statistics similarly exceeded the upper threshold $t^*=+3.678$ at 5.4 – 54.3% ($p<0.001$), 73.5 – 79.2%
246 ($p=0.001$), 88.4 – 100% ($p<0.001$).

247 In the anteroposterior direction (Figure 2c), **statistically** significant negative **linear correlation** was
248 found during loading response to mid-stance (10.6 – 29.9%), terminal stance (45.4 – 55.3%), and
249 from mid-swing phase (72.2 – 100%). The clusters exceeded the threshold $t^*= -3.667$ with p-values
250 <0.001 . No significant difference was observed for the medio-lateral component (Figure 2d).

251

252 ***Age***

253

254 There was a **statistically** significant negative **linear correlation** between age and the magnitude of
255 the total HCF (Figure 3a), however this was limited to the terminal swing phase (90.7 – 98.7%), with
256 the cluster exceeding the critical threshold $t^*=-3.660$ with $p<0.001$. This indicates that younger
257 patients **are more likely to** experience higher contact forces during this phase. The same trend was
258 observed for the proximo-distal component, for which the test statistics similarly exceeded the
259 lower threshold $t^*=-3.659$ at 90.7 – 98.7% of the gait cycle, with an associated p-value <0.001 (Figure
260 3b), and for the medio-lateral component at 91.8 – 97.7% of the gait cycle ($t^*=-3.633$, $p=0.002$)
261 (Figure 3d). In the anteroposterior direction, no **statistically** significant **linear correlation** was found
262 (Figure 3c).

263

264 ***Function***

265

266 The mean gait speed for the functional ability stratum was $0.82 \text{ m}\cdot\text{s}^{-1}$ (SD; ± 0.08), $1.10 \text{ m}\cdot\text{s}^{-1}$ (± 0.09)
267 and $1.37 \text{ m}\cdot\text{s}^{-1}$ (± 0.09) for LF, NF and HF, respectively. There was a **statistically** significant **linear**
268 **correlation** between functional ability and the magnitude of the total HCF (Figure 4a). Patients with a
269 higher function demonstrated significantly increased HCF during initial contact to loading response
270 (0 – 16% gait cycle), terminal stance to initial swing (43.8 – 74.1%), and terminal swing (87.8 –

271 100%). A statistically significant negative linear correlation was instead found during mid-stance
272 (27.9-34.9%). All the supra-threshold clusters exceeded the critical threshold $t^*=\pm 3.668$, with the
273 chances of observing similar clusters in repeated random samplings being $p<0.001$.
274 The same trends were observed for the proximo-distal component (the dominant component in
275 terms of magnitude), with the corresponding supra-threshold ($t>t^*=\pm 3.666$) areas spanning from 0 –
276 15.3%, 45.1 – 73%, 87.7 – 100%, and 27.4 – 35%, respectively (Figure 4b). In the anteroposterior
277 direction, statistically significant negative linear correlation was found during initial contact to
278 loading response (0.6 – 16.3%) and terminal swing (91.6 – 100%), indicating that higher function
279 demonstrated a significantly increased posterior force during these phases (Figure 4c), while a
280 statistically significant positive linear correlation was found during mid-stance (27.3 – 45.9%). All the
281 clusters exceeded the critical threshold $t^*=\pm 3.658$ with p-values <0.001 . Statistically significant
282 positive linear correlations were observed for the medio-lateral component during initial contact to
283 leading response (0-19.8%), terminal stance to mid-swing (43.8 – 75.4%), and late swing phase (91.6
284 – 100%) (Figure 4d).

285

286 Discussion

287

288 This is the first study to explore the effect of patient characteristics on joint loading through
289 multibody modelling in a large cohort. We found that resultant HCF varies between different patient
290 groups and identified systematic differences between strata for BMI and functional ability. The BMI
291 strata displayed statistically significant differences in the resultant force throughout most of stance
292 phase. Few differences were observed between the age strata, whereas the functional strata,
293 represented by gait speed, displayed the greatest range of statistically significant differences across
294 the time series (over approx. 60% of the whole gait cycle). Patients with a high functionality had
295 increased peak loads during the stance phase of the gait cycle, while low functioning patients

296 displayed a pathological HCF, with a flattening of the typical double hump (Figure 4a). These trends
297 were similar when observing the difference in the proximo-distal component of the HCF, albeit
298 unsurprisingly considering this is the main contributor to the resultant HCF. Our average peak HCF
299 (2449N) was of a similar magnitude to the HCFs measured with instrumented implants by Bergmann
300 *et al.*¹² (2225.7N) (Table 2). No past research has considered the effect of patient characteristics on
301 HCF and comparison to previous literature is difficult. However, previous work has found that joint
302 kinematics and forces acting around the joint are affected by different patient characteristics⁶⁻⁸ and
303 altered gait variables can affect the magnitude of joint contact forces³¹, and therefore this variability
304 in HCF would be expected.

