



Citation for published version:

Talbott, H, Jha, S, Gulati, A, Brockett, C, Mangwani, J & Pegg, E 2023, 'Clinically useful finite element models of the natural ankle – a review', *Clinical Biomechanics*, vol. 106, 106006.
<https://doi.org/10.1016/j.clinbiomech.2023.106006>

DOI:

[10.1016/j.clinbiomech.2023.106006](https://doi.org/10.1016/j.clinbiomech.2023.106006)

Publication date:

2023

Document Version

Peer reviewed version

[Link to publication](#)

Publisher Rights

CC BY-NC-ND

University of Bath

Alternative formats

If you require this document in an alternative format, please contact:
openaccess@bath.ac.uk

General rights

Copyright and moral rights for the publications made accessible in the public portal are retained by the authors and/or other copyright owners and it is a condition of accessing publications that users recognise and abide by the legal requirements associated with these rights.

Take down policy

If you believe that this document breaches copyright please contact us providing details, and we will remove access to the work immediately and investigate your claim.

Clinically useful finite element models of the natural ankle – a review

Harriet Talbott¹, Shilpa Jha², Aashish Gulati³, Claire Brockett⁴, Jitendra Mangwani², Elise C Pegg^{5*}

¹ School of Engineering, University of Hull, Hull, UK

² University Hospitals of Leicester, Leicester, UK

³ Sandwell and West Birmingham Hospitals NHS Trust, Birmingham, UK

⁴ Department of Mechanical Engineering, University of Sheffield, Sheffield, UK

⁵ Department of Mechanical Engineering, University of Bath, Bath, UK

* Corresponding author: Elise C Pegg, University of Bath, Claverton Down, Bath, BA2 7AY, UK. Email: e.c.pegg@bath.ac.uk

Abstract

Background

Biomechanical simulation of the foot and ankle complex is a growing research area but compared to simulation of joints such as hip and knee, it has been under investigated and lacks consistency in research methodology. The methodology is variable, data is heterogenous and there are no clear output criteria. Therefore, it is very difficult to correlate clinically and draw meaningful inferences.

Methods

The focus of this review is finite element simulation of the native ankle joint and we will explore: the different research questions asked, the model designs used, ways the model rigour has been ensured, the different output parameters of interest and the clinical impact and relevance of these studies.

Findings

The 72 published studies explored in this review demonstrate wide variability in approach. Many studies demonstrated a preference for simplicity when representing different tissues, with the

27 majority using linear isotropic material properties to represent the bone, cartilage and ligaments; this
28 allows the models to be complex in another way such as to include more bones or complex loading.
29 Most studies were validated against experimental or in vivo data, but a large proportion (40%) of
30 studies were not validated at all, which is an area of concern.

31 *Interpretation*

32 Finite element simulation of the ankle shows promise as a clinical tool for improving outcomes.
33 Standardisation of model creation and standardisation of reporting would increase trust, and enable
34 independent validation, through which successful clinical application of the research could be realised.

35 Keywords: Ankle; Finite Element; Clinical relevance; Modelling; Orthopaedics

36 Abstract word count: 241

37 Main text word count: 5889

38 1 Introduction

39 Finite Element (FE) Analysis of the foot and ankle has been a rapidly growing field in the past two
40 decades, and it shows great promise as a tool to inform clinical planning and treatment. As simulation
41 studies do not put the patient at any risk, it is possible to assess the potential of more risky and radical
42 treatments, and to predict patient-specific outcomes of surgical treatment enabling optimisation prior
43 to intervention. The purpose of this literature review was to capture the current trends in simulation
44 studies of the natural ankle, and explore ways in which their clinical translation can be facilitated.

45 1.1 Review methodology

46 The literature was identified from the Scopus database using the search words 'ankle' AND 'finite
47 element', OR 'in silico', OR 'computational' OR 'tibiotalar'. No limit was set on the publication date
48 and the search was performed in December 2022; this resulted in 144 documents. These were then
49 reviewed and papers excluded if:

- 50 • The primary focus of the paper was on a prosthesis rather than the natural ankle
- 51 • The study did not include the natural ankle or whole foot
- 52 • Finite element methods were not used as a primary part of the study

53 This resulted in 72 studies which were investigated; themes within these were identified and used to
54 structure this article.

55 1.2 Clinical Need

56 FE models have been utilised extensively in the knee to evaluate the biomechanics of the cruciate
57 ligaments [3] and there is certainly a role for models in the ankle to similarly research the
58 pathomechanics of ankle instability and ligament injury. The effect of each sequential ligament injury
59 (lateral, syndesmotic and medial) in multi-ligament patterns, as well as repair of each in turn, could
60 be examined to determine the effect on joint displacement and contact stresses, providing
61 information on optimal repair strategy, prognosis, and how aggressively such injuries should be

62 treated. The timing of surgery after ligament injury of the ankle has been identified as a priority for
63 UK Foot & Ankle research [4]. Furthermore, FE models could be utilised to optimise techniques (e.g.
64 number of bone anchors and bone anchor positions in lateral ligament reconstruction, effect of
65 fibretape or hamstring augmentation of lateral ligament reconstruction, screws versus dynamic
66 stabilisation of the syndesmosis) as well as quantify the effect of bony alignment (e.g. heel varus) and
67 the impact of correction. FE could also be a useful tool in examining fixation at the enthesis – with
68 previous groups using this method to evaluate and compare rotator cuff repair methods [5]. In this
69 way FE could also examine treatment of insertional Achilles tendinopathy and re-attachment
70 strategies.

71 Osteoporotic ankle fractures are the third most common fragility fracture [6] with an anticipated
72 increase in incidence with an aging population. FE models have been used to evaluate fixation
73 methods in osteoporotic bone in the proximal humerus and proximal femur [7]. An *in silico* ankle
74 model based on osteoporotic bone could provide useful insights in optimising fixation and minimising
75 risks of fatigue failure of implants. Locking technology, fibula-pro-tibia screws, number of screws and
76 alternative fixation strategies such as tibio-talo-calcaneal nailing and fibular nailing could be examined
77 to assess strain across the fracture site as well as loading of the implant expanding our understanding
78 of these difficult injuries and the most biomechanically sound strategies for fixing them.

79 In the longer term, patient-specific FE models derived from the individual's CT may have a role both
80 in augmenting the design of customised instrumentation in arthroplasty, as well as pre-operative
81 planning in re-alignment surgery to enhance biomechanical outcome for example in supramalleolar
82 osteotomy for treatment of eccentric ankle arthritis.

83 One of the critical barriers to the translation of FE models into clinical practice is trust; clinicians need
84 to be sure that the data from a simulation can be relied upon to be correct before any treatment
85 decisions are made. A first step towards this is to improve clinical understanding of how such models

86 work, what they can (and cannot) represent, and what questions to ask in order to ascertain how
87 trustworthy a model and its results are.

88 1.3 Biomechanics Simulation Overview

89 Finite element (FE) research of the ankle has focussed on a range of applications such as impact [8, 9],
90 injury [10-14], surgery [15-20], shoe [21, 22] or orthotic [23] design, and arthroplasty [24-31].
91 Depending on their application, these models have been formulated of several structures (Figure 1);
92 some models consider only one bone [25, 29, 32], while others model just the bones of the ankle [1,
93 15, 16, 18, 28, 30, 31, 33-45], and some have generated whole foot models which often include the
94 soft tissues encapsulating the bones [2, 8, 9, 11-14, 21-23, 46-66]. The level of complexity required
95 from any model is dependent on the research question being asked and it is the role of the research
96 team to ensure the model captures all the key factors which may affect the result. This has led to large
97 variability in research methods limiting the opportunities for independent validation.

98 Reproducibility and standardisation are contentious issues within orthopaedic modelling [67];
99 reporting methods to ensure they are reproducible and robust is key to increasing confidence in model
100 outputs. However, this does not come without its challenges and there is currently a lack of
101 standardisation of methods in FE of the foot and ankle. Differences between methods start from
102 model generation and can be seen through to results analysis.

103 The initial step in creating a FE model is replication of the patient tissue geometry within the computer
104 from medical scans. Digitisation of the patient geometry can be achieved through segmentation of
105 both Magnetic Resonance Imaging (MRI) and Computed Tomography (CT) images using a range of
106 software; this provides accurate patient specific tissue geometries. Once the geometry has been
107 digitised, the 3D models generated are then discretised or 'meshed' to enable numerical calculation
108 (Figure 2). Clinically relevant loads and material properties can be then input to the model, and lastly
109 the model can be solved (Figure 3). There are many different software packages available for each of
110 these steps, and no standardised combination used within the published research.

