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Development of a Hip Joint Model for Finite Volume Simulations

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Abstract

This paper establishes a procedure for numerical analysis of hip joint using the Finite Volume method. Patient-specific hip joint geometry is segmented directly from computed tomography and magnetic resonance imaging datasets and the resulting bone surfaces are processed into a form suitable for volume meshing. A high resolution continuum tetrahedral mesh has been generated where a sandwich model approach is adopted; the bones are represented as a stiffer cortical shells surrounding more flexible cancellous cores. Cartilage is included as a uniform thickness extruded layer and the effect of layer thickness is investigated. To realistically position the bones, gait analysis has been performed giving the 3-D positions of the bones for the full gait cycle. Three phases of the gait cycle are examined using a Finite Volume based custom structural contact solver implemented in open-source software OpenFOAM.

Keywords: hip joint, gait analysis, bone segmentation, volume meshing,

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1. Introduction

During routine activities such as walking or stair climbing, the hip joint experiences complex loading scenarios determined by the musculotendon forces, inertia and gravity. As a consequence of the limitations of *in vivo* and *in vitro* hip joint studies, numerical models of the hip joint have increasingly been considered to better understand the mechanics of the joint. Potentially, these *in silico*¹ studies of the hip joint may provide an effective tool for analysis of hip joint mechanics and stability, helping orthopaedists make confident surgical decisions.

Numerical methods and computing power have progressed considerably since Brekelmans et al. [1] first developed a 2-D finite element model of the femur in the early seventies. In preference to model geometry derived from radiographs, recent realistic hip models are now typically captured directly from patient-specific computed tomography (CT) or magnetic resonance imaging (MRI) datasets [2–9]. The hip bones are commonly represented as *sandwich structures* consisting of two discrete types of bone [2, 3, 5]; a stiffer outer shell of cortical bone surrounding a more flexible inner core of cancellous bone. In more recent years, CT-based material property assignment has become more popular [10–13], where the stiffness of each finite element is assigned based on CT Hounsfield pixel intensities. This approach represents an improvement on the more traditional bi-material sandwich model

¹The phrase *in silico*, coined in 1989 as an analogy to the Latin phrases *in vivo* and *in vitro*, is an expression meaning “performed on computer or via computer simulation”.

22 approach, however, the exact empirical relationship between Hounsfield in-
23 tensity and stiffness can be difficult to determine and verification is not trivial
24 [11, 12].

25 Although models of the hip joint have progressed considerably, there are
26 still a number of shortcomings. Due to computational limits, the mesh is
27 often of insufficient resolution to capture the true anatomy, cortical bone is
28 often represented by degenerative shell elements, and bone surface meshes
29 can be over-smoothed to deal with the understandably complex task of
30 volumetric meshing. In spite of the steady increase in mesh densities, recent
31 models are still yet to approach the high resolution required to capture large
32 local stress gradients present in the contacting regions, giving possible ex-
33 planation of why hip contact pressure predictions of more recent studies are
34 larger than older studies [2, 4, 5, 14].

35 Numerical analysis of hip joint mechanics may be performed using one
36 of many approaches, however, due to its well established role in computa-
37 tional structural mechanics, finite element (FE) analysis is the most widely
38 employed method [1, 3, 4, 14–20]. Nonetheless, the Finite Volume (FV)
39 method, being attractively simple yet strongly conservative in nature, has
40 become a viable alternative in many solid mechanics applications [21–39].
41 Accordingly, the current work employs a cell-centred FV structural contact
42 procedure to numerically examine the hip joint, implemented in open-source
43 C++ based software OpenFOAM (Open Source Field Operation and Manipu-
44 lation, version 1.6-ext) [40–42], where the implemented physics and numerical
45 algorithms may be viewed and are open to academic scrutiny, which is not
46 possible with *black box* commercial codes.

47 2. Methods

48 2.1. Geometry Generation

49 Extraction of the hip bone geometry from tomography images is not a
50 trivial task and care must be taken to avoid the loss of significant geomet-
51 rical information. This section establishes a procedure for the generation
52 of faithful hip bone volume meshes from patient specific tomography image
53 sets.

54 CT and MRI scans were acquired of the hip joint of a 23-year-old male
55 subject with no congenital or acquired pathology. The CT images (512×512
56 pixels, $0.7422 \times 0.7422 \times 1.2500$ mm) and MRI images (256×256 pixels,
57 $1.6797 \times 1.6797 \times 2.9999$ mm) spanning from mid femur to second lumbar
58 vertebra were obtained using the GE medical systems LightSpeed VCT [45]
59 and GE medical systems Signa HDxt [45] scanners, respectively.