305

306 ***BMI***

307

308 We found a systematic trend for HCFs to increase with an increasing BMI, and this was expected due
309 to the increase in body mass which has been previously reported to increase linearly with joint
310 contact force³². These systematic changes in magnitude are a consistent finding in the literature
311 comparing obese and healthy weight participants when force data are non-normalised, and the
312 differences between BMI groups tend to disappear when normalised to body mass³³, which is
313 common practice in the biomechanical literature exploring function. In our study we specifically
314 chose not to normalise HCF to body weight, as we were interested in the absolute magnitude of the
315 real world forces to which the bearing surface would be exposed. Analysing non-normalised HCFs
316 may help to explain observed BMI dependant revision rates², as increased loads in preclinical
317 hardware simulator testing has been shown to increase wear volume and wear particle size³⁴.

318

319 ***Age***

320

321 When stratified by age there were very few differences observed in HCF in our patient cohort, with
322 **statistically** significant differences only found during the terminal swing phase in the proximo-distal
323 and resultant forces (90.7 – 98.7%) and medio-lateral component (91.8 – 97.7%), where the hip is
324 relatively unloaded. Differences in terminal swing phase may be related to the capacity for
325 individuals to energetically drive the limb forward. Compared to the functional strata, the temporal
326 range of significance was much less, indicating that grouping patients by age, as a measure of
327 function, does not differentiate well between patients. No other study has considered the effect of
328 age on HCF measures specifically, however in a gait study using conventional motion capture
329 analysis, Bennett *et al.* ^{7,8} observed little kinematic or kinetic differences between age groups in THR
330 patients. As noted previously, the absolute risk of revision in younger patients, can be up to ten
331 times higher than in older patients ² and it is likely that other factors such as overall activity level in
332 younger patients being higher or younger patients undertaking more demanding adverse loading
333 activities may contribute more than age-related variability in loads during normal walking.

334

335 ***Functional ability***

336

337 Our results suggest the functional capability of the patient, identified by biomechanical
338 characteristics, best identifies differences between patient groups. When stratifying patients by gait
339 speed, not only were peak forces increased in the HF group, but the waveform in the LF group
340 displayed pathological patterns with a flattening of the transition phase between the two peaks of
341 axial forces (Figure 4a). A trend was also observed in joint contact forces derived at different walking
342 speeds, with the slower walking speeds exhibiting a reduced force during the transition between the
343 peaks ³⁵. This GRF/HCF waveform has been associated with pathological symptoms in patients with
344 OA or other neurological pathologies ³⁶, suggesting that amongst our patient cohort, all of whom
345 during screening had self-reported as well-functioning, were patients who were indeed pathological,
346 identified by different HCF waveforms. Furthermore, those with higher walking speeds exhibit

347 increased GRFs and joint moments ³⁷, a trend also observed in our HCFs in the function strata.
348 Patient characteristics such as age and BMI are often controlled for in preclinical testing, whereas
349 the real-world functional capability of THR patient is frequently overlooked. Our results suggest that
350 the functional capability of patients could be the most influential factor in determining forces at the
351 bearing surface.

352

353 ***Limitations***

354

355 Previous work has identified that simulating different activities in preclinical testing also leads to
356 increased wear volume ³⁸. In the current study we only analysed walking and in reality patients
357 perform a number of other daily tasks which can change the overall loading conditions ³⁹. Walking is
358 the most commonly performed daily task ⁴⁰ however, and it is reasonable to suggest that walking
359 would have a **clinically relevant** impact on implant performance post-surgery. Within the multibody
360 modelling, a number of simulations were run from scaled generic models, and a certain level of error
361 associated with soft-tissue artefacts and the lack of subject-specific bone geometry and muscle
362 physiology information might persist. These models have been previously validated against in-vivo
363 data from different subjects however ^{14, 15, 19} with good agreement. The overall agreement with the
364 range of measurements from instrumented patients further supports the validity of the current
365 models' predictions.

366 **It could be expected that follow up time could have an effect on patient gait and hip contact force**
367 **and short-term follow up has shown as much ^{31, 41}. However, patients were recruited between 1-5**
368 **years post operatively in an attempt to avoid abnormalities due to post-surgery recovery and**
369 **patients mean follow-up time were similar in all groups (Table 1).**

370 **Finally, as this study was exploratory in nature we did not analyse any interactions between the**
371 **strata. It would be expected that there could be some interactions, for example, between age and**
372 **function ²³, which could potentially be more clinically relevant. However the analysis of interactions**

373 is not possible in *spm1D* and therefore we decided to keep the focus of the paper on the temporal
374 analysis in the individual strata, as this is relevant for other applications where full waveform data is
375 required, such as preclinical testing.

