



Deposited via The University of Leeds.

White Rose Research Online URL for this paper:

<https://eprints.whiterose.ac.uk/id/eprint/211322/>

Version: Accepted Version

Article:

Raju, V. and Koorata, P.K. (2022) Influence of material heterogeneity on the mechanical response of articulated cartilages in a knee joint. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 236 (9). pp. 1340-1348. ISSN: 0954-4119

<https://doi.org/10.1177/09544119221116263>

This is an author produced version of an article published in Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine. Uploaded in accordance with the publisher's self-archiving policy.

Reuse

Items deposited in White Rose Research Online are protected by copyright, with all rights reserved unless indicated otherwise. They may be downloaded and/or printed for private study, or other acts as permitted by national copyright laws. The publisher or other rights holders may allow further reproduction and re-use of the full text version. This is indicated by the licence information on the White Rose Research Online record for the item.

Takedown

If you consider content in White Rose Research Online to be in breach of UK law, please notify us by emailing eprints@whiterose.ac.uk including the URL of the record and the reason for the withdrawal request.

1 **Influence of material heterogeneity on the mechanical response of**
2 **articulated cartilages in a knee joint**

3 Vaishakh Raju^a and Poornesh K Koorata^{a,*}

4 *^aApplied Solid Mechanics Laboratory, Department of Mechanical Engineering,*
5 *National Institute of Technology Karnataka, Surathkal 575025, India;*

6

7 *Corresponding Author

8 Poornesh K Koorata

9 Applied Solid Mechanics Laboratory, Mechanical Engineering Department,

10 National Institute of Technology Karnataka, Surathkal 575025, India.

11 Email: kpkoorata@nitk.edu.in

12 Tel.: +91-824-247-3650

13 Fax: +91-824-247-4003

14

15

16

17

18

19

20

21

22

23

24

25

26

27

28

29 **Abstract**

30 Structurally, the articular cartilages are heterogeneous owing to nonuniform distribution
31 and orientation of its constituents. The oversimplification of this soft tissue as a
32 homogeneous material is generally considered in the simulation domain to estimate
33 contact pressure along with other physical responses. Hence, there is a need for
34 investigating knee cartilages for their actual response to external stimuli. In this article,
35 impact of material and geometrical heterogeneity of the cartilage is resolved using well
36 known material models. The findings are compared with conventional homogeneous
37 models. The results indicate vital differences in contact pressure distribution and tissue
38 deformation. Further, this study paves way for standardising material models to extract
39 maximum information possible for investigating knee mechanics with variable
40 geometry and case specific parameters.

41

42 **Keywords:** Articular cartilage; Contact Pressure; Finite element analysis; Knee joint;
43 Material heterogeneity; Superficial layer;

44 **1. Introduction**

45 The knee is a complicated joint that includes tibiofemoral and patella-femoral
46 articulations^{1,2}. Mechanically, articular cartilage functioning similar to frictionless
47 bearing between the surfaces³. The knowledge of contact pressure and stress fields is
48 essential in predicting the onset of functional damage of these tissues⁴⁻⁶. The
49 computational model for whole-knee biomechanics is a useful clinical tool for
50 determining the onset and progression of disease and injury. The critical challenge in
51 simulating soft tissues like articular cartilage is the complexity of structure,
52 compounded by the heterogeneous distribution of collagen fibers throughout its cross-
53 section^{7,8}. A dense extracellular matrix (ECM) with a random distribution of

54 chondrocyte cells constitutes the cartilage structure. ECM is primarily composed of
55 collagen fibers, proteoglycan, and water fraction⁹. These components jointly help
56 maintain ECM water, critical in keeping properties like sudden impact strength and high
57 compressive strength¹⁰. A cross-section of articular cartilage reveals the direction of
58 collagen fibers into three zones: fibers oriented parallel to articulating surface in the
59 superficial zone (SZ), randomly oriented in transitional zone (TZ) and oriented
60 perpendicular to subchondral bone in the deep zone (DZ)^{11,12}. The zonal thickness of
61 SZ, TZ, and DZ is about 12%, 32%, and 56% of cartilage's total thickness, respectively
62 ¹³.

63 The most common material models for simulating the mechanical behavior of
64 cartilage are isotropic elastic (IE), isotropic poroelastic (IPE), transversely isotropic
65 poroelastic/ transversely isotropic elastic (TIPE/TIE), and fibril-reinforced
66 poroviscoelastic/poroelastic (FRPVE/FRPE)^{10,14,23,15-22}. The isotropic elastic model
67 predicts the instantaneous cartilage response faster compared with its alternative models
68 ²⁴⁻²⁷. Essentially these models give a qualitative understanding of the response.
69 Nevertheless, in reality, articular cartilages comprise a porous matrix saturated with
70 water (68% - 88% of cartilage weight). The response of cartilage tissue is influenced by
71 fluid pressure, according to studies ^{28,29}. Hence the biphasic characteristics of cartilage
72 are generally studied with IPE models. In IPE models, the fluid flow in cartilage is
73 modelled with Darcy's law related to permeability(k)³⁰.

74 Apart from biphasic characteristics, the articular cartilage constitutes a non-fibril matrix
75 and collagen fibril network. FRPE model may suit well to simulate the response of such
76 structure³¹. The material orientation is assigned such that it mimics contribution of
77 collagen fibril orientation (arcade-like structure). However, this model is
78 computationally not cost effective, lacks their relevance in clinical applications. TIE

79 models are also widely used in modelling articular cartilage³²⁻³⁴. From recent studies,
80 the TIE model can predict intact articular cartilage uniaxial compression responses with
81 higher accuracy³⁵. The highly heterogeneous nature of superficial zone can simulate
82 well with such models^{36,37} the most influential compartment on articular cartilage to
83 mechanical response³⁸.