111 The geometry of the foot and ankle is highly complex; hence, many simplifications are made at various
112 stages of the modelling process. This review aims to demonstrate the processes required to produce
113 a reproducible, standardised foot or ankle model for use in clinical applications (Figure 4). In recent
114 years, the uptake of *in silico* methods has increased vastly in foot and ankle research, hence it is
115 important that these models are appropriately addressing their research questions and that the
116 biomechanics community work together to ensure clinically relevant and consistent data.

117 2 Model Generation

118 The processes involved in model generation are key to an accurate simulation being possible, ensuring
119 geometries are correctly represented. MRI or CT have been used separately or in combination to
120 generate the 3D volume from which finite element models are produced. Bone is more appropriately
121 modelled by CT, and cartilage from MRI, due to them being the respective gold standards for obtaining
122 geometry. The optimal model composition would therefore be bone extracted from CT, and cartilage
123 from MRI, however this has associated costs for additional scanner time for per research participant.
124 Multimodal image coregistration CT can be automated and carried out in licensed software such as
125 Matlab (The Math Works Inc.) or freeware such as ImageJ (NIH). Scanner hardware can be
126 programmed so that the image resolutions are similar for the two image modalities, however, some
127 postprocessing would still be involved, meaning there would also be additional researcher time
128 required.

129 There have been no studies published contrasting models generated entirely from CT or MRI on the
130 same foot and ankle geometry to understand as to what extent this influences model outputs. Other
131 imaging parameters, such as slice thickness, slice gap, and voxel size, will also influence model
132 accuracy. However, this information is rarely cited in literature and is not standardised due to different
133 scanners set-ups; this is a current knowledge gap in the literature which needs to be filled. A recent
134 literature review by Barkaoui *et al.* explored several different imaging modalities used to create
135 orthopaedic finite element model and emphasised that each modality has its strengths and

136 weaknesses [68]. MRI scans are much more suited to representing soft tissues as these are rarely
137 visible within CT scans, but CT is more suited to represent hard tissues such as bone. In relation to the
138 foot-ankle complex they discuss an unusual study by Wong *et al.* [69] who created an MRI-based FE
139 model to look at talus and calcaneus impact fracture risk; due to the modality chosen the authors
140 could not represent the heterogeneous material properties of the bones so they simplified this to
141 homogeneous orthotropic properties for the cortical and trabecular regions. Interestingly the model
142 was successfully validated against experimental data from a cadaveric experiment, though the
143 cadaveric results had large variability.

144 Conversely, models generated from CT need to make assumptions about the soft tissue elements of a
145 model, for example, utilising a uniform cartilage thickness [44, 45, 70] or excluding cartilage from the
146 model [10, 12, 17, 23, 48, 49, 71]. From MRI, bone boundaries are less clearly defined, but subject
147 specific cartilage segmentations can be carried out for a more accurate representation of the
148 geometry. This soft tissue definition can be taken advantage of, with some models including
149 segmentations of ligaments [57] and muscles [50, 56]. The ligaments are commonly simplified to linear
150 spring elements, trusses or excluded from models [15, 28, 37, 41, 51, 72], while muscles are usually
151 accounted for in the forces applied in the models, and their absence noted as a limitation of the
152 methodology. This is deemed appropriate unless the aim of the model is to consider the functions of
153 the muscle [50].

154 2.1 Discretisation

155 FE models discretise the structure into small elements with simple shapes. In orthopaedics, this
156 complex geometry is commonly discretised into tetrahedral or hexahedral elements which have
157 robust meshing algorithms. These can be described as first order, or second order, which relates to
158 the number of nodes present in an element; as the nodes are the locations at which the partial
159 differential equations are solved, an increased number of nodes would suggest an increased model
160 accuracy. There is a limit to this, and this is known as the point at which the mesh converges; it is

161 important to show that the meshing parameters do not influence the model outputs, while keeping
162 the computational expense as low as possible.

163 One mesh convergence of a human foot model was found in the literature [73]; the study considered
164 five mesh densities, using the same element types, under different foot loading conditions. The
165 findings showed a sensitivity of the mesh convergence to the loading conditions; however, it was
166 suggested a 2.5 mm edge length may be appropriate for most models. This falls within the region of
167 studies using converged global mesh densities, where target edge lengths have ranged from the region
168 of 1 to 1.5 mm [37, 39, 41], to 3 mm [48] or 4 mm [18, 32]. On the face of it an element size of 2.5 mm
169 might be considered far too large to get accurate results considering the dimensions of different
170 tissues within the foot, but when considering discretisation it is essential to always consider what
171 geometry is being represented. In this study the authors were interested in the contact stress
172 underneath the toes, and the inner geometry of the structures within the foot were greatly simplified
173 enabling a relatively large element size to be used. So although the literature gives a general idea of
174 commonly used mesh sizes, but it is important to note that the optimal mesh size to use will always
175 depend on the particular model, the research question, and the output parameter of interest.

176 The variability based on the different loading conditions highlights the importance of assessing for a
177 converged mesh, however, very few studies [15, 25] detail if a mesh convergence has been carried
178 out. Studies tend to only report element type, with some giving total number of elements or nodes
179 [62, 63, 71, 74], or element edge lengths. In studies that detail the total number of nodes or elements,
180 it is unclear if the element size is constant throughout the volume, or whether the model uses different
181 mesh densities in different regions. For example, higher mesh densities where articulations occur [19,
182 20, 56], in thinner regions to avoid the loss of morphology [50], or in regions of interest for outputs
183 [25, 31, 43]. If it is the latter then knowing the overall number of elements is insufficient information
184 to define the mesh.

185 Relatively few studies report the additional mesh qualities of aspect ratios, internal angles, and
186 Jacobians [12, 32, 56]. A perfectly shaped tetrahedral element would have an aspect ratio of 1, a
187 Jacobian of 1 and internal angles that do not deviate greatly from 60 degrees. There are other factors
188 to consider in mesh quality checks such as skewness that do not seem to have been reported for any
189 foot and ankle models. Finite element models with large aspect ratios, or negative Jacobians may not
190 produce accurate results and can lead to localised outliers in the results data, and consequently mesh
191 quality checks should be considered when generating a model for clinical use.

192 2.2 Material Properties

193 The level of detail required from material properties depends on the model application; simplifications
194 are common due to the complex nature of biological tissues. Where there are unknowns, or
195 assumptions made, sensitivity analyses should be carried out to ensure these do not influence the
196 outcomes of studies. There have been examples in foot and ankle literature of sensitivity studies
197 regarding cartilage thickness [51], frictional properties [19, 20, 29], loading conditions [60, 64], and
198 material properties [12, 33, 34, 47, 59, 61, 63]. Other studies considering osteochondral defects have
199 also checked sensitivity to defect size [35, 40].

200 The sensitivity to material properties is key, as properties are often simplified due to the lack of raw
201 data on mechanical properties. Properties for the bone and cartilage are often simplified to linear
202 elastic models, with hyper-elastic models sometimes used for the soft tissues. When simplified to
203 homogenous isotropic materials, the properties of bone (Table 1) often assume an elastic modulus (E)
204 of 7,300 MPa. This value originates from a weighted average of the cortical and trabecular elastic
205 moduli in a healthy ankle [75]. Some studies have considered treating all bone as cortical bone [35,
206 53], while others have used significantly higher unreferenced values [14]. One study has compared
207 the influence of four different elastic moduli (7,300 MPa, 14,600 MPa, 21,900 MPa, and 29,200 MPa)
208 on a foot and ankle model [59], the study found increased elastic moduli could lead to an
209 underestimation of bone stresses and strains.

210 When modelling cortical and trabecular bone as separate linear elastic homogenous isotropic
211 materials, there is less agreement on the properties to be used. The most common cortical bone
212 elastic modulus is 17,000 MPa, which has been used in combination with trabecular values ranging
213 from 400 to 700 MPa. However, these models assume that all cortical and trabecular bone has one
214 constant material property, which is a significant simplification of the real tissue [76].

215 Bone material models can be further improved by considering the anisotropic nature of bone [57], or
216 modelling more precisely the inhomogeneity. Hounsfield Unit (HU) based properties from CT are likely
217 to provide the most accurate representation of bone across the model, provided they are correctly
218 calibrated. Heterogenous material assignment has been applied in some studies, where bone is
219 modelled with subject specific mechanical properties [15, 19, 20, 29, 32, 33, 39, 51, 72]. This, however,
220 increases the model complexity and for those models not considering bone outputs this may be an
221 unnecessary expense, but if a patient is osteoporotic or the focus of the study relates directly to the
222 bone such as fracture healing then it is crucial that the distribution of mechanical properties be
223 accurately assigned.