376

377

378 In conclusion, we have found that the HCF predicted at the bearing surface is highly dependent on
379 the characteristics of the patient. Conversely, current preclinical laboratory testing standards reflect
380 only one loading scenario while our study has shown systematic differences in loading patterns
381 between patient groups (Figures 2-4). To our knowledge these differences are also not considered
382 in any *in-silico* wear prediction models, although more complex waveforms, compared to ISO, have
383 resulted in greater predicted differences wear volume^{42, 43}. By extension, if future modelling included
384 patient variability, our data suggest that it is possible that differences in wear rates would also be
385 predicted. We have to accept that failure of an implant is multi factorial and patient factors and
386 surgical factors need to be taken into consideration. However if pre-clinical testing were robust
387 enough to check how implants would perform in different types of patients then patient-dependant
388 failures could potentially be better predicted. Importantly, patient variability is not considered at all
389 in current preclinical hardware simulator testing, which determines whether a device new to market
390 is fit for purpose. It was beyond the realm of this work to test this experimentally in full, but if the
391 loading profiles generated in this study were used in preclinical hardware tests, it would be expected
392 that the variability between patient groups found in this study would also be seen in experimental
393 wear testing⁴⁴. There is certainly a movement towards using different/updated testing procedures
394 with a number of authors suggesting wear testing under more adverse loads is warranted⁴⁴.
395 Improved preclinical testing, both *in silico* and *in vitro*, using more patient stratified waveforms
396 would highlight where and in whom failures are more likely to occur, allowing for better implant
397 design and more informed decision making at the time of THR planning for surgeons. Future work

398 should focus on using patient specific waveforms for *in vitro* testing to check whether the
399 differences observed in this study influence experimental wear rates.

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401

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409 Morten E. Lund, for developing the toolkit *AnyPyTools* and helping with automatizing such a large-
410 scale analysis.

411

412 **Author contributions**

413

414 All authors were involved in the conception and design of the study. DEL and EDP performed data
415 acquisition, data processing and analysis. All authors were involved in interpreting the data, revising
416 the manuscript for critically important intellectual content and approved the final version to be
417 submitted.

418

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421 The funding source had no role in the study design, collection, analysis and interpretation of the
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423

424 **Competing interest statement**

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426 The authors have no competing interests to declare

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428 **Supplementary data**

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430 **Supplementary data associated with this article can be found in the online version.**

431 Data associated with this research, in C3d format, can be found at <https://doi.org/10.5518/345>. This

432 data can be subsequently used with AnyBody Modelling software to calculate joint contact forces.

433 Musculoskeletal models for all trials in the data repository have been implemented with the

434 AnyBody Modelling software and are freely available at Zenodo (DOI: 10.5281/zenodo.1254286)

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558 **Figure Legends**

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560 **Figure 1.** Predicted HCF across the patients' cohort compared to the measured HCF from the
561 Orthoload dataset (<https://orthoload.com/test-loads/standardized-loads-acting-at-hip-implants/>)¹².
562 Resultant force (blue) and single components – proximo-distal (red), antero-posterior (orange),
563 medio-lateral (green) – are reported as mean across the cohort (solid line) and overall range of
564 variation (shaded area) and compared to the corresponding mean and range of variations from the
565 Orthoload measurements (in grey). Peak values reported in Table 2 are indicated in each plot.

566

567 **Figure 2.** Predicted hip contact forces across patients reported as a) resultant magnitude, and
568 individual components: b) proximo-distal, c) antero-posterior, and d) medio-lateral component. The
569 patients were stratified in *Healthy Weight* (blue), *Overweight* (purple) and *Obese* (red) according to
570 their BMI score. The upper panels report the averages for each patient strata (solid line) and their
571 relative 95% confidence intervals. Additionally, the loading profile from the ISO14242-1 testing
572 standard (dashed grey line) is compared to the proximo-distal forces for each group. The
573 corresponding lower panels report the results of the SPM linear regression analysis. The significance
574 α -level was set to 0.01 for each analysis and the corresponding threshold t^* are reported (horizontal
575 dashed lines). Whenever the test statistics **continuum** SPM{t} exceeds the threshold, significance is
576 reached and the p-values associated with the supra-threshold clusters (shaded grey areas) are
577 reported.