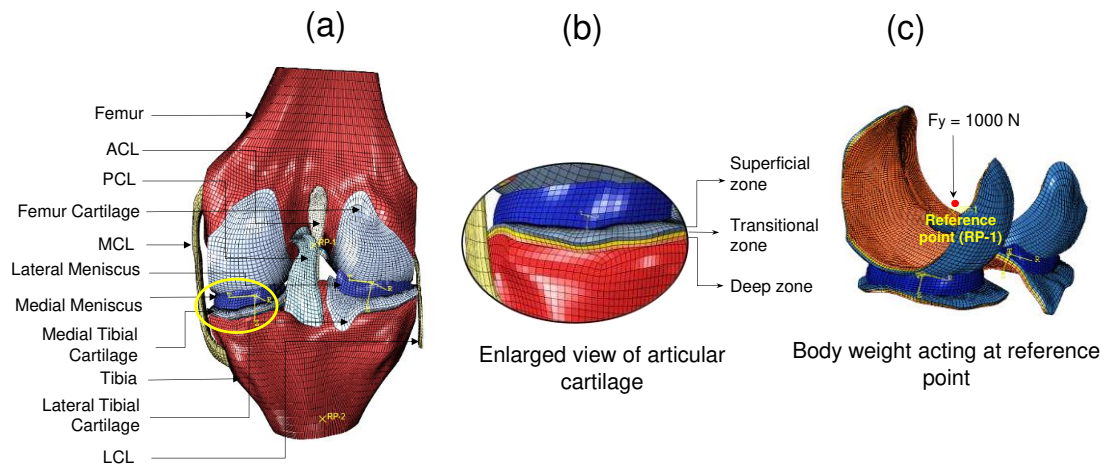
84 Although many studies were conducted on the articular cartilage behavior with
85 multiple constitutive models³⁹⁻⁴⁴, the mechanical response of heterogeneous cartilage at
86 maximum loading position in a gait is not adequately compared. Therefore, our primary
87 objective is to address this issue by considering graded cartilage material with multiple
88 heterogeneous constitutive models. The scope of the paper extends further to identify
89 the critical differences among material models in predicting contact pressure and stress
90 distributions. Even though the cartilage structure is complex and comprises dense
91 collagen fibre networks and extracellular matrixes, the hypothesis addressed for the
92 current study assumes cartilage as elastic models. The present study focused on the
93 tibio-femoral articulation, ignoring the effects of muscle, tendon, and patella forces on
94 cartilage response prediction at full extension position in a gait cycle.

95 **2. Materials and Methods**

96 *2.1 Geometry and Finite Element model*

97 An existing knee joint geometry (open knee) developed at Computational Bio modeling
98 Core and Department of Biomedical Engineering, Cleveland Clinic from a female
99 corpse (70 years & 77 kg) is used for the current study^{45,46}. An expanded view of knee
100 geometry (Figure 1(a)) and the 3D knee substructure is imported into Abaqus CAE for
101 analysis. The model contains four ligaments anterior and posterior cruciate ligaments
102 (ACL and PCL) and medial and lateral collateral ligaments (MCL and LCL), two

103 cartilages (femur and tibia cartilage), two meniscus (lateral and medial meniscus) and
104 two bones (femur and tibia).



105

106 **Fig 1.** (a) [The posterior view of 3D finite element knee joint model](#) (b) enlarged view of tibial
107 cartilage with three layers in different colours representing inhomogeneity; (c) Bodyweight
108 acting on knee joint simulated by applying 1000N force in vertical direction acting downwards
109 at the reference point which is constrained with femur cartilage and meniscus.

110 The coordinate system of the geometry is synchronized with the Abaqus, such
111 that the x-axis is anterior-posterior, where the anterior or posterior force component can
112 apply. Similarly, the y-axis is the proximal-distal where the vertical ground reaction
113 force can use, and the z-axis is the medial-lateral direction where medial or lateral joint
114 force components can apply. The ground reaction force becomes the body weight (BW)
115 when the body is in a full extension position during the stance phase of a gait cycle. The
116 valgus or varus rotation is about the x-axis, internal or external rotation is about the y-
117 axis, and the flexion or extension rotation is about the z-axis.

118 In this article, the heterogeneity of the cartilage is defined in terms of material
119 and geometrical heterogeneities. The material heterogeneity is the inhomogeneous
120 distribution of material constituents (such as fiber density and orientation) and it is
121 modeled using the corresponding constitutive models. The geometrical heterogeneity is

122 the inhomogeneity in terms of cartilage structure (such as superficial, transverse/middle,
123 and deep zones) as shown in Figure 1(b). It may be noted here is that native architecture
124 of cartilage may have these individual layers with different thicknesses depending on
125 the collagen fiber structural inhomogeneity. The current study, however, assumes mesh
126 size of each zone is to be same for numerical convergence purpose. These soft tissues
127 are discretized into 56433 hexahedral elements with an element size of 0.5 mm each.
128 The cartilage, meniscus, and ligaments are meshed with hexahedral brick elements
129 (element type: C3D8) and the femur and tibia are meshed with shell elements (element
130 type: S4) as per Abaqus/Standard user's manual ⁴⁷. The use of an 8-node element in
131 contact modelling has the potential to improve contact response and numerical
132 convergence than higher node elements.

133 2.2 *Interface, constraints, loading and boundary conditions*

134 The interaction between the cartilages at the articulating surface is assigned as
135 frictionless contact. A rigid body tie constraint ties the ligaments and cartilage with
136 bone's nodes at the bone insertion points to retain their position. Another rigid body
137 constraint connects the tibia and femur to corresponding reference points (RP-1 and RP-
138 2), such that the tibia and femur act as rigid body. The RP-1 is at the centre position of
139 lateral and medial femoral epicondyles for the femur. The femur can rotate about RP-1,
140 and the meniscus is constrained such that its position is maintained between two
141 cartilages.

142 All rotational and linear motions (6DOF) of tibia is constrained and the femur is
143 set free to move in all five degrees of freedom but restricted in knee flexion. Since this
144 study focus on the maximum extension position of gait cycle, when flexion angle is
145 zero. Thus RP-1 is subjected to a load of 1000N (compressive) as shown in Figure 1(c)

146 2.3 *Material Models*

147 The influence of homogeneous and heterogeneous (both material and geometrical case)
 148 cartilage surface texture is compared for mechanical response using well-known
 149 material models. [These are IE, IPE, and TIE models. Note that the basic models such as](#)
 150 [IE and IPE assumes collagen fibers is homogenized with rest of the cartilage](#)
 151 [constituents. The TIE model is extended version of these basic models where collagen](#)
 152 [contribution is assumed as reinforcements.](#) The material constants of these models (the
 153 constitutive relation of all these models are given supplementary information S1) are
 154 provided in Table 1.