60 Using an automated thresholding technique, implemented in open-source
61 software 3DSlicer (version 4.0) [46], the cortical bone is extracted by select-
62 ing pixels in the range 400–1585 Hounsfield Units (HU), while the cancellous
63 bone is extracted selecting pixels in the range 200–400 HU [47]. Subse-
64 quently, the exterior bone surfaces are clearly discernible, and with minimum
65 user effort, manual separation of the femur, pelvis and sacrum is performed.
66 However, the cortical-cancellous bone interface can be much more difficult to
67 distinguish necessitating time consuming manual segmentation. During the
68 difficult task of distinguishing the cortical-cancellous bone interface, the MRI
69 image set can be combined with the CT images, using a *fast rigid registration*
70 procedure [46], limiting the subjectivity of the segmentation.

71 Once the bone pixels of interest have been selected, an enclosing triangu-

lated surface is constructed using a marching cubes procedure [46], producing a castellated surface mesh of the bone², illustrated in Figure 1(a).

To remove unwanted noise, the castellated surface meshes are smoothed to 10 iterations using a volume conserving Laplacian smoothing algorithm (Figure 1(b)) [48], as implemented in the open-source software Meshlab [49].

Figure 1: Volume Conservative Smoothing of the Bone Surfaces

77

Decimation, the process of combining small faces together, is conducted using a *quadric based edge collapse decimation* procedure, making the surface files more manageable for subsequent meshing procedures. A decimation factor of 0.2 is employed reducing the number of faces by a factor of five, with negligible effect to the bone features. This has been verified using open-source software Metro [50] by examining the geometric deviation between the original surface mesh and the decimated mesh. The mean geometric deviation is 0.0059 mm and the maximum is 0.1441 mm, which is considered acceptable.

²In this context, *castellated* refers to the resemblance of the mesh to the battlements on top of a medieval castle.

87 Finally, cleaning operations are performed on the surface mesh where
88 close vertices are merged, short edges are removed, small holes are filled,
89 and isolated faces and vertices are eliminated [51], and the final surfaces
90 (Figure 2) are exported in stereolithography (STL) format - a facet based
91 surface composed of triangles - suitable for input to most volumetric meshing
software.

Figure 2: Final Processed Bone Exterior Surfaces Embedded in a Frontal CT Slice

92

93 2.2. Volume Meshing

94 Although, hexahedral meshes have been found more accurate than tetra-
95 hedral meshes [52, 53], generation of fully hexahedral hip joint meshes is far
96 from a trivial process. Consequently, tetrahedral meshes are often employed,
97 as is the case in the current study.

98 The femur, pelvis and cortical-cancellous STL surfaces are imported into
 99 ANSYS ICEM CFD [54], and partitioned into patches of interest - distal
 100 femur, femur head, acetabulum, iliosacral joint and pubic symphysis joint
 101 - for application of boundary conditions. The cortical and cancellous bone
 102 volumes are meshed using the patch independent Delaunay tetrahedral ap-
 103 proach, and incrementally smoothed to improve the quality. Triangular
 104 prisms are *grown* from the boundary surface and cortical-cancellous inter-
 105 face, which is favourable for accurate boundary stresses, and the entire vol-
 106 ume mesh is incrementally smoothed. The volume meshes are exported via
 107 the ANSYS Fluent “.msh” format and converted to the OpenFOAM format
 108 using the OpenFOAM utility `fluent3DMeshToFoam`.

109 To overcome inferior quality cells in thin cortical bone regions, smaller
 110 local cell sizes are required, and the minimum cortical bone thickness has
 111 been limited to 1.5 mm in troublesome areas.

112 To create the articular cartilage volume meshes, the femur and pelvis ar-
 113 ticular surface meshes are extruded in the surface normal direction by 0.6 mm
 114 using the OpenFOAM utility `extrudeMesh`. The cartilage thickness, t , has
 115 been determined using the approximate acetabulum radius, R_a , and the ap-
 116 proximate femoral head radius, R_f [4]:

$$t = \frac{R_a - R_f}{2} \quad (1)$$

117 where $R_a = 27.6$ mm and $R_f = 26.4$ mm have been determined by manually
 118 fitting spheres.