578

579 **Figure 3.** Predicted hip contact forces across patients reported as a) resultant magnitude, and
580 individual components: b) proximo-distal, c) antero-posterior, and d) medio-lateral component. The
581 patients were stratified according to their age in five groups: 54:64 (orange), 65:69 (red), 70:74
582 (grey), 75:79 (blue) and ≥ 80 (green). The upper panels report the averages for each patient strata
583 (solid line) and their relative 95% confidence intervals. Additionally, the loading profile from the
584 ISO14242-1 testing standard (dashed grey line) is compared to the proximo-distal forces for each
585 group. The corresponding lower panels report the results of the SPM linear regression analysis. The
586 significance α -level was set to 0.01 for each analysis and the corresponding threshold t^* are
587 reported (horizontal dashed lines). Whenever the test statistics **continuum** SPM{t} exceeds the
588 threshold, significance is reached and the p-values associated with the supra-threshold clusters
589 (shaded grey areas) are reported.

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591 **Figure 4.** Predicted hip contact forces across patients reported as a) resultant magnitude, and
592 individual components: b) proximo-distal, c) antero-posterior, and d) medio-lateral component. The
593 patients were stratified in Low Functioning (purple), Normal Functioning (blue) and High Functioning
594 (green) according to their self-selected gait speed. The upper panels report the averages for each
595 patient strata (solid line) and their relative 95% confidence intervals. Additionally, the loading profile
596 from the ISO14242-1 testing standard (dashed grey line) is compared to the proximo-distal forces for
597 each group. The corresponding lower panels report the results of the SPM linear regression analysis.
598 The significance α -level was set to 0.01 for each analysis and the corresponding threshold t^* are
599 reported (horizontal dashed lines). Whenever the test statistics **continuum** SPM{t} exceeds the
600 threshold, significance is reached and the p-values associated with the supra-threshold clusters
601 (shaded grey areas) are reported.
602

603 **Table 1.** Patient demographics for each classification strata. Values are reported as mean (SD) unless
 604 otherwise stated.

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		Number of patients	Female:Male	Age (Years)	BMI (kg/m ²)	Post-surgery (Years)
All		132	66:66	71.6 (7.6)	28.2(3.8)	2.8 (1.4)
BMI	Healthy Weight	29	18:11	70.1(8.2)	23.4(1.2)	2.6(1.2)
	Overweight	67	31:36	73.2(7.2)	27.6(1.3)	2.8(1.4)
	Obese	36	17:20	69.7(7.0)	33.2(2.2)	3.0(1.6)
Age	54-64	22	11:11	60.4 (2.9)	28.5(5.3)	2.9(1.5)
	65-69	37	17:20	67.0(1.4)	28.9(3.4)	2.8(1.6)
	70-74	23	14:9	72.3(1.0)	27.8(4.2)	2.1(1.1)
	75-79	28	14:14	77.4(1.2)	28.2(3.0)	2.7(1.3)
	>=80	22	10:12	82.4(3.0)	27.1(2.7)	3.0(1.5)
Function	HF	18	7:11	69.3(6.1)	27.1(2.8)	3.6(1.4)
	NF	97	48:49	71.3(7.7)	28.2(3.8)	2.7(1.4)
	LF	17	11:6	75.8(6.3)	29.3(4.4)	2.7(1.2)

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Table 2. A comparison of measured peak contact forces ¹² and the calculated peak contact forces from our study. Values are reported as mean and ranges (min-max). The reported values are highlighted in the corresponding graphs in Figure 1.

Dataset	Peak resultant force 1 st peak (R1) (min-max range)	Peak resultant force 2nd peak (R2) (min-max range)	Peak Proximal/Distal force 1st peak (PD1) (min-max range)	Peak Proximal/Distal force 2nd peak (PD2) (min-max range)	Peak posterior force(P1) (min-max range)	Peak Anterior forces (A1) (min-max range)	Peak Medial/Lateral force 1st peak (ML1) (min-max range)	Peak Medial/Lateral force 2nd peak (ML2) (min-max range)
LLJ dataset	2449.1 (1310.9 , 3913.5)	2279.0 (1093.8 , 3920.5)	2254.3 (1179.8 , 3694.4)	2197.3 (1030.8 , 3849.1)	-466.1 (-838.0 , -232.9)	-60.5 (-365.3 , 297.2)	826.0 (459.4 , 1353.5)	599.0 (273.2 , 1063.3)
Orthoload	2225.7 (1793.4 , 3147.0)	2149.9 (1721.2 , 2546.8)	2085.8 (1670.1 , 3006.5)	2073.6 (1643.8 , 2475.2)	-405.7 (-650.4 , -111.4)	23.5 (-193.0 , 211.7)	641.3 (366.7 , 819.5)	600.0 (341.1 , 807.2)