155 **Table 1**

156 The material parameters used for modeling homogeneous and heterogeneous articular
 157 cartilage.

Material models	Homogeneous (<i>non-gradient</i>)	Heterogeneous (<i>gradient</i>)	Source
IE	$E = 15 \text{ MPa}$ $\nu = 0.475$	$E_s = 15 \text{ MPa}$ $E_t = 10 \text{ MPa}$ $E_d = 5 \text{ MPa}$ $\nu = 0.475$	27*
	$E = 15 \text{ MPa}$ $\nu = 0.475$ $S_l = 1$ $e = 4$ $k = 0.001 \text{ mm}^4/\text{Ns}$	$E_s = 15 \text{ MPa}$ $E_t = 10 \text{ MPa}$ $E_d = 5 \text{ MPa}$ $\nu = 0.475$ $S_l = 1, e = 4$ $k = 0.001 \text{ mm}^4/\text{Ns}$	
TIE	$E_p = 5.8 \text{ MPa}$ $E_t = 0.46 \text{ MPa}$ $\nu_p = 0.87 (-)^{**}$	$E_{ps} = 5.8 \text{ MPa}$ $E_{ts} = 0.46 \text{ MPa}$ $\nu_{ps} = 0.87 (-)$ $\nu_{ts} = 0.03(-)$	29,49,50*

$$\nu_t = 0.03(-)$$

$$G_{ts} = 2.5 \text{ MPa}$$

$$G_t = 2.5 \text{ MPa}$$

$$E_{pt} = 4 \text{ MPa}$$

$$E_{tt} = 0.46 \text{ MPa}$$

$$\nu_{pt} = 0.87 (-)$$

$$\nu_{tt} = 0.05(-)$$

$$G_{tt} = 2 \text{ MPa}$$

$$E_{pd} = 2 \text{ MPa}$$

$$E_{td} = 0.46 \text{ MPa}$$

$$\nu_{pd} = 0.87 (-)$$

$$\nu_{td} = 0.2 (-)$$

$$G_{td} = 1 \text{ MPa}$$

158 Notes: E = Elastic modulus, ν = Poisson's ratio, E_s, E_t, E_d = Elastic moduli of the superficial,
159 transitional and deep layer, E_{ps}, E_{pt}, E_{pd} and E_{ts}, E_{tt}, E_{td} are in-plane and out of plane
160 Young's moduli for the three layers, similarly $\nu_{ps}, \nu_{pt}, \nu_{pd}$ and $\nu_{ts}, \nu_{tt}, \nu_{td}$ are in-plane and out
161 of plane Poisson's ratio for the three layers. G_{ts}, G_{tt}, G_{td} are out of plane shear modulus for all
162 layers respectively, S_l = specific weight of wetting liquid, k = permeability and e = void ratio.

163 *Source for the homogeneous (non-gradient) model.

164 **Poisson's ratio in the in-plane direction has been chosen from article ²⁹, but the value is
165 altered to match the material's consistency in Abaqus.

166 The rest of the joint parts are modelled as per Table 2. Even though the bone
167 (femur and tibia) consists of the cortical and cancellous parts, we approximated it as a
168 uniform rigid body. Similar to articular cartilage, the meniscus also has complicated
169 architecture, including a network of collagen fibers. To reduce the complexity in
170 modelling, we modelled the meniscus (lateral and medial) with TIE material ⁵¹. The

171 ligaments are modeled with Neo-Hookean isotropic hyperelastic material (nearly
 172 incompressible) model.

173 **Table 2**

174 The material parameters for the components of the knee joint other than cartilage

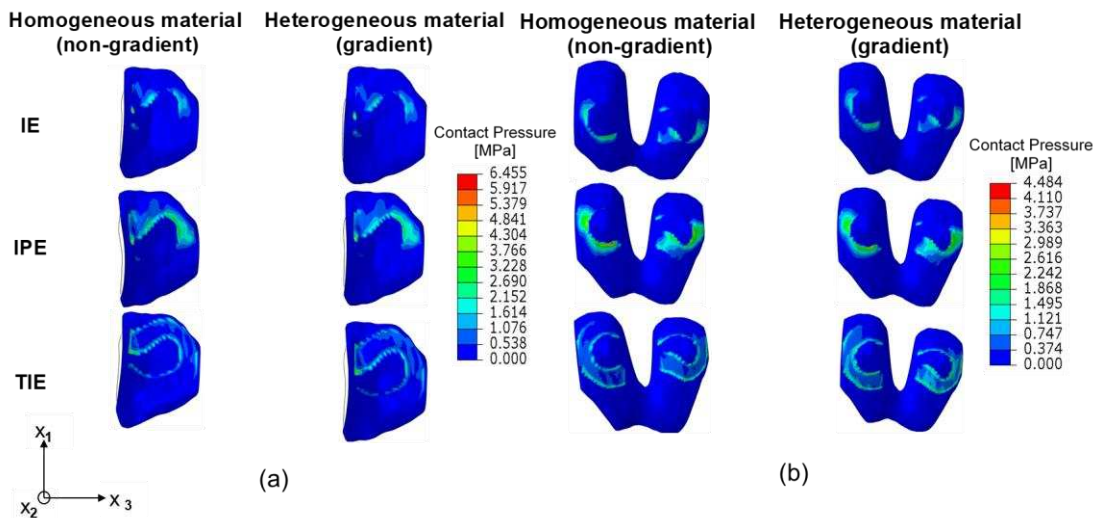
Parts	Components	C_1 (MPa)	D_1 (MPa) ⁻¹	Source
Ligaments isotropic hyperelastic (Neo-Hookean)	ACL	1.95	0.00683	52,53
	PCL	3.25	0.0041	
	MCL	1.44	0.00126	
	LCL	1.44	0.00126	
Meniscus (TIE)	$E_p = 120 \text{ MPa}, E_t = 20 \text{ MPa}$ $\nu_p = 0.2 (-), \nu_t = 0.3 (-)$ $G_p = 8.33 \text{ MPa}, G_t = 57.7 \text{ MPa}$			53

175 Notes: C_1 and D_1 are Neo-Hookean material constants, E_p and E_t are in-plane and transverse plane
 176 elastic modulus, ν_p and ν_t are in-plane and transverse plane Poisson's ratio, G_p and G_t are in-
 177 pane and transverse plane shear modulus.