119 The final high resolution hip joint volume mesh, containing a total of 569
 120 418 cells (266 817 cortical, 253 316 cancellous, 49 285 cartilage), is shown in
 121 Figure 3. The average cell breadth on the articular surfaces is approximately

0.5 mm, ensuring good resolution of contact stress gradients. However, ul-

Figure 3: Hip Joint Model Material Distribution (Cortical Bone in Red, Cancellous Bone in Green and Cartilage in Yellow, Cells Removed for Visualisation)

122

123 timately the geometric accuracy of the model is limited by the resolution of
124 the tomography images.

125 *2.3. Gait Analysis*

126 Gait analysis, the systematic study of locomotion, has been performed
127 on the subject from which the tomography images have been obtained, us-
128 ing the CODA (Codamotion V6.69H-CX1/MPX30) movement analysis sys-

tem [55, 56]. The kinematic data of 3-D positions, measured at 200 Hz, has been analysed and processed using a custom written OpenFOAM utility `visualiseGaitData` [36, 43]. The *Visualisation Toolkit* (VTK) files generated by the `visualiseGaitData` utility allow visualisation of a *stick-man* representation of the gait cycle (Figure 4) in open-source software ParaView [58], enabling accurate determination of the beginning of each gait event.

Figure 4: Gait Cycle Data Processed with `visualiseGaitData` Utility

135

A developed custom OpenFOAM utility `rotateRigidFemur` employs the kinematic gait data to calculate the time-varying position of the femur relative to the pelvis, allowing accurate positioning of the femur volume mesh relative to the pelvis for any phase of the gate cycle.

140 2.4. Finite Volume Structural Solver & Contact Procedure

A FV-based transient structural contact solver, `elasticContactNon-LinULSolidFoam`, has been specifically developed to analyse the hip joint [36, 43, 59], with inertia and body forces being neglected in the current analyses. Here, linear elastic material properties are assumed and the updated

145 Lagrangian mathematical model is applied. Special attention is given to the
146 contact algorithm, where a recently developed FV procedure based on the
147 frictionless penalty method has been used [33]. Two potentially contacting
148 surfaces are designated as the master and slave surfaces, and during the iter-
149 ative procedure, the slave vertices are checked for penetration of the master
150 surface. If a slave vertex does penetrate, an increment of interface force is
151 applied between the slave vertex and the master surface.

152 The contact algorithm is controlled via three main parameters, namely,
153 the interface stiffness, synonymous with the penalty factor or penalty stiff-
154 ness, the gap tolerance and the contact correction frequency. The penalty
155 factor controls the addition of interface force increments. Its choice affects the
156 convergence of the contact procedure; if the penalty factor is too high then
157 the contact may not converge, if too low the contact may take a prohibitively
158 long time to converge. The gap tolerance specifies the extent of penetration.
159 If the gap tolerance is too large, the bodies will penetrate by a large amount
160 and the contact pressures will be underestimated. Conversely, if too small,
161 the convergence of the contact procedure will be adversely affected. The
162 contact correction frequency factor specifies how often the contact procedure
163 is to be invoked during the inner iteration loop. It should be noted that as
164 long as the procedure converges, the penalty factor and contact correction
165 frequency do not affect the predicted mechanics, as an iterative procedure
166 is employed ensuring the contact constraints are respected within the user
167 defined gap tolerance.

168 3. Results

169 The hip joint is numerically analysed at three different phases of the gait
170 cycle, namely:

- 171 • The **mid-stance** phase;
- 172 • The hip force peak following the **heel-strike** phase;
- 173 • The hip force peak prior to the **toe-off** phase.

174 3.1. Hip Joint Model Setup

175 The three hip joint models are shown in Figure 5. All materials are
176 represented as hypoelastic, homogenous and isotropic and the material dis-
177 tribution is displayed previously in Figure 3. The cortical bone is assigned a
178 Young’s modulus of 17 GPa and a Poisson’s ratio of 0.3 [2, 4–7, 14, 60–63],
179 the cancellous bone is assigned a Young’s modulus of 800 MPa and a Pois-
180 son’s ratio of 0.2 [63, 64], and the articular cartilage is assigned a Young’s
181 modulus of 12 MPa and a Poisson’s ratio of 0.45 [19, 65].