224 Cartilage, when included in a model, is simplified in its properties (Table 2); either linear elastic, or
225 hyper-elastic properties are used, however these both still assume isotropic behaviour. Articular
226 cartilage is both orthotropic [77] and biphasic [78] in nature. The biphasic nature of cartilage has
227 previously been investigated in representative 2D [79] or 3D [80] FE models, not macroscale joint
228 models; these characteristics would be extremely complex to translate to a whole foot or ankle model.

229 There are a considerable number of models of the natural foot and ankle that do not include additional
230 structures to represent the cartilage [10, 12, 17, 22, 23, 48, 49, 51, 53, 74], instead frictionless contact
231 definitions between bony segments are used to represent the lubricated nature of the articular
232 cartilage surfaces during loading [17]. This simplification may be appropriate in some modelling
233 scenarios where the outputs of interest are in the soft tissues or ligamentous structures.

234 When cartilage is modelled as a linear elastic material, both the elastic modulus (E) and Poisson's ratio
235 (ν) vary. The elastic modulus represents by how much a material will deform in response to a particular
236 load, and the Poisson's ratio represents by how much a material volume compresses in response to
237 load, where a material with a Poisson's ratio of 0.5 is completely incompressible and if it is -1 then
238 the material has no resistance to volume change. These values independently influence the
239 mechanical behaviour of the cartilage and in turn impact the model results. With the exception of one
240 study [37] cartilage has been modelled with a significantly lower elastic modulus than the bone. The
241 Poisson's ratios used tend to fall within a range of 0.4 to 0.49 which shows that cartilage does not
242 undergo much volume change under load but it can readily change shape enabling it to conform to
243 the articular surfaces while providing impact resistance. This is largely accepted as appropriate for
244 modelling cartilage in the foot and ankle; however, when considering cartilage outputs such as contact
245 pressures a hyper-elastic material model would be more appropriate as it can replicate the time-
246 dependent properties of the tissue deformation under load. There is no agreement in literature on
247 the foot and ankle as to which model is the most appropriate for this application, with Yeoh, Neo-
248 Hookean, Mooney-Rivlin and Ogden all used. In a 2D simulation carried out on the knee [81], Neo-
249 Hookean and Mooney-Rivlin models were compared; through validation against a mathematical
250 model it was seen that non-linear Neo-Hookean provided the most appropriate model for
251 characterising cartilage behaviour.

252 Ligaments are commonly modelled as trusses with a cross sectional area, with all ligaments modelled
253 with the same Elastic Modulus; 260MPa is used most commonly [14, 21, 23, 48, 49, 54, 55, 58, 59];
254 however, 20MPa [40] and 350 MPa [74] have also been reported in literature. For added complexity
255 some studies have given each ligament its own Elastic Modulus, these have used a variety of ranges
256 including 7 to 18.44 MPa [36], 100 to 320 MPa [53] and 99.5 to 512 MPa [1, 11].

257 Another method of modelling ligaments is using linear spring elements; these are defined with
258 stiffness values (Table 3). Some of the stiffness values have been derived from mechanical test data,

259 but for ligaments with no associated mechanical testing data a stiffness of 70 N/mm has been
260 assumed. There is some variation between studies with different sources of original values [18], but it
261 is not possible to identify the impact of this variation on the model outputs due to other differences
262 in the study designs.

263 In the models that include soft tissue structures, these tend to be the focus of the outputs, hence
264 efforts are made to accurately represent the hyper-elastic nature of plantar tissues. The two hyper-
265 elastic models used were Ogden [11, 22, 46, 82] and Second Order Polynomial [21, 54, 56, 63, 64, 74].
266 There are reports that simplify the soft tissues properties to a homogenous isotropic linear elastic
267 material [14, 55, 83], however, this simplification might influence any plantar contact pressure outputs
268 should this be a focus of the study.

269 2.3 Contact Definitions

270 In models composed of more than one part, the contact definitions are key to how the parts interact
271 with one another. The interaction between bone and cartilage is most frequently defined using tie
272 constraints, which means that there is no relative movement between the two parts treating them as
273 one entity. As previously mentioned, there are numerous foot models that do not require these tie
274 constraints as they do not include cartilage components. The tangential contact properties they define
275 between bone to reproduce articular motion are also required between articular cartilage surfaces.
276 These are largely defined as frictionless [38, 52, 54, 55, 58, 63, 74, 84], or with a low coefficient of
277 friction such as 0.001 [40], 0.01 [2, 18, 30, 35, 36, 44, 47], or 0.02 [12]. Higher coefficients of friction
278 (0.1) have also been assumed in some models [43, 57], which is still within the bounds of coefficients
279 of friction found experimentally [85].

280 Some models have also included normal contact behaviours, which have predominantly been defined
281 as “hard” [15, 38, 41, 51, 52] using either the Penalty or Augmented Lagrange algorithm. Some models
282 also use a non-linear normal contact definition [33, 47, 49]. These contacts tend to be defined using a

283 surface-to-surface discretisation rather than node-to-surface, as it produces lower errors for contact
284 pressures, however, it is more computationally expensive.

285 2.4 Loading and Boundary Conditions

286 Once the contact is defined to allow for relative movement at articulating regions, the loading and
287 boundary conditions must be appropriately defined to simulate motion. Correctly representing the
288 loading conditions of the foot and ankle is a complex task due to the number of joints and articulations
289 to consider with the force distribution [86].

290 These loading and boundary conditions can be used to simulate both static and dynamic motion;
291 examples include replicating surgical procedures [41] and experimental setups [8, 12, 25, 43, 64], such
292 as anterior drawer tests [34, 57], or simulating biomechanical data [2, 11, 15, 21, 32, 51-55, 58, 82,
293 83]. The application of biomechanical data to models has been both dynamic and simplified to quasi-
294 static at points in the gait cycle depending on the outputs of interest.

295 This data is either scaled to represent a subject specific bodyweight (BW), or an average BW, which
296 tends to be set at 600 N [1, 16, 35, 36, 38, 47]. A proportion of this BW is used in static models
297 simulating neutral standing. These have been done with various degrees of simplifications and
298 assumptions depending on the model setup. For example, in models that include soft tissues, and a
299 flat surface to represent the ground, 50% BW might be applied to the ground which has a contact
300 defined with the plantar surface [14, 23]. In many of these cases, an opposite load is applied through
301 the Achilles to represent the Achilles tendon force [33, 62-64, 74]. Other loading conditions considered
302 have been single leg standing, where 100% BW is applied [36, 71], up to 320% BW [44], and 520% BW
303 [29, 72], which was selected to represent the peak load experienced in stance phase of gait [29]. In all
304 cases, some sort of constraint must be applied to the model to stop free body movement.

305 Some models, loaded proximally, have also taken into consideration that the load transfer through
306 the tibia, and through the fibula is not equal. For example, one study simulating a participant
307 bodyweight of 98 kg used the assumption that 84.3 % BW is transferred through the tibial column,

308 and 15.7% through the fibula. Giving a tibial loading of 810.44 N, and a load of 150.94 N through the
309 fibula [37]. A second study used slightly different proportions, with 93.59 %BW transferred through
310 the tibia, and 6.41 %BW transferred through the fibula [40]. This is not a split that could be accounted
311 for in the models loaded distally.

312 2.5 Solving

313 The underlying numerical calculations performed by the simulation software, whether an implicit [31,
314 43, 57] or explicit [12, 46, 53, 83] solver is used, is another important difference in models of the foot
315 and ankle, but surprisingly it is not frequently reported which is another barrier to reproducibility.
316 These two solver types calculate the state of the model in different ways, taking time into
317 consideration differently. Therefore, the two different solvers may not give the same solution to a
318 simulation, require different verification, and they are appropriate for different strain rates.

319 Simulations of the foot and ankle have been carried out using dynamic, static, and quasi-static
320 methods, depending on the application and the required level of complexity. For example, those
321 models considering impact [8, 9] require dynamic simulations; however, gait has been considered
322 using both quasi-static and dynamic methods depending on whether the whole gait cycle is simulated
323 or independent events of stance phase.