178 **3. Results**

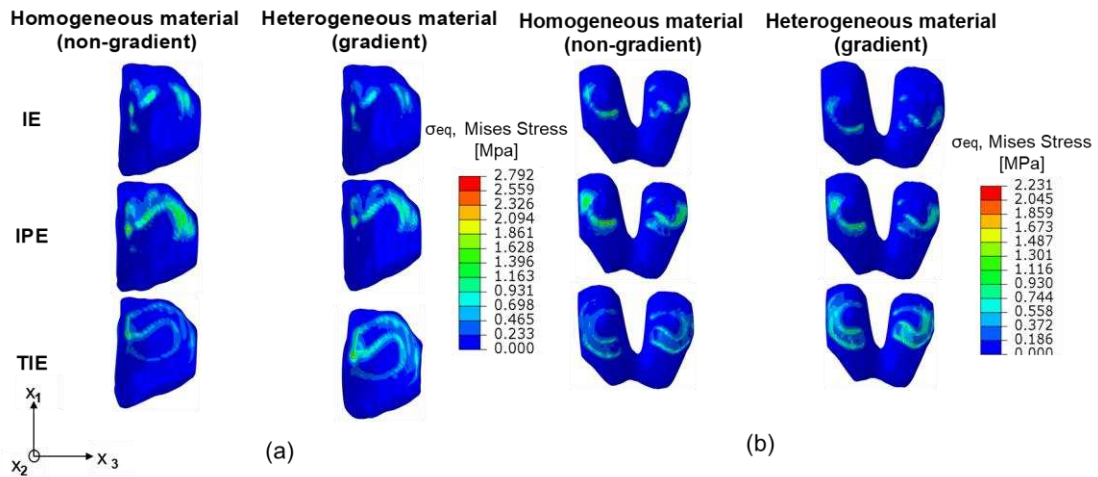
179 [The contact pressure distribution in the tibial cartilage surface from the meniscus](#)
 180 [impact are compared for homogeneous and heterogeneous cartilage cases are given in](#)
 181 [Figure 2](#). Though no significant difference is observable in Figure 2(a), the
 182 homogeneous model provides higher contact pressure distribution compared to
 183 heterogeneous case. Among all the constitutive models compared, TIE and IPE models
 184 shows a clear impression of contact pressure on the cartilage surface.

185 Also the equivalent stress has an essential role in predicting knee pathologies.
 186 Figure 3(a) depicts the stresses distribution on the tibial surface. Compared with the IE
 187 and TIE model, the IPE model showed maximum equivalent stress generated in the
 188 femur cartilage. This indicates the biphasic tissue model supports more load than the
 189 simple model during the load transfer mechanism. Also it is observed here that the IE
 190 and IPE geometrical heterogeneous models provide less uniformity in stress distribution
 191 with respect to the TIE model. Figure 3(b) shows the femoral stress distribution and it
 192 follows similar pattern for all models, where the TIE model gives a clear understanding
 193 of the stress impression with no stress concentration. [The maximum stress generated on](#)
 194 [the TIE model are 2.559 MPa and 2.792 MPa for the tibial cartilage, 2.045 MPa and](#)
 195 [2.231 MPa for the femoral cartilage for the material heterogeneous and homogeneous](#)
 196 [cases, respectively.](#)



197

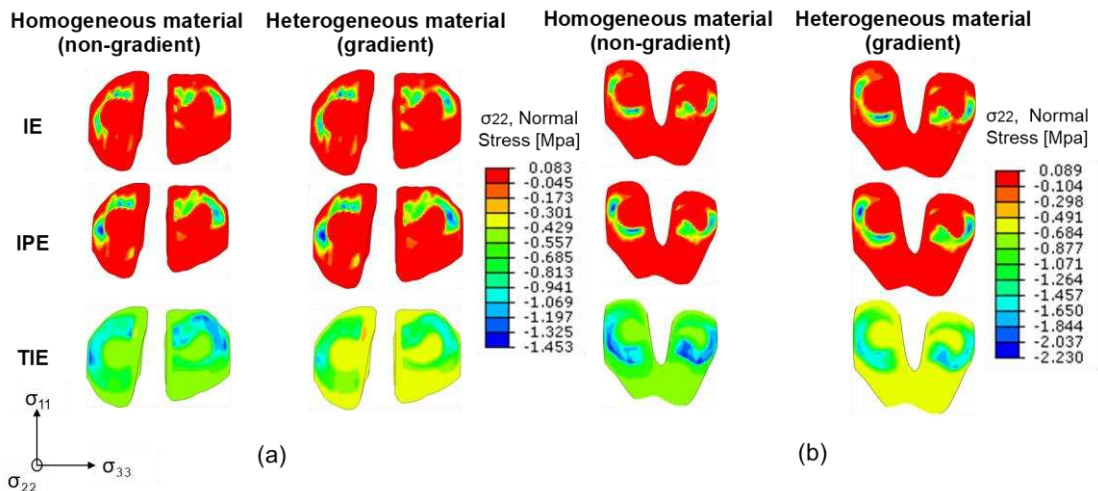
198 **Fig 2.** [Distribution of contact pressure on \(a\) the tibial lateral surface; \(b\) the femoral surface](#)



199

200 **Fig 3.** The contours for equivalent stress generation on (a) the tibial lateral surface; (b) the femoral
 201 surface

202 Directional stress may give better insights into the knee mechanics. Figure 4(a) and 4(b)
 203 clearly distinguish the directional stress impression in tibial and femoral surface. Figure
 204 4(a) shows a uniform stress distribution in heterogeneous (both material and
 205 geometrical) case for all models compared to homogeneous model. In TIE model the
 206 tibial cartilage has a lower stress than the femur cartilage -2.2 MPa and -1.4 MPa
 207 respectively. Also the compression stress in the homogeneous situation is relatively
 208 high for the TIE model. This could be owing to the high rigidity provided by the
 209 cartilage's surface.