182 As boundary conditions, shown graphically in Figure 6, the pelvis is fixed
183 at the iliosacral and pubic symphysis joints, and the distal femur is displaced
184 in the femur axial direction into the acetabulum such that resulting total hip
185 joint force is as measured *in vivo* by Bergmann et al. [44]. At *mid-stance*
186 and *toe-off*, the hip joint force is twice body weight, 1 611 N, in the femur
187 axial direction. For the *heel-strike* model, the femur axial force is 1 917 N,
188 equivalent to 2.38 times body weight.

189 The remaining femur and pelvis surfaces are specified as traction-free.
190 Custom boundary conditions with non-orthogonal corrections, previously de-
191 scribed [36, 59], are employed and the gradient terms are calculated using

Figure 5: Hip Joint Models

Figure 6: Boundary Conditions

192 the least squares method³. The models are solved in one load increment and
193 inertia and gravity forces are neglected.

194 For the contact procedure, the pelvis articular cartilage surface is des-
195 ignated as the master and the femur articular surface as the slave. The
196 contact penetration distances have been calculated using the *contact-spheres*
197 approach [35]. A contact gap tolerance of 9×10^{-6} m is employed, the penalty
198 factor is 6×10^8 and the contact correction frequency is 40.

199 3.2. Solution Process

200 The linear system formed by the discretisation of the momentum equation
201 is iteratively solved using a geometric multi-grid⁴ [66] segregated approach,
202 where each of the three components of momentum equation are separately
203 solved in terms of displacement increment. The final solution tolerance is set
204 to 10^{-7} . An additional *relative* residual, defined as the maximum difference
205 between successive displacements field solutions [36], must also reach the
206 prescribed tolerance.

207 The models have been solved in parallel on a distributed memory super-
208 computer using 32 CPU cores (Intel Xeon E5430 Quad Core 2.66 GHz) in
209 an approximate clock time of 10 hours.

210 Figure 7(a) displays the solution convergence for a typical model, showing
211 that both the solver residual and the relative residual reach the predefined
212 tolerance of 10^{-7} in approximately 160 000 outer iterations. Although a rel-

³The OpenFOAM `extendedLeastSquares` gradient scheme is employed as it as-
sumes non-orthogonal boundary cells, unlike the `leastSquares` gradient scheme.

⁴The OpenFOAM GAMG linear solver has been employed, where the coarsest solution
grid is set to the square-root of the average number of cells on each processor.

213 atively aggressive under-relaxation factor of 0.05 has been applied to achieve
 214 convergence, high frequency oscillations are still visible in the residuals most
 215 likely due to the nonlinear nature of the contact as well as lower quality
 216 cells. Figure 7(b) shows the convergence of the contact penetration to the
 217 predefined gap tolerance 9×10^{-6} m, where it can be seen that the contact
 218 converges at approximately 60 000 outer iterations and the remaining outer
 iterations are required to converge the displacements increments.

Figure 7: Typical Model Solution Convergence

219
 220 The collection of softwares employed in the current work is summarised
 221 in Appendix A.

222 3.3. Contact Stress Analysis

223 The von Mises stress distribution of the *mid-stance* model is shown in
 224 Figure 8. The most highly stressed areas, of 30 to 50 MPa, are found in the
 225 ilium directly above the acetabulum, the acetabular roof bone as well as near
 226 the fixed iliosacral joint. The scale of Figure 8 ranges from 0 to 10 MPa to
 227 allow clearer viewing of the highly stressed regions.

228 Examining the contact pressure distribution, shown in Figure 8(a), three
 229 distinct contact regions are discernible, occurring in anterior superior, pos-

Figure 8: *Mid-Stance* Model

230 terior superior and superior regions of the acetabulum. The maximum pre-
 231 dicted contact pressure is 26 MPa occurring in the most superior contact
 232 region. The predicted contact area is $3.96 \times 10^{-4} \text{ m}^2$ and has been calculated
 233 by summing the articular surface faces with a pressure greater than 1 kPa.
 234 The average contact pressure is 6.28 MPa and has been calculated by divid-
 235 ing the total contact normal force by the contact area. The total contact
 236 normal force, C_n , is calculated by:

$$C_n = \sum_f \frac{\mathbf{\Gamma}_f}{|\mathbf{\Gamma}_f|} \cdot (\mathbf{\Gamma}_f \cdot \boldsymbol{\sigma}_f) \quad (2)$$

237 where $\sum_f$ refers to the summation over all the faces of the articular surface
 238 and $\mathbf{\Gamma}_f$ is the face area vector.