324 3 Results analysis (post-processing)

325 Once the simulation is carried out, appropriate postprocessing and interpretation of results is a vital
326 next step enabling clinical assertions to be made from models. The relevant output depends on the
327 question raised, and the tissue of interest. Despite models often including all structures of the foot
328 and ankle, results might only consider one specific region, for example the plantar tissues, or a specific
329 bony region. These regions can have one or more output reported for them, however, the justification
330 for the choice of output is not always explicitly clear with how it relates to the research question.

331 The postprocessing of results is key in delivering clinically relevant outputs; clear examples of outputs
332 chosen due to their application are micromotions when considering joint fixations [19], bone
333 remodelling in total ankle replacements [72], or plantar pressures when looking at shoe or insole
334 designs [62, 63]. The interaction with a device, and understanding of potential failure mechanisms due
335 to this, give rise to obvious postprocessing choices. However, this is not always obvious, and studies
336 may consider an output based on the ability to carry out validation [64] with a more tenuous clinical
337 significance.

338 Understanding the clinical relevance of specific model outputs is a key starting point in the
339 postprocessing of results, as FE software has the capacity to potentially produce hundreds of output
340 types for a mechanical model. In orthopaedics stresses, strains, and contact pressures are common
341 outputs of interest which would all have clinical relevance. These, however, are generic terms, and
342 each will have types (e.g. principal stresses), as well as directional components, which will have
343 differing relationship with tissue health. Von Mises Stress is the default stress output in most FE solvers
344 and is consequently frequently used to consider orthopaedic model outputs, without any comment
345 on whether this is appropriate. Von Mises Stress is a good indicator of stress distribution, and stress
346 concentration location. However, it does not indicate whether stresses are occurring in compression,
347 tension, or shear, which is highly relevant to orthopaedics as bone has high compressive strength but
348 poor tensile and shear strengths. Furthermore when modelling bone as a heterogeneous material, the
349 von Mises results will be dependent on the modulus assignment and so strain-based outputs are more
350 clinically informative.

351 The FE method is best implemented when there is a clear failure mode, and therefore stresses or
352 strains – once validated – can be directly contrasted with this, to suggest the likelihood of failure under
353 a certain condition. For example, excessive strains in ligaments might highlight the potential for
354 overstretching or rupture. Or in bone, changes in stresses and strains leading to bone remodelling can

355 be implemented using bone adaptation laws which can highlight potential changes in bone mass after
356 treatment[15, 72].

357 The applicability of these results is partially governed by the material properties used; for example,
358 contact pressures and their distribution may not be appropriate in regions where linear elastic
359 material properties have been used. This is because this material property influences the contact
360 mechanics, which directly relates to the contact pressures. Careful consideration must be taken when
361 defining material properties that are appropriate for the purpose of the model. Therefore, it is
362 important that the research question is defined, and failure mechanism considered, at the beginning
363 of the modelling process.

364 4 Validation

365 Care must be taken in models with clinical implications that input parameters accurately represent
366 the model and simulation [87], to ensure that the outputs of a model are meaningful. Alongside this
367 it is pertinent that the model is validated, to draw any conclusions, or make any predictions from its
368 outputs. Validation is the process of determining the degree to which the model is an accurate
369 representation of the real world from the perspective of the intended use for the model [88]. Once a
370 validation is successfully achieved, there will be increased confidence in the interpretation and value
371 of the results.

372 Validation against previously published data is the most common technique used in foot and ankle
373 models (Figure 5), while assessing against a gold standard experimental technique is the preferred
374 validation method; model-model validation [59] is also possible, but less frequently used.

375 The importance of validation is much reported in literature [89-91], however, some foot and ankle
376 studies do not report validation methods [13, 34, 35, 41, 44, 65, 66, 70, 92], appear not to be validated,
377 or state the lack of validation is a limitation of the model [2, 15, 17, 19, 26, 31, 37, 72, 74, 82, 93].
378 Models where this is the case tend to only be able to consider the relative change between cases using

379 the same methodology, and not absolute values. They are therefore still valuable, but not to clinical
380 practice.

381 In those that do carry out *in vivo* or *in vitro* validation, this may be the main objective of the model
382 generation [45], or be possible due to the data collected for input parameters; such as joint angles [9],
383 loading [8] or GRF [58]. The standard for validation is to carry this out for the primary model output
384 [90], such as; plantar pressures [46, 56, 94], peak strains [10, 25] or joint contact pressures [45, 71].

385 Of the studies which have been validated experimentally they can be split into five main groups:

- 386 1. **High experimental variability:** where the FE results match the experimental data, but the
387 experimental data is highly variable (particularly common with cadaver experiments) so it is
388 hard to know for sure if the FE results and trends are representative [8, 43, 59, 95].
- 389 2. **Large difference between model and experiment:** where the difference between the model
390 and the experiment, for some or all data, is high (>10%) [12, 28, 71].
- 391 3. **Trends validated but inaccurate magnitude:** where the model shows the same trends but the
392 values output are very different to the experimental data [64, 94].
- 393 4. **Insufficient experiment sample size:** where the model and the experiment matches, but the
394 variability in the experimental data is uncertain [45].
- 395 5. **True match between experiment and model data:** where the experimental data has been
396 acquired robustly and the FE results closely match [25, 40, 42, 57].

397 Many of the above validated models are still useful and are able to answer clinical questions but it is
398 important that studies are open about the limitations behind their model validation.

399 Validations carried out against literature will be limited to studies considering the primary output; or
400 may guide users to select a primary output based on what can be validated, rather than the research
401 question. The simulation is also limited to replicating the model setup used in the previously reported
402 study, as model parameters such as loading conditions and material properties should be kept in
403 common for the purpose of validation. After the model is validated, it is then possible to ascertain the

404 sensitivity to any parameter changed. For example, parametric tests could be used to examine the
405 biomechanical effects of different sizes, shapes, and positions of osteochondral defects of the talus,
406 which may assist our surgical decision-making as well as provide guidance on the likely natural history
407 without intervention.

408 These validated models then have the capacity to accelerate the research area, without the use of
409 new resources each iteration, due to one model having the potential to simulate a multitude of
410 scenarios.

411 5 Clinical relevance

412 Of the studies explored in this review, few have led to changes in clinical practice. Viceconti *et al.*
413 suggested some basic requirements for a clinically useful numerical model [89]: 1) clear rationale
414 behind why a numerical model is needed, 2) evidence that the model has been verified, 3) clinically
415 relevant input parameters, 4) sensitivity analysis to understand how uncertainty (patient and other
416 variability) impacts on the model, 5) experimental validation of the data, 6) risk-benefit analysis so
417 clinicians can quantitatively assess the risk involved in using the model. None of the papers examined in
418 this literature review met all of these requirements, but also many did not seek to answer a direct
419 clinical question and instead were focused on the methodology of the model development [45, 94,
420 95] with a promise of later application. This approach can be fruitful but relies upon work continuing
421 into the long term, and if a model were to be developed for general use it would need to be sufficiently
422 flexible and accessible to enable a variety of clinical questions to be answered.

423 The Open Knee model [96] is an example of an orthopaedic FE model that has allowed for clinical
424 translation; this open-source model has allowed investigators outside of the developers to utilise the
425 model in research relating to the biomechanics and tissue structures of the knee joint. This has not
426 only expedited the research capacity in the field but has standardised some aspects of model
427 generation. An equivalent model, to the best of the authors' knowledge, is not available for the whole

428 foot or ankle; therefore, the authors propose a similar strategy is required for this joint to improve
429 translation into clinical practice. This could be either a single patient specific model, or several models
430 to create an in-silico population. Alongside this a general template or checklist on good practice when
431 creating a model of the ankle could improve the clinical usefulness of models; but as evidenced by
432 Viceconti's paper, such recommendations are not always followed. Our view is that provision of
433 standard materials that make it easier to create good quality models of the ankle, alongside guidelines,
434 is the way forward.

435 6 Conclusion

436 The value of foot and ankle modelling is clear in clinical practice; however, to date the translation has
437 not been as successful as various other orthopaedic or cardiovascular systems. Finite element models
438 of the foot and ankle are becoming increasingly prevalent, however, the lack of standardisation and
439 clear reporting in the field hinders its success and wider application. Therefore, a conclusion of this
440 review relates to the improvement of these aspects to allow for better clinical translation.

441 This is highly challenging, due to the complexity of the foot and ankle; hence, models tend to be
442 purpose built with simplifications to elements of lesser interest in the model. Therefore, often models
443 may not be appropriate for processing results relating to the bone or cartilage as their complex
444 material properties are often simplified, or not modelled – as is sometimes the case with cartilage in
445 whole foot models. These simplifications and their potential impact on results are often brushed over,
446 however, are key in understanding if a model is robust and can therefore provide inciteful, clinically
447 relevant outcomes.

448 Acknowledgements

449 Financial support was provided by Orthopaedic Research UK (ORUK).

450 References

- 451 1. Li, J., et al., *The effect of talus osteochondral defects of different area size on ankle joint*
452 *stability: a finite element analysis*. BMC Musculoskeletal Disorders, 2022. **23**(1): p. 500.
- 453 2. Wang, C., et al., *Three-Dimensional Finite Element Analysis and Biomechanical Analysis of*
454 *Midfoot von Mises Stress Levels in Flatfoot, Clubfoot, and Lisfranc Joint Injury*. Med Sci Monit,
455 2021. **27**: p. e931969.
- 456 3. Pfeiffer, F.M., *The Use of Finite Element Analysis to Enhance Research and Clinical Practice in*
457 *Orthopedics*. J Knee Surg, 2016. **29**(2): p. 149-58.
- 458 4. Mangwani, J., *James Lind Alliance Priority Setting Partnership - 'Top 10' research priorities in*
459 *foot and ankle surgery*. Journal of Trauma and Orthopaedics, 2022. **10**(1).
- 460 5. Tits, A. and D. Ruffoni, *Joining soft tissues to bone: Insights from modeling and simulations*.
461 Bone Rep, 2021. **14**: p. 100742.
- 462 6. Sporer, S.M., J.N. Weinstein, and K.J. Koval, *The geographic incidence and treatment variation*
463 *of common fractures of elderly patients*. J Am Acad Orthop Surg, 2006. **14**(4): p. 246-55.
- 464 7. Lewis, G.S., et al., *Finite Element Analysis of Fracture Fixation*. Curr Osteoporos Rep, 2021.
465 **19**(4): p. 403-416.
- 466 8. Bandak, F.A., R.E. Tannous, and T. Toridis, *On the development of an osseo-ligamentous finite*
467 *element model of the human ankle joint*. International Journal of Solids and Structures, 2001.
468 **38**(10): p. 1681-1697.
- 469 9. Huang, J., C. Huang, and F. Mo, *Analysis of Foot-Ankle-Leg Injuries in Various Under-Foot*
470 *Impact Loading Environments With a Human Active Lower Limb Model*. Journal of
471 Biomechanical Engineering, 2022. **144**(1).
- 472 10. Wei, F., et al., *Development and Validation of a Computational Model to Study the Effect of*
473 *Foot Constraint on Ankle Injury due to External Rotation*. Annals of Biomedical Engineering,
474 2011. **39**(2): p. 756-765.
- 475 11. Bae, J.Y., et al., *Analysis of the Effects of Normal Walking on Ankle Joint Contact Characteristics*
476 *After Acute Inversion Ankle Sprain*. Annals of Biomedical Engineering, 2015. **43**(12): p. 3015-
477 3024.
- 478 12. Smolen, C. and C.E. Quenneville, *A Finite Element Model of the Foot/Ankle to Evaluate Injury*
479 *Risk in Various Postures*. Annals of Biomedical Engineering, 2017. **45**(8): p. 1993-2008.
- 480 13. Zhou, L., et al., *Finite element analysis of the influence of anterior talofibular ligament injury*
481 *on Hook test results*. Injury, 2017. **48**(7): p. 1499-1502.
- 482 14. Sun, D., et al., *Workflow assessing the effect of Achilles tendon rupture on gait function and*
483 *metatarsal stress: Combined musculoskeletal modeling and finite element analysis*.
484 Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in
485 Medicine, 2022. **236**(5): p. 676-685.
- 486 15. Anderson, R.T., et al., *Finite element analysis of a pseudoelastic compression-generating*
487 *intramedullary ankle arthrodesis nail*. Journal of the Mechanical Behavior of Biomedical
488 Materials, 2016. **62**: p. 83-92.
- 489 16. Zhou, L., et al., *Biomechanical effect of anterior talofibular ligament injury in Weber B lateral*
490 *malleolus fractures after lateral plate fixation: A finite element analysis*. Foot and Ankle
491 Surgery, 2020. **26**(8): p. 871-875.
- 492 17. Mercan, N., A. Yildirim, and Y. Dere, *Biomechanical Analysis of Tibiofibular Syndesmosis Injury*
493 *Fixation Methods: A Finite Element Analysis*. The Journal of Foot and Ankle Surgery, 2022.
- 494 18. Ni, M., et al., *Quantitative initial safety range of early passive rehabilitation after ankle*
495 *fracture surgery*. Injury, 2022. **53**(6): p. 2281-2286.
- 496 19. Vázquez, A.A., et al., *Finite element analysis of the initial stability of ankle arthrodesis with*
497 *internal fixation: flat cut versus intact joint contours*. Clinical Biomechanics, 2003. **18**(3): p.
498 244-253.

- 499 20. Alonso-Vázquez, A., et al., *Initial stability of ankle arthrodesis with three-screw fixation. A finite*
500 *element analysis*. Clinical Biomechanics, 2004. **19**(7): p. 751-759.
- 501 21. Yu, J., et al., *Biomechanical simulation of high-heeled shoe donning and walking*. Journal of
502 Biomechanics, 2013. **46**(12): p. 2067-2074.
- 503 22. Nonogawa, M., et al., *Developing a three-dimensional numerical foot model and identifying*
504 *the loading condition for designing a stable sole for running shoes*. Mechanical Engineering
505 Journal, 2021. **8**(5): p. 21-00200-21-00200.
- 506 23. Peng, Y., et al., *Computational models of flatfoot with three-dimensional fascia and bulk soft*
507 *tissue interaction for orthosis design*. Medicine in Novel Technology and Devices, 2021. **9**: p.
508 100050.
- 509 24. Reggiani, B., et al., *Finite element analysis of a total ankle replacement during the stance phase*
510 *of gait*. Journal of Biomechanics, 2006. **39**(8): p. 1435-1443.
- 511 25. Terrier, A., et al., *Development and experimental validation of a finite element model of total*
512 *ankle replacement*. Journal of Biomechanics, 2014. **47**(3): p. 742-745.
- 513 26. Martinelli, N., et al., *Contact stresses, pressure and area in a fixed-bearing total ankle*
514 *replacement: a finite element analysis*. BMC Musculoskeletal Disorders, 2017. **18**(1): p. 493.
- 515 27. Martinelli, N., et al., *Contact stresses in a fixed-bearing total ankle replacement: A finite*
516 *element analysis*. Foot and Ankle Surgery, 2017. **23**: p. 7-8.
- 517 28. Taghizadeh, Y., et al., *Total ankle replacement along with subtalar joint arthrodesis: In-vitro*
518 *and in-silico biomechanical investigations*. International Journal for Numerical Methods in
519 Biomedical Engineering, 2021. **37**(9): p. e3514.
- 520 29. Jyoti, S. Mondal, and R. Ghosh, *Biomechanical analysis of three popular tibial designs for TAR*
521 *with different implant-bone interfacial conditions and bone qualities: A finite element study*.
522 Medical Engineering & Physics, 2022. **104**: p. 103812.
- 523 30. Liu, T., et al., *The evaluation of artificial talus implant on ankle joint contact characteristics: a*
524 *finite element study based on four subjects*. Medical & Biological Engineering & Computing,
525 2022. **60**(4): p. 1139-1158.
- 526 31. van Hoogstraten, S.W.G., et al., *Malalignment of the total ankle replacement increases peak*
527 *contact stresses on the bone-implant interface: a finite element analysis*. BMC Musculoskeletal
528 Disorders, 2022. **23**(1): p. 463.
- 529 32. Xu, C., et al., *An Integrated Musculoskeletal-Finite-Element Model to Evaluate Effects of Load*
530 *Carriage on the Tibia During Walking*. Journal of Biomechanical Engineering, 2016. **138**(10).
- 531 33. Mondal, S. and R. Ghosh, *A numerical study on stress distribution across the ankle joint: Effects*
532 *of material distribution of bone, muscle force and ligaments*. Journal of Orthopaedics, 2017.
533 **14**(3): p. 329-335.
- 534 34. Ji, Y., et al., *Analysis of 3-dimensional finite element after reconstruction of impaired ankle*
535 *deltoid ligament*. Exp Ther Med, 2016. **12**(6): p. 3913-3916.
- 536 35. Li, J., Y. Wei, and M. Wei, *Finite Element Analysis of the Effect of Talar Osteochondral Defects*
537 *of Different Depths on Ankle Joint Stability*. Med Sci Monit, 2020. **26**: p. e921823.
- 538 36. Guan, M., et al., *Finite element analysis of the effect of sagittal angle on ankle joint stability in*
539 *posterior malleolus fracture: A cohort study*. International Journal of Surgery, 2019. **70**: p. 53-
540 59.
- 541 37. Kılıçaslan, Ö.F., et al., *Effect of cartilage thickness mismatch in osteochondral grafting from*
542 *knee to talus on articular contact pressures: A finite element analysis*. Joint diseases and
543 related surgery, 2021. **32**(2): p. 355-362.
- 544 38. Wu, X., et al., *Finite Element Analysis of a Novel Approach for Knee and Ankle Protection during*
545 *Landing*. Applied Sciences, 2021. **11**(4): p. 1912.
- 546 39. Zhang, Y., et al., *Articular geometry can affect joint kinematics, contact mechanics, and*
547 *implant-bone micromotion in total ankle arthroplasty*. Journal of Orthopaedic Research, 2022.
548 **n/a**(n/a).

- 549 40. Ramos, A., C. Rocha, and M. Mesnard, *The effect of osteochondral lesion size and ankle joint*
550 *position on cartilage behavior - numerical and in vitro experimental results*. Medical
551 Engineering & Physics, 2021. **98**: p. 73-82.
- 552 41. Bing, F., et al., *Biomechanical finite element analysis of typical tibiotalar arthrodesis*. Medicine
553 in Novel Technology and Devices, 2021. **11**: p. 100087.
- 554 42. Palazzi, E., et al., *Estimating the stabilizing function of ankle and subtalar ligaments via a*
555 *morphology-specific three-dimensional dynamic model*. Journal of Biomechanics, 2020. **98**: p.
556 109421.
- 557 43. Forestiero, A., et al., *Investigation of the biomechanical behaviour of hindfoot ligaments*.
558 Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in
559 Medicine, 2013. **227**(6): p. 683-692.
- 560 44. Li, W., et al., *Patient-specific finite element analysis of chronic contact stress exposure after*
561 *intraarticular fracture of the tibial plafond*. Journal of Orthopaedic Research, 2008. **26**(8): p.
562 1039-1045.
- 563 45. Anderson, D.D., et al., *Physical validation of a patient-specific contact finite element model of*
564 *the ankle*. Journal of Biomechanics, 2007. **40**(8): p. 1662-1669.
- 565 46. Mo, F., et al., *A three-dimensional finite element foot-ankle model and its personalisation*
566 *methods analysis*. International Journal of Mechanical Sciences, 2022. **219**: p. 107108.
- 567 47. Malakoutikhah, H., E. Madenci, and L.D. Latt, *The impact of ligament tears on joint contact*
568 *mechanics in progressive collapsing foot deformity: A finite element study*. Clinical
569 Biomechanics, 2022. **94**: p. 105630.
- 570 48. Peng, Y., et al., *Biomechanical comparison among five mid/hindfoot arthrodeses procedures*
571 *in treating flatfoot using a musculoskeletal multibody driven finite element model*. Computer
572 Methods and Programs in Biomedicine, 2021. **211**: p. 106408.
- 573 49. Peng, Y., et al., *Influence of arch support heights on the internal foot mechanics of flatfoot*
574 *during walking: A muscle-driven finite element analysis*. Computers in Biology and Medicine,
575 2021. **132**: p. 104355.
- 576 50. Bocanegra, M.A.M., et al., *Numerical Assessment of the Structural Effects of Relative Sliding*
577 *between Tissues in a Finite Element Model of the Foot*. Mathematics, 2021. **9**(15): p. 1719.
- 578 51. Muralidharan, L., et al., *A patient-specific numerical model of the ankle joint for the analysis*
579 *of contact pressure distribution*. Proceedings of the Institution of Mechanical Engineers, Part
580 H: Journal of Engineering in Medicine, 2020. **234**(9): p. 909-920.
- 581 52. Park, S., et al., *Finite element analysis of knee and ankle joint during gait based on motion*
582 *analysis*. Medical Engineering & Physics, 2019. **63**: p. 33-41.
- 583 53. Chen, T.L.-W., et al., *Foot arch deformation and plantar fascia loading during running with*
584 *rearfoot strike and forefoot strike: A dynamic finite element analysis*. Journal of Biomechanics,
585 2019. **83**: p. 260-272.
- 586 54. Wang, Y., et al., *Finite element analysis of biomechanical effects of total ankle arthroplasty on*
587 *the foot*. Journal of Orthopaedic Translation, 2018. **12**: p. 55-65.
- 588 55. Akrami, M., et al., *Subject-specific finite element modelling of the human foot complex during*
589 *walking: sensitivity analysis of material properties, boundary and loading conditions*.
590 Biomechanics and Modeling in Mechanobiology, 2018. **17**(2): p. 559-576.
- 591 56. Morales-Orcajo, E., et al., *Non-linear finite element model to assess the effect of tendon forces*
592 *on the foot-ankle complex*. Medical Engineering & Physics, 2017. **49**: p. 71-78.
- 593 57. Forestiero, A., et al., *Numerical model for healthy and injured ankle ligaments*. Australasian
594 Physical & Engineering Sciences in Medicine, 2017. **40**(2): p. 289-295.
- 595 58. Chen, Y.-N., et al., *Finite Element Analysis of Plantar Fascia During Walking: A Quasi-static*
596 *Simulation*. Foot & Ankle International, 2015. **36**(1): p. 90-97.
- 597 59. Niu, W.X., et al., *Effects of Bone Young's Modulus on Finite Element Analysis in the Lateral*
598 *Ankle Biomechanics*. Applied Bionics and Biomechanics, 2013. **10**: p. 818414.

- 599 60. Ramlee, M.H., M.R.A. Kadir, and H. Harun. *Three-dimensional modeling and analysis of a*
600 *human ankle joint*. in *2013 IEEE Student Conference on Research and Development*. 2013.
- 601 61. laquinto, J.M. and J.S. Wayne, *Computational Model of the Lower Leg and Foot/Ankle*
602 *Complex: Application to Arch Stability*. Journal of Biomechanical Engineering, 2010. **132**(2).
- 603 62. Cheung, J.T.-M. and M. Zhang, *A 3-dimensional finite element model of the human foot and*
604 *ankle for insole design*. Archives of Physical Medicine and Rehabilitation, 2005. **86**(2): p. 353-
605 358.
- 606 63. Cheung, J.T.-M., et al., *Three-dimensional finite element analysis of the foot during standing—*
607 *a material sensitivity study*. Journal of Biomechanics, 2005. **38**(5): p. 1045-1054.
- 608 64. Cheung, J.T.-M., M. Zhang, and K.-N. An, *Effect of Achilles tendon loading on plantar fascia*
609 *tension in the standing foot*. Clinical Biomechanics, 2006. **21**(2): p. 194-203.
- 610 65. Cheung, J.T.M., M. Zhang, and B.M. Nigg, *Three-dimensional finite element analysis of the*
611 *human foot and ankle during the stance phases of gait*. Journal of Biomechanics, 2006. **39**: p.
612 S180.
- 613 66. Camacho, D.L., et al., *A three-dimensional, anatomically detailed foot model: a foundation for*
614 *a finite element simulation and means of quantifying foot-bone position*. J Rehabil Res Dev,
615 2002. **39**(3): p. 401-10.
- 616 67. Halloran, J.P., et al., *Assessment of reporting practices and reproducibility potential of a cohort*
617 *of published studies in computational knee biomechanics*. Journal of Orthopaedic Research,
618 2022. **41**(2): p. 325-334.
- 619 68. Barkaoui, A., I. Ait Oumghar, and R. Ben Kahla, *Review on the use of medical imaging in*
620 *orthopedic biomechanics: finite element studies*. Computer Methods in Biomechanics and
621 Biomedical Engineering: Imaging & Visualization, 2021. **9**(5): p. 535-554.
- 622 69. Wong, D.W.-C., et al., *Finite Element Analysis of Foot and Ankle Impact Injury: Risk Evaluation*
623 *of Calcaneus and Talus Fracture*. PLOS ONE, 2016. **11**(4): p. e0154435.
- 624 70. Goreham-Voss, C.M., T.O. McKinley, and T.D. Brown, *A finite element exploration of cartilage*
625 *stress near an articular incongruity during unstable motion*. Journal of Biomechanics, 2007.
626 **40**(15): p. 3438-3447.
- 627 71. Filardi, V., *Tibio talar contact stress: An experimental and numerical study*. Journal of
628 Orthopaedics, 2020. **17**: p. 44-48.
- 629 72. Bouguecha, A., et al., *Numerical simulation of strain-adaptive bone remodelling in the ankle*
630 *joint*. BioMedical Engineering OnLine, 2011. **10**(1): p. 58.
- 631 73. CHEN, W.-M., et al., *OPTIMAL MESH CRITERIA IN FINITE ELEMENT MODELING OF HUMAN*
632 *FOOT: THE DEPENDENCE FOR MULTIPLE MODEL OUTPUTS ON MESH DENSITY AND LOADING*
633 *BOUNDARY CONDITIONS*. Journal of Mechanics in Medicine and Biology, 2021. **21**(09): p.
634 2140034.
- 635 74. Filardi, V., *Finite element analysis of the foot: Stress and displacement shielding*. Journal of
636 Orthopaedics, 2018. **15**(4): p. 974-979.
- 637 75. Gefen, A., et al., *Biomechanical Analysis of the Three-Dimensional Foot Structure During Gait:*
638 *A Basic Tool for Clinical Applications*. Journal of Biomechanical Engineering, 2000. **122**(6): p.
639 630-639.
- 640 76. Taddei, F., et al., *The material mapping strategy influences the accuracy of CT-based finite*
641 *element models of bones: An evaluation against experimental measurements*. Medical
642 Engineering & Physics, 2007. **29**(9): p. 973-979.
- 643 77. Fischenich, K.M., et al., *Human articular cartilage is orthotropic where microstructure,*
644 *micromechanics, and chemistry vary with depth and split-line orientation*. Osteoarthritis
645 Cartilage, 2020. **28**(10): p. 1362-1372.
- 646 78. Ateshian, G.A., et al., *The correspondence between equilibrium biphasic and triphasic material*
647 *properties in mixture models of articular cartilage*. Journal of Biomechanics, 2004. **37**(3): p.
648 391-400.

- 649 79. Wu, J.Z., W. Herzog, and S. Federico, *Finite element modeling of finite deformable, biphasic*
650 *biological tissues with transversely isotropic statistically distributed fibers: toward a practical*
651 *solution*. Z Angew Math Phys, 2016. **67**.
- 652 80. Hislop, B.D., C.M. Heveran, and R.K. June, *Development and analytical validation of a finite*
653 *element model of fluid transport through osteochondral tissue*. bioRxiv, 2020: p.
654 2020.10.26.356188.
- 655 81. Butz, K.D., et al., *Stress distributions and material properties determined in articular cartilage*
656 *from MRI-based finite strains*. Journal of Biomechanics, 2011. **44**(15): p. 2667-2672.
- 657 82. Chen, W.-M., et al., *Role of gastrocnemius–soleus muscle in forefoot force transmission at heel*
658 *rise — A 3D finite element analysis*. Journal of Biomechanics, 2012. **45**(10): p. 1783-1789.
- 659 83. Qian, Z., et al., *A Dynamic Finite Element Analysis of Human Foot Complex in the Sagittal Plane*
660 *during Level Walking*. PLOS ONE, 2013. **8**(11): p. e79424.
- 661 84. Xu, C., et al., *Biomechanical evaluation of tenodesis reconstruction in ankle with deltoid*
662 *ligament deficiency: a finite element analysis*. Knee Surgery, Sports Traumatology,
663 Arthroscopy, 2012. **20**(9): p. 1854-1862.
- 664 85. Krishnan, R., M. Kopacz, and G.A. Ateshian, *Experimental verification of the role of interstitial*
665 *fluid pressurization in cartilage lubrication*. Journal of Orthopaedic Research, 2004. **22**(3): p.
666 565-570.
- 667 86. Barton, T., F. Lintz, and I. Winson, *Biomechanical changes associated with the osteoarthritic,*
668 *arthrodeseed, and prosthetic ankle joint*. Foot Ankle Surg, 2011. **17**(2): p. 52-7.
- 669 87. Morales-Orcajo, E., J. Bayod, and E. Barbosa de Las Casas, *Computational Foot Modeling:*
670 *Scope and Applications*. Archives of Computational Methods in Engineering, 2016. **23**(3): p.
671 389-416.
- 672 88. Schwer, L.E., *An overview of the PTC 60/V&V 10: guide for verification and validation in*
673 *computational solid mechanics*. Engineering with Computers, 2007. **23**(4): p. 245-252.
- 674 89. Viceconti, M., et al., *Extracting clinically relevant data from finite element simulations*. Clinical
675 Biomechanics, 2005. **20**(5): p. 451-454.
- 676 90. Erdemir, A., et al., *Considerations for reporting finite element analysis studies in biomechanics*.
677 Journal of Biomechanics, 2012. **45**(4): p. 625-633.
- 678 91. Cooper, R.J., R.K. Wilcox, and A.C. Jones, *Finite element models of the tibiofemoral joint: A*
679 *review of validation approaches and modelling challenges*. Medical Engineering & Physics,
680 2019. **74**: p. 1-12.
- 681 92. Giddings, V.L., et al., *Calcaneal loading during walking and running*. Medicine & Science in
682 Sports & Exercise, 2000. **32**(3): p. 627-634.
- 683 93. Hannah, I., et al., *Evaluation of a Kinematically-Driven Finite Element Footstrike Model*. Journal
684 of Applied Biomechanics, 2016. **32**(3): p. 301-305.
- 685 94. Liang, J., et al., *Deformation and stress distribution of the human foot after plantar ligaments*
686 *release: A cadaveric study and finite element analysis*. Science China Life Sciences, 2011. **54**(3):
687 p. 267-271.
- 688 95. Liacouras, P.C. and J.S. Wayne, *Computational Modeling to Predict Mechanical Function of*
689 *Joints: Application to the Lower Leg With Simulation of Two Cadaver Studies*. Journal of
690 Biomechanical Engineering, 2007. **129**(6): p. 811-817.
- 691 96. Erdemir, A., *Open Knee: Open Source Modeling and Simulation in Knee Biomechanics*. The
692 journal of knee surgery, 2016. **29**(2): p. 107-116.

693

695 Table 1 Summary of bone properties reported for foot and ankle models

	E (MPa)		ν		References
Homogenous isotropic	7,300		0.3		[1, 11, 21, 22, 28, 54, 55, 62-64, 74, 82, 83]
	7,000		0.3		[58]
	17,000		0.3		[35, 53]
	10,000		0.3		[23, 48, 49]
	29,200		0.3		[14]
Cortical and Trabecular	Cortical	Trabecular			
	19,100	1,000	0.3		[37]
	19,000	531	0.3		[8]
	17,000	700	0.3		[17, 56, 71]
	17,000	500	0.3		[31]
	17,000	477	0.3		[18]
	17,000	400	0.3		[12, 40]
	13,000	1,000	0.3		[41]
	12,100	530	0.3		[36]
	7300	1100	0.3		[60]
Isotropic Cortical, CT greyscale Trabecular	Cortical	Trabecular			
	19,000		0.3		[29, 33]
	Cortical	Trabecular	Cortical	Trabecular	

Anisotropic cortical bone	E1= 17,500	E2= 450	$\nu_{12}=0.31$ $\nu_{13}=0.43$	0.3	[43] G13= 3,500MPa
Isotropic trabecular bone	E3=11,500				
CT Greyscale Based					[15, 19, 20, 32, 39, 51]
Cortical, cancellous and Interface densities (ρ)	$3790 \cdot \rho^3$				[72]

696

697 *Table 2 Summary of cartilage properties reported for foot and ankle models*

	E (MPa)	ν	References
Linear Elastic	0.7	0.49	[8, 34, 84]
	0.83	0.49	[36]
	1	0.4	[2, 14, 21, 28, 54-56, 62-64, 82]
	10	0.4	[11, 29, 33, 56]
	10	0.45	[40]
	12	0.4	[18]
	12	0.42	[1, 35, 44, 45]
	12	0.45	[52]
	50	0.1	[58]
	1230	0.42	[37]
Hyper Elastic	Yeoh: C1=23.42MPa, C2=21.89MPa, C3= 5.28MPa, C4=29.94MPa, C5= 20.09MPa,		[43]

	D1=25.60MPa, D2=215.14MPa, D3=215.89MPa, D4=254.18MPa and D5=31.66MPa.	
	Neo-Hookean: 0.49, $C_{10} = E/(4(1+n))$	[83]
	Mooney-Rivlin: $C_{01}=0.41$ MPa and $C_{10}=4.1$ MPa	[60]
	Ogden: $\mu=2.43$, $\alpha=12.45$, $D=0.176$	[30]

698

699 *Table 3 Spring stiffness, k (N/mm) reported in literature where ligaments were modelled as*
 700 *linear spring elements*

	[18]	[33]	[60]	[95]
Anterior talofibular	142		90	90
Anterior tibiofibular	78	90	78	90 (proximal), 70 (distal)
Anterior tibiotalar	123	90	70	70
Calcaneofibular	70	70	70	70
Interosseous 1-4	400	400	400	400
Interosseous talocalcaneal		70	70	70
Lateral talocalcaneal	70	70	70	70
Medial talocalcaneal	70	70	70	70
Posterior talocalcaneal	70	70	70	70
Posterior talofibular	82	70	70	70
Posterior tibiofibular	101	90	101	90 (proximal), 101 (distal)
Posterior tibiotalar	60	80	80	80
Tibiocalcaneal	122	122	122	122

701

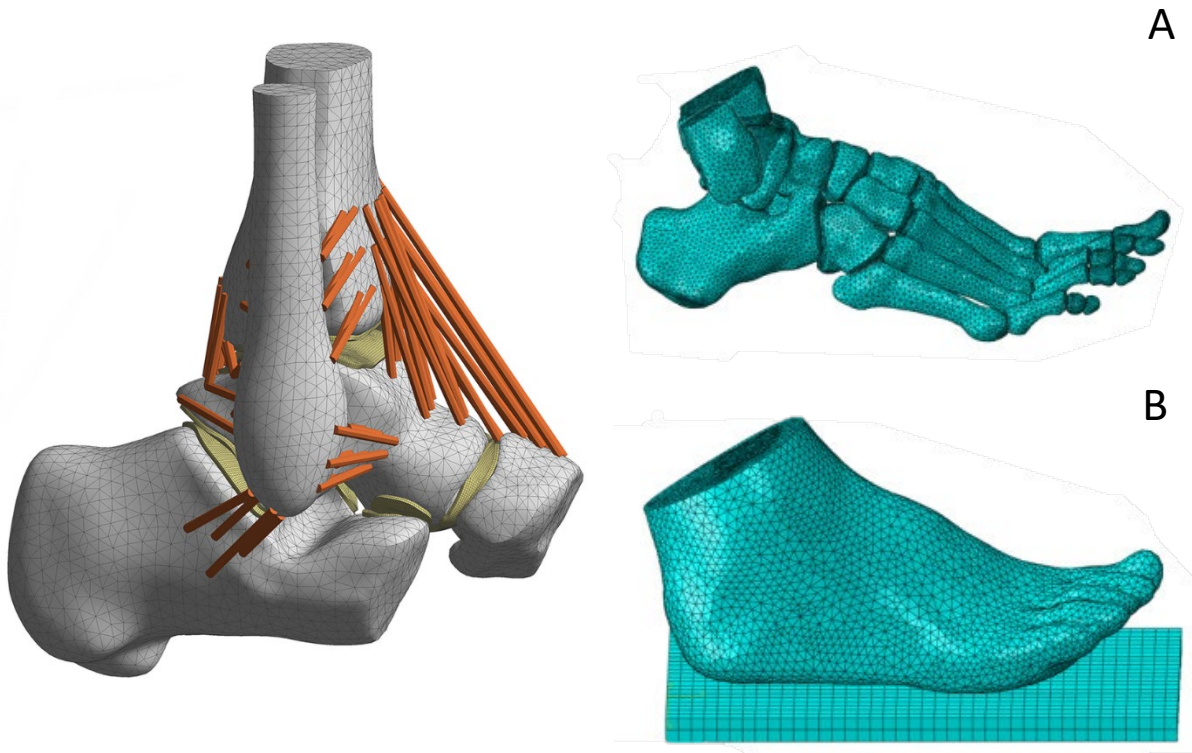


Figure 1 FE model of the bones of the ankle with 3D renderings of 2D truss elements used for ligaments (left) [1], and of the whole foot (right) adapted from Wang et al. where A) shows the underlying bony structures of the foot and ankle modelled, and B) shows the encapsulated tissues and rigid plate included in the simulation[2].

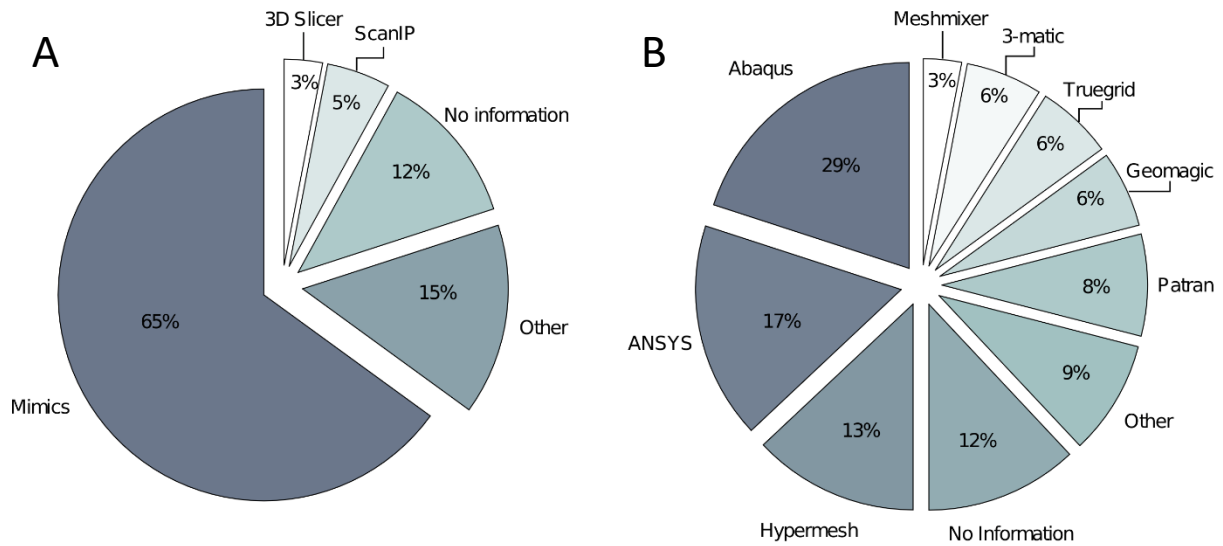


Figure 2 Proportion of software used for A) segmentation and B) meshing based on analysis of foot and ankle FE literature, between 2001 and June 2022

703 7
704

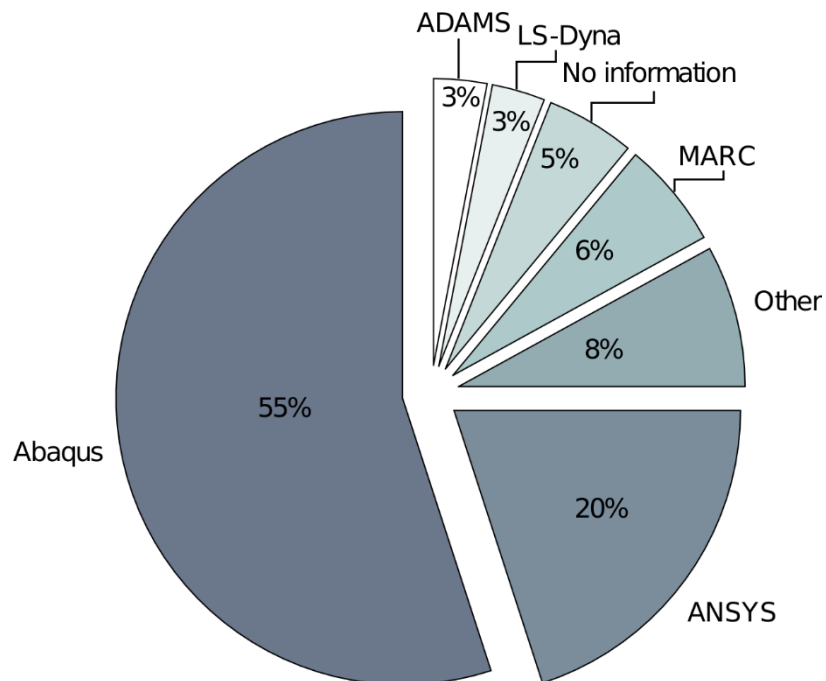