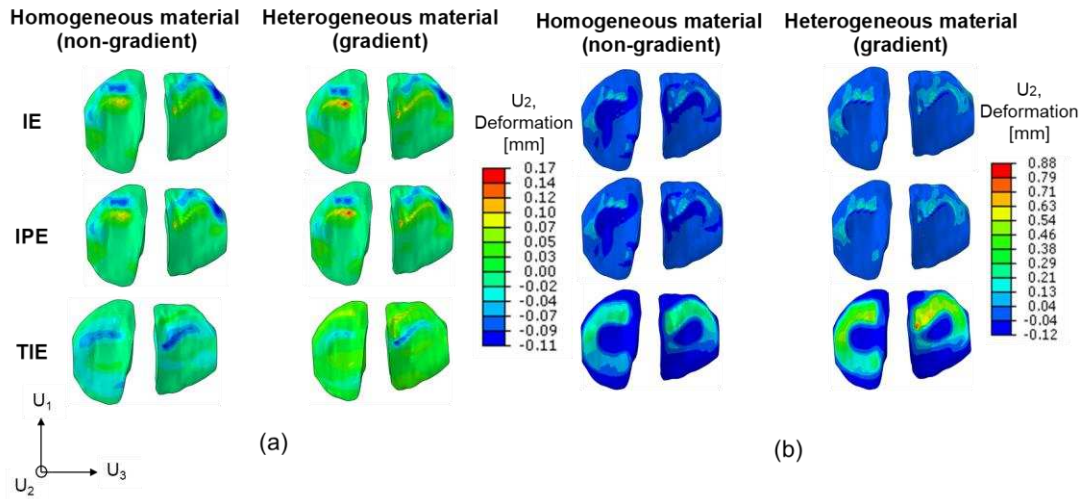


210

211

212

Fig 4. The normal stress distribution on (a) the tibial surface; (b) the femoral surface



213
 214 **Fig 5.** The cartilage deformation (a) U1 (anterior-posterior) on the tibial surface; (b) U2 (proximal-distal)
 215 on the tibial surface at full extension position after loading.

216 **Table 3**

217 [The estimated over-prediction of mechanical measures in percentage for homogeneous](#)
 218 [model relative to its heterogeneous alternative.](#)

Comparison of models	Zones	Mises stress (%)	Max. prin. Stress (%)	Max. prin. log. strain (%)	Min. prin. log. strain (%)	Max. def. (%)	Max. pres. generated (%)
Homo. IE	SZ	25	32	9	28	6	0.3
Vs	TZ	12	5	10	13	7	6
Hetro. IE	DZ	4	15	22	28	6	12
Homo. IPE	SZ	25	36	49	42	5	0.7
Vs	TZ	14	6	8	99	4	6
Hetro. IPE	DZ	22	15	23	4	5	12
Homo. TIE	SZ	60	42	4	95	3	47
Vs	TZ	4	88	71	98	3	37
Hetro. TIE	DZ	37	215	42	27	2	47

219 Note: Homogeneous isotropic elastic (Homo. IE), heterogeneous isotropic elastic (Hetro. IE),
 220 homogeneous isotropic poroelastic (Homo. IPE), heterogeneous isotropic poroelastic (Hetro. IPE),
 221 homogeneous transversely isotropic elastic (Homo. TIE), heterogeneous transversely isotropic elastic

222 (Hetro. IE), maximum principal (Max. prin.), maximum principal logarithmic strain (Max. prin. log.
223 strain), maximum deformation (Max. def.), the maximum pressure (Max. pres.)

224 The tibial cartilage deformation U_2 (anterior-posterior) over the medial and lateral
225 compartment of the tibial cartilage surface are shown for different material models and
226 also for the deformation in U_1 (proximal-distal) given by Figure 5(a) and 5(b). The
227 maximum tibial cartilage deformation for the IPE geometrical heterogeneous model is
228 0.17 mm in the posterior and 0.88 mm in the proximal direction, where the solid model
229 has lesser deformation. It indicates that porosity impacts tibial deformation and pressure
230 distribution in a knee joint during the standing position. It can be observed from Figure
231 5 that even with the homogeneous TIE model, the in-plane (anterior-posterior) and
232 through plane (proximal-distal) cartilage deformation is more pronounced and clear
233 than the material heterogeneous IPE model. Therefore TIE models can safely assumed
234 to be reliable in predicting the onset and progression of OA.

235 In addition, Table.3 compares the maximum variation in Mises stress, principal
236 stress, strain, deformation, and pressure generated on the cartilage with the
237 homogeneous and heterogeneous entity. The maximum percentage change in stresses
238 and strains is observed higher in the TIE model than in the IE and IPE models. The
239 maximum primary stress varies about 200 percentage in TIE models between
240 homogeneous and heterogeneous entity.

241 **4. Discussion**

242 This study examines the influence of heterogeneous material characteristics as a
243 function of tissue depth from superficial to deep zone with multiple constitutive
244 material models. Also, investigate the impact of stresses and strains during the full
245 extension position (standing position) using the finite element knee model. Verifying

246 simulation findings is a critical step, and we double-checked that our findings for the
247 intact knee model match those found in the literature⁵⁴⁻⁵⁹. The IE, IPE, and TIE
248 cartilage material models are compared with homogeneous and heterogeneous entities.
249 The contact pressure distribution is observed not so evident from IE models compared
250 to IPE or TIE models, and the material heterogeneity produces a relatively lower
251 magnitude of pressure distribution.

252 According to some studies, the material property of articular cartilage varies
253 enormously with distance from the articular surface, especially in the superficial region;
254 hence heterogeneous constitutive models suit well for such studies⁶⁰. Also, many
255 constitutive models are proposed for implementing intact and OA heterogeneous
256 characteristics. The heterogeneous behavior of a finite element cartilage model with an
257 incompressible, poroelastic solid matrix reinforced by an inhomogeneous, distributed
258 fibre filled with an incompressible fluid in the collagen–proteoglycan solid matrix's is
259 well predicted⁷. The split-line patterns are utilized for FRPE inhomogeneous cartilage
260 models to illustrate diverse cartilage influenced by collagen fibres. [The average Mises](#)
261 [stresses in the homogeneous IE model are 2.7 MPa in the tibial lateral compartment and](#)
262 [2.2 MPa in the femoral lateral component, similar to the range reported in the previous](#)
263 [work^{61,62}. Also, with a load of 1000 N, the maximum contact pressure generated on](#)
264 [cartilage surface varies between 6 and 16 MPa⁶³](#). Particularly in the geometrically
265 heterogeneous model, the maximum principal stresses increased significantly in the
266 cartilage's middle zone²¹. The early OA model showed increased compressive strains in
267 the articulating layer, as well as decreased stresses and fibril strains, especially in the
268 intermediate zone^{33,64}.

269 We assumed that IE models predict more accurate findings under short-term
270 loading, similar to the assumptions made in other investigations^{10,65}. Moreover, in

271 compression, the IE material model (elastic) showed the highest primary stresses,
272 whereas the other models indicated tension. It's because the IE material model doesn't
273 include fluid pressure. According to previous research, the load supported by fluid in
274 cartilage can be as high as 5–15 MPa, which can support 80–90% of the BW when
275 walking^{15,28,42} However, different parameters undoubtedly likewise influence the
276 material heterogeneity characteristics in cartilage. This might alter minimally if more
277 attributes were included (heterogeneity in fluid flow across cartilage thickness), but it
278 should not change any conclusions about the correlation between the material models.
279 [The IE and IPE models are basic models, which do not take into account of material](#)
280 [heterogeneity of the cartilage. From the present study, material heterogeneous TIE](#)
281 [models show better impression of meniscus reaction on articular cartilage compared to](#)
282 [its homogeneous alternative. While IE and IPE geometrically heterogeneous models](#)
283 [predict poor impression on cartilage surface. Hence material heterogeneous TIE model](#)
284 [can be used as a better alternative model to fiber reinforced model in knee biomechanics](#)
285 [study.](#)

286 In a wide range of biomedical engineering applications, it is becoming
287 increasingly important to develop better constitutive models for modelling soft tissue
288 deformation. [Researchers can use these articular cartilage model comparisons to look](#)
289 [into tissue-joint mechanism, implant material design, and have a better understanding of](#)
290 [microscale response of tissues.](#)

291 **5. Conclusions**

292 In summary, the following conclusions drawn: (1) the maximum cartilage
293 contact pressure induced by the knee joint with the geometrically heterogeneous
294 material model is lower compared to the homogeneous model; (2) the maximum Von-
295 Mises stress may not present a quantitative assessment of cartilage damage; (3)

296 Poroelastic cartilage model can be helpful to in estimative anterior/posterior
297 deformation whereas material heterogeneous TIE model is suitable in understanding
298 proximal/distal deformation limits. (4) The maximum change in stresses and strains are
299 observed in TIE models than IE and IPE models.

300 The study has some limitations in terms of model generation, input, and
301 assumptions. The knee kinematics is very complex, and hard to simulate the exact
302 motion; the tibia stresses are a combination of loading (compression), shear, and tensile.
303 In this article, only the knee joint's standing (full extension) position is considered as a
304 simple case of loading. Other soft tissues are left out (patella, patellar tendon, joint
305 capsule, and skin) in the models because the focus of this investigation is to compare
306 three distinct material models of cartilage having the geometry under same applied load.

307 **Acknowledgement**

308 KKP gratefully acknowledges the financial support of the DST-INSPIRE Faculty
309 Award (File No.: DST/INSPIRE/04/2016/000735), Govt. of India, in executing the
310 computational part of this work.

311 **Declaration of conflicting interests:**

312 The author(s) declared no potential conflicts of interest with respect to the research,
313 authorship and/or publication of this article.

314 **Funding:** DST-INSPIRE Faculty Award, Govt. of India

315 **Ethical Approval:** Not required

316 **References**

- 317 1. Gill TJ, Van De Velde SK, Wing DW, et al. Tibiofemoral and patellofemoral kinematics
318 after reconstruction of an isolated posterior cruciate ligament injury: In vivo analysis
319 during lunge. *Am J Sports Med* 2009; 37: 2377–2385.

- 320 2. Drake RL, Vogel AW, Mitchell AWM. *Gray's Anatomy for Students*. 4th ed.
321 Philadelphia: Elsevier, 2019.
- 322 3. Shirazi R, Shirazi-Adl A. Computational biomechanics of articular cartilage of human
323 knee joint: Effect of osteochondral defects. *J Biomech* 2009; 42: 2458–2465.
- 324 4. Smith CR, Vignos MF, Lenhart RL, et al. The influence of component alignment and
325 ligament properties on tibiofemoral contact forces in total knee replacement. *J Biomech*
326 *Eng*; 138. Epub ahead of print 2016. DOI: 10.1115/1.4032464.
- 327 5. Naghibi Beidokhti H, Janssen D, Khoshgoftar M, et al. A comparison between dynamic
328 implicit and explicit finite element simulations of the native knee joint. *Med Eng Phys*
329 2016; 38: 1123–1130.
- 330 6. Haris A, Beng Chye Tan V. Stress response envelopes of intact tibiofemoral joint and
331 knee osteoarthritis. *Proc Inst Mech Eng Part H J Eng Med* 2020; 234: 1151–1161.
- 332 7. Pierce DM, Ricken T, Holzapfel GA. A hyperelastic biphasic fibre-reinforced model of
333 articular cartilage considering distributed collagen fibre orientations: Continuum basis,
334 computational aspects and applications. *Computer Methods in Biomechanics and*
335 *Biomedical Engineering* 2013; 16: 1344–1361.
- 336 8. Jurvelin JS, Buschmann MD, Hunziker EB. Mechanical anisotropy of the human knee
337 articular cartilage in compression. *Proc Inst Mech Eng Part H J Eng Med* 2003; 217:
338 215–219.
- 339 9. Sophia Fox AJ, Bedi A, Rodeo SA. The basic science of articular cartilage: Structure,
340 composition, and function. *Sports Health* 2009; 1: 461–468.
- 341 10. Tanska P, Mononen ME, Korhonen RK. A multi-scale finite element model for
342 investigation of chondrocyte mechanics in normal and medial meniscectomy human
343 knee joint during walking. *J Biomech* 2015; 48: 1397–1406.
- 344 11. Meng Q, An S, Damion RA, et al. The effect of collagen fibril orientation on the
345 biphasic mechanics of articular cartilage. *J Mech Behav Biomed Mater* 2017; 65: 439–
346 453.
- 347 12. Changoor A, Nelea M, Méthot S, et al. Structural characteristics of the collagen network
348 in human normal, degraded and repair articular cartilages observed in polarized light and
349 scanning electron microscopies. *Osteoarthr Cartil* 2011; 19: 1458–1468.
- 350 13. Kazemi M, Dabiri Y, Li LP. Recent advances in computational mechanics of the human
351 knee joint. *Comput Math Methods Med*; 2013. Epub ahead of print 2013. DOI:
352 10.1155/2013/718423.