239 Inspecting the model contact gap, as shown in Figure 8(b), the anterior
 240 superior and posterior superior regions show the most negative contact gap,
 241 relative to the superior region of the femoral head.

242 When the von Mises stress distribution of the *toe-off* and *heel-strike* mod-
 243 els are examined, the most highly stressed areas occur in the bone superior
 244 and posterior to the acetabulum. As with the *mid-stance* model, the acetab-
 245 ular roof is highly stressed, in particular directly above the contact regions.
 246 Additionally, the bone around the iliosacral joint experiences high stresses.

247 Inspecting the contact regions of the *toe-off* and *heel-strike* models, shown
 248 side-by-side in Figures 9, distinct contact areas are once again visible. As
 249 with the *mid-stance* model, three contact regions occur in the *heel-strike*
 250 model, where the maximum predicted contact pressure is 26 MPa, the contact
 251 area is $3.83 \times 10^{-4} \text{ m}^2$ and the average pressure is 10.1 MPa. In contrast, two
 252 contact regions occur in the *toe-off* model, where the maximum predicted

253 contact pressure is 23 MPa, the contact area is $4.62 \times 10^{-4} \text{ m}^2$ and the average
 254 pressure is 5.93 MPa.

Figure 9: Contact Pressures of the *Toe-Off* & *Heel-Strike* Models (in MPa)

254
 255 To ensure the predicted results are mesh independent, an additional *mid-*
 256 *stance* simulation has been performed using a more dense globally refined
 257 mesh. The refined mesh, containing a total of 769 529 cells (25% more cells
 258 than the original mesh) with an average articular surface cell breadth of ap-
 259 proximately 0.42 mm, predicted a maximum contact pressure of 26.67 MPa,
 260 which is within 3% of the original mesh predictions, and stress distributions
 261 close to the original mesh, demonstrating mesh independence.

262 3.4. Effect of Cartilage Thickness

263 As the current models have approximated the articular cartilage as a 0.6
 264 mm constant thickness layer, two additional simulations of *mid-stance* have
 265 been performed with increased cartilage thickness by 20% to 0.72 mm, and
 266 decreased thickness by 20% to 0.48 mm.

For the thinner cartilage model (0.48 mm), the maximum contact pressure is 34.58 MPa, the average contact pressure is 9.30 MPa, and the contact area is $2.51 \times 10^{-4} \text{ m}^2$. For the thicker cartilage model (0.72 mm), the maximum predicted contact pressure is 21.59 MPa, the average contact pressure is 6.86 MPa, and the contact area is $3.59 \times 10^{-4} \text{ m}^2$. Table 1 summarises the results for the three distinct cartilage thicknesses.

Cartilage Thickness (in mm)	Maximum Pressure (in MPa)	Average Pressure (in MPa)	Area (in m^2)
0.48	34.58	9.30	2.50×10^{-4}
0.60	26	6.28	3.96×10^{-4}
0.72	21.59	6.86	3.59×10^{-4}

Table 1: Effect of Cartilage Thickness on the Contact Mechanics of the *Mid-Stance* Model

4. Discussion

Inspecting the regions of greatest stress in the *mid-stance* model, the pelvis is relatively highly stressed in the acetabular roof, the ilium above the acetabulum and at the iliosacral joint. On closer analysis of the femur, the body of the femur holds much of the stress where significant bending occurs. These predictions agree well with results from numerical models in literature [2, 4, 14, 16, 19, 61, 63, 67], and it has been noted the large predicted stresses in the vicinity of the iliosacral joint are contributed to by the rigid fixture boundary condition [68].

282 Examining the predicted contact pressure distributions, the analyses sug-
 283 gest that the hip joint contact is not perfectly congruent and that there are
 284 distinct regions of local high contact pressure, agreeing with previous FE pre-
 285 dictions [4, 14]. Comparing the contact mechanics of the *mid-stance* model
 286 with values from literature, the predicted contact pressures are larger than
 287 the reported values. Peak contact pressures in literature vary from 1 to 18
 288 MPa [2, 4, 14, 16, 19, 61, 63, 67], whereas the maximum predicted contact
 289 pressure in the *mid-stance* model is 26 MPa. A possible explanation for this
 290 difference may be attributed to the simplifying assumptions commonly made
 291 in literature with regard to spherical congruent contact surfaces that overes-
 292 timate contact areas, hence underestimating contact pressures. Additionally,
 293 many of the published numerical models represent the contact surfaces with
 294 relatively low resolution grids essentially averaging local contact stress peaks.
 295 A contributing factor that may result in the current model overestimating the
 296 contact pressure may be ascribed to simplifying assumptions with regard to
 297 the uniform bone mechanical properties and the constant thickness articular
 298 cartilage. Employing a stiffness of 17 GPa for the acetabular cortical bone
 299 may overestimate the stiffness of the pelvis, reducing the natural congruency
 300 of the joint and increasing contact stresses. A more realistic representation of
 301 varying cartilage thickness in addition to spatially varying tomography-based
 302 bone properties would certainly result in a more faithful model.