- 353 14. Kazemi M, Dabiri Y, Li LP. Recent Advances in Computational Mechanics of the
354 Human Knee Joint. *Comput Math Methods Med* 2013; 2013: 1–27.
- 355 15. Bursac PM, Obitz TW, Eisenberg SR, et al. Confined and unconfined stress relaxation of
356 cartilage: appropriateness of a transversely isotropic analysis. *J Biomech* 1999; 32:
357 1125–1130.
- 358 16. Shirazi R, Shirazi-Adl A, Hurtig M. Role of cartilage collagen fibrils networks in knee
359 joint biomechanics under compression. *J Biomech* 2008; 41: 3340–3348.
- 360 17. Orozco GA, Tanska P, Mononen ME, et al. The effect of constitutive representations and
361 structural constituents of ligaments on knee joint mechanics. *Sci Rep* 2018; 8: 1–15.
- 362 18. Julkunen P, Wilson W, Jurvelin JS, et al. Stress-relaxation of human patellar articular
363 cartilage in unconfined compression: Prediction of mechanical response by tissue
364 composition and structure. *J Biomech* 2008; 41: 1978–1986.
- 365 19. Halonen KS, Dzialo CM, Mannisi M, et al. Workflow assessing the effect of gait
366 alterations on stresses in the medial tibial cartilage - Combined musculoskeletal
367 modelling and finite element analysis. *Sci Rep* 2017; 7: 1–14.
- 368 20. Meng Q, Jin Z, Wilcox R, et al. Computational investigation of the time-dependent
369 contact behaviour of the human tibiofemoral joint under body weight. *Proc Inst Mech
370 Eng Part H J Eng Med* 2014; 228: 1193–1207.
- 371 21. Halonen KS, Mononen ME, Jurvelin JS, et al. Importance of patella, quadriceps forces,
372 and depthwise cartilage structure on knee joint motion and cartilage response during
373 gait. *J Biomech Eng* 2016; 138: 1–11.
- 374 22. Faisal TR, Adouni M, Dhaher YY. The effect of fibrillar degradation on the mechanics
375 of articular cartilage: a computational model. *Biomech Model Mechanobiol* 2019; 18:
376 733–751.
- 377 23. Venäläinen MS, Mononen ME, Salo J, et al. Quantitative Evaluation of the Mechanical
378 Risks Caused by Focal Cartilage Defects in the Knee. *Sci Rep* 2016; 6: 1–12.
- 379 24. Trad Z, Barkaoui A, Chafra M, et al. Finite element analysis of the effect of high tibial
380 osteotomy correction angle on articular cartilage loading. *Proc Inst Mech Eng Part H J
381 Eng Med* 2018; 232: 553–564.
- 382 25. Donahue TLH, Hull ML, Rashid MM, et al. A finite element model of the human knee
383 joint for the study of tibio-femoral contact. *J Biomech Eng* 2002; 124: 273–280.
- 384 26. Zhang K, Li L, Yang L, et al. Effect of degenerative and radial tears of the meniscus and
385 resultant meniscectomy on the knee joint: a finite element analysis. *J Orthop Transl*

- 386 2019; 18: 20–31.
- 387 27. Li L, Yang X, Yang L, et al. Biomechanical analysis of the effect of medial meniscus
388 degenerative and traumatic lesions on the knee joint. *Am J Transl Res* 2019; 11: 542–
389 556.
- 390 28. Ateshian GA, Wang H, Lai WM. The role of interstitial fluid pressurization and surface
391 porosities on the boundary friction of articular cartilage. *J Tribol* 1998; 120: 241–248.
- 392 29. Klets O, Mononen ME, Tanska P, et al. Comparison of different material models of
393 articular cartilage in 3D computational modeling of the knee: Data from the
394 Osteoarthritis Initiative (OAI). *J Biomech* 2016; 49: 3891–3900.
- 395 30. Suh JK, Disilvestro MR. Biphasic poroviscoelastic behavior of hydrated biological soft
396 tissue. *J Appl Mech Trans ASME* 1999; 66: 528–535.
- 397 31. Orozco GA, Tanska P, Mononen ME, et al. The effect of constitutive representations and
398 structural constituents of ligaments on knee joint mechanics. *Sci Rep* 2018; 8: 1–15.
- 399 32. Bolcos PO, Mononen ME, Mohammadi A, et al. Comparison between kinetic and
400 kinetic-kinematic driven knee joint finite element models. *Sci Rep* 2018; 8: 1–11.
- 401 33. Klets O, Mononen ME, Liukkonen MK, et al. Estimation of the Effect of Body Weight
402 on the Development of Osteoarthritis Based on Cumulative Stresses in Cartilage: Data
403 from the Osteoarthritis Initiative. *Ann Biomed Eng* 2018; 46: 334–344.
- 404 34. Liukkonen MK, Mononen ME, Tanska P, et al. Application of a semi-automatic
405 cartilage segmentation method for biomechanical modeling of the knee joint. *Comput*
406 *Methods Biomech Biomed Engin* 2017; 20: 1453–1463.
- 407 35. Deneweth JM, McLean SG, Arruda EM. Evaluation of hyperelastic models for the non-
408 linear and non-uniform high strain-rate mechanics of tibial cartilage. *J Biomech* 2013;
409 46: 1604–1610.
- 410 36. Jeffery AK, Blunn GW, Archer CW, et al. Three-dimensional collagen architecture in
411 bovine articular cartilage. *J Bone Jt Surg - Ser B* 1991; 73: 795–801.
- 412 37. Clark JM. Variation of collagen fiber alignment in a joint surface: A scanning electron
413 microscope study of the tibial plateau in dog, rabbit, and man. *J Orthop Res* 1991; 9:
414 246–257.
- 415 38. Mizrahi J, Maroudas A, Lanir Y, et al. The ‘instantaneous’ deformation of cartilage:
416 Effects of collagen fiber orientation and osmotic stress. *Biorheology* 1986; 23: 311–330.
- 417 39. Marchi BC, Arruda EM, Coleman RM. The Effect of Articular Cartilage Focal Defect

- 418 Size and Location in Whole Knee Biomechanics Models. *J Biomech Eng*; 142. Epub
419 ahead of print 2020. DOI: 10.1115/1.4044032.
- 420 40. Hosoda N, Sakai N, Sawae Y, et al. Finite Element Analysis of Articular Cartilage
421 Model Considering the Configuration and Biphasic Property of the Tissue. *IFMBE Proc*
422 2009; 23: 1883–1887.
- 423 41. Marchi BC, Arruda EM. A study on the role of articular cartilage soft tissue constitutive
424 form in models of whole knee biomechanics. *Biomechanics and Modeling in*
425 *Mechanobiology* 2017; 16: 117–138.
- 426 42. Halonen KS, Mononen ME, Jurvelin JS, et al. Deformation of articular cartilage during
427 static loading of a knee joint - Experimental and finite element analysis. *J Biomech*
428 2014; 47: 2467–2474.
- 429 43. Tomic A, Grillo A, Federico S. Poroelastic materials reinforced by statistically oriented
430 fibres-numerical implementation and application to articular cartilage. *IMA J Appl Math*
431 *(Institute Math Its Appl* 2014; 79: 1027–1059.
- 432 44. Raju V, Koorata PK, Kamat Y. Case study for contact pressure improvisation with
433 graded implant material in articular cartilages of knee joint. *J Mech Sci Technol* 2021;
434 35: 1049–1054.
- 435 45. Erdemir A, Sibole S. Open Knee: A Three-Dimensional Finite Element Representation
436 of the Knee Joint. *User's Guid Version 100*, <https://simtk.org/home/openknee> (2010).
- 437 46. Erdemir A. Open Knee: Open Source Modeling and Simulation in Knee Biomechanics. *J*
438 *Knee Surg* 2014; 29: 107–116.
- 439 47. Smith M. *ABAQUS/Standard User's Manual, Version 6.9*. Providence, United States:
440 Dassault Systèmes Simulia Corp, 2009.
- 441 48. Wilson W, Van Donkelaar CC, Van Rietbergen B, et al. Stresses in the local collagen
442 network of articular cartilage: A poroviscoelastic fibril-reinforced finite element study. *J*
443 *Biomech* 2004; 37: 357–366.
- 444 49. Vaziri A, Nayeb-Hashemi H, Singh A, et al. Influence of meniscectomy and meniscus
445 replacement on the stress distribution in human knee joint. *Annals of Biomedical*
446 *Engineering* 2008; 36: 1335–1344.
- 447 50. Wilson W, Van Rietbergen B, Van Donkelaar CC, et al. Pathways of load-induced
448 cartilage damage causing cartilage degeneration in the knee after meniscectomy. *J*
449 *Biomech* 2003; 36: 845–851.
- 450 51. Imeni M, Seyfi B, Fatourae N, et al. Constitutive modeling of menisci tissue: a critical

- 451 review of analytical and numerical approaches. *Biomech Model Mechanobiol*. Epub
452 ahead of print 2020. DOI: 10.1007/s10237-020-01352-1.
- 453 52. Peña E, Calvo B, Martínez MA, et al. A three-dimensional finite element analysis of the
454 combined behavior of ligaments and menisci in the healthy human knee joint. *J Biomech*
455 2006; 39: 1686–1701.
- 456 53. Łuczkiwicz P, Daszkiewicz K, Witkowski W, et al. Influence of meniscus shape in the
457 cross sectional plane on the knee contact mechanics. *J Biomech* 2015; 48: 1356–1363.
- 458 54. Morimoto Y, Ferretti M, Ekdahl M, et al. Tibiofemoral Joint Contact Area and Pressure
459 After Single- and Double-Bundle Anterior Cruciate Ligament Reconstruction. *Arthrosc -*
460 *J Arthrosc Relat Surg* 2009; 25: 62–69.
- 461 55. Marzo JM, Gurske-DePerio J. Effects of medial meniscus posterior horn avulsion and
462 repair on tibiofemoral contact area and peak contact pressure with clinical implications.
463 *Am J Sports Med* 2009; 37: 124–129.
- 464 56. Allaire R, Muriuki M, Gilbertson L, et al. Biomechanical consequences of a tear of the
465 posterior root of the medial meniscus: Similar to total meniscectomy. *J Bone Jt Surg -*
466 *Ser A* 2008; 90: 1922–1931.
- 467 57. Lee SJ, Aadalen KJ, Malaviya P, et al. Tibiofemoral contact mechanics after serial
468 medial meniscectomies in the human cadaveric knee. *Am J Sports Med* 2006; 34: 1334–
469 1344.
- 470 58. Paletta GA, Manning T, Snell E, et al. The effect of autograft meniscal replacement on
471 intraarticular contact area and pressures in the human knee: A biomechanical study. *Am*
472 *J Sports Med* 1997; 25: 692–698.
- 473 59. Halonen KS, Mononen ME, Töyräs J, et al. Optimal graft stiffness and pre-strain restore
474 normal joint motion and cartilage responses in ACL reconstructed knee. *J Biomech*
475 2016; 49: 2566–2576.
- 476 60. Chen AC, Bae WC, Schinagl RM, et al. Depth- and strain-dependent mechanical and
477 electromechanical properties of full-thickness bovine articular cartilage in confined
478 compression. *J Biomech* 2001; 34: 1–12.
- 479 61. Yang NH, Canavan PK, Nayeb-Hashemi H. The effect of the frontal plane tibiofemoral
480 angle and varus knee moment on the contact stress and strain at the knee cartilage. *J*
481 *Appl Biomech* 2010; 26: 432–443.
- 482 62. Yang NH, Canavan PK, Nayeb-Hashemi H, et al. Protocol for constructing subject-
483 specific biomechanical models of knee joint. *Comput Methods Biomech Biomed Engin*

- 484 2010; 13: 589–603.
- 485 63. Mononen ME, Mikkola MT, Julkunen P, et al. Effect of superficial collagen patterns and
486 fibrillation of femoral articular cartilage on knee joint mechanics-A 3D finite element
487 analysis. *J Biomech* 2012; 45: 579–587.
- 488 64. Liukkonen MK, Mononen ME, Klets O, et al. Simulation of subject-specific progression
489 of knee osteoarthritis and comparison to experimental follow-up data: Data from the
490 osteoarthritis initiative. *Sci Rep* 2017; 7: 1–14.
- 491 65. Mononen ME, Jurvelin JS, Korhonen RK. Implementation of a gait cycle loading into
492 healthy and meniscectomised knee joint models with fibril-reinforced articular cartilage.
493 *Comput Methods Biomech Biomed Engin* 2015; 18: 141–152.
- 494
- 495