303 Analysing the effect of cartilage thickness has shown a considerable in-
 304 fluence on the predicted mechanics. As expected, it has been found that
 305 there is an inverse relationship between the cartilage thickness and maxi-
 306 mum contact pressure, where the maximum contact pressure increases by

307 approximately 60% as the cartilage thickness decreases by 33% from 0.72
 308 to 0.48 mm. However, surprisingly at first, the middle thickness (0.6 mm)
 309 model predicts the lowest average contact pressure. The reason for this may
 310 be deciphered by noticing that the contact pressure increases in the thicker
 311 cartilage model in the anterior and posterior regions of the acetabulum and
 312 lowers in the superior region. In reality, it is expected that the physiological
 313 cartilage thickness is greater on the superior femur head surface, therefore
 314 increasing the overall cartilage thickness may overestimate the thickness in
 315 the anterior and posterior regions of the acetabulum and femur head.

316 Of the three phases of gait examined, as anticipated the largest stresses
 317 and contact pressures occur in the *heel-strike* model, corresponding to the
 318 peak in gait cycle total joint forces. For all models, the predicted stress
 319 values and locations are consistent with previous FE studies [2, 4, 14, 16,
 320 19, 36, 61, 63, 67], however, the magnitude of maximum contact pressures
 321 are greater than previously predicted. The effect of cartilage thickness has
 322 been examined and as expected the maximum predicted contact pressure
 323 has been found to be inversely related to the cartilage thickness, with the
 324 case of the thinnest cartilage producing the greatest maximum contact pres-
 325 sure. However, the largest average contact pressure has been found in the
 326 thickest cartilage case, most likely due to the assumption of constant thick-
 327 ness cartilage resulting in underestimation of superior cartilage thickness and
 328 overestimation of anterior/posterior cartilage thickness.

329 All presented results show that this novel numerical model, based on Fi-
 330 nite Volume Method as implemented in an open-source software, represent
 331 a powerful alternative to well-established commercial FE-based softwares.

332 As its object-oriented nature permits complex custom routines to be imple-
333 mented in an efficient manner, further improvement of the current model will
334 be focused on implementation of material properties based on CT Hounsfield
335 pixel intensities and on employment of physiologically realistic musculoten-
336 don loading models [36, 43].

337 5. Conflict of Interest Statement

338 There is no conflict of interest.

339 6. Acknowledgements

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342 Clinic Dublin for conducting the gait analysis.

343 7. Appendix A

344 The softwares employed in the current work are summarised in Table 2.

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Phase of Model Development	Software/Utility	Description
CT/MRI segmentation	3DSlicer [46]	Open-source image segmentation and surface generation software
Surface processing	Meshlab [49]	Open-source software for general surface processing
Surface processing	Metro [50]	Open-source software allowing verification of decimated surfaces
Volumetric meshing	ANSYS ICEM CFD [54]	Commercial meshing software for generation of volumetric meshes
Volumetric meshing	OpenFOAM/fluvent3D-MeshToFoam [41, 42]	OpenFOAM utility to convert mesh to OpenFOAM format
Volumetric meshing	OpenFOAM/extrudeMesh [41, 42]	OpenFOAM utility to create cartilage volumetric mesh
Gait analysis	OpenFOAM/visualise-GaitData [43]	Custom OpenFOAM utility to convert gait data into a form suitable for viewing
Gait analysis	OpenFOAM/rotate-RigidFemur [36]	Custom OpenFOAM utility for relative positioning of the femur using gait data
Solving	OpenFOAM/elastic-ContactNonLinUL-SolidFoam [59]	Custom OpenFOAM application for numerical calculation of displacements, stresses and strains by Finite Volume Method
Post-processing	ParaView [58]	Open-source visualisation software for post processing results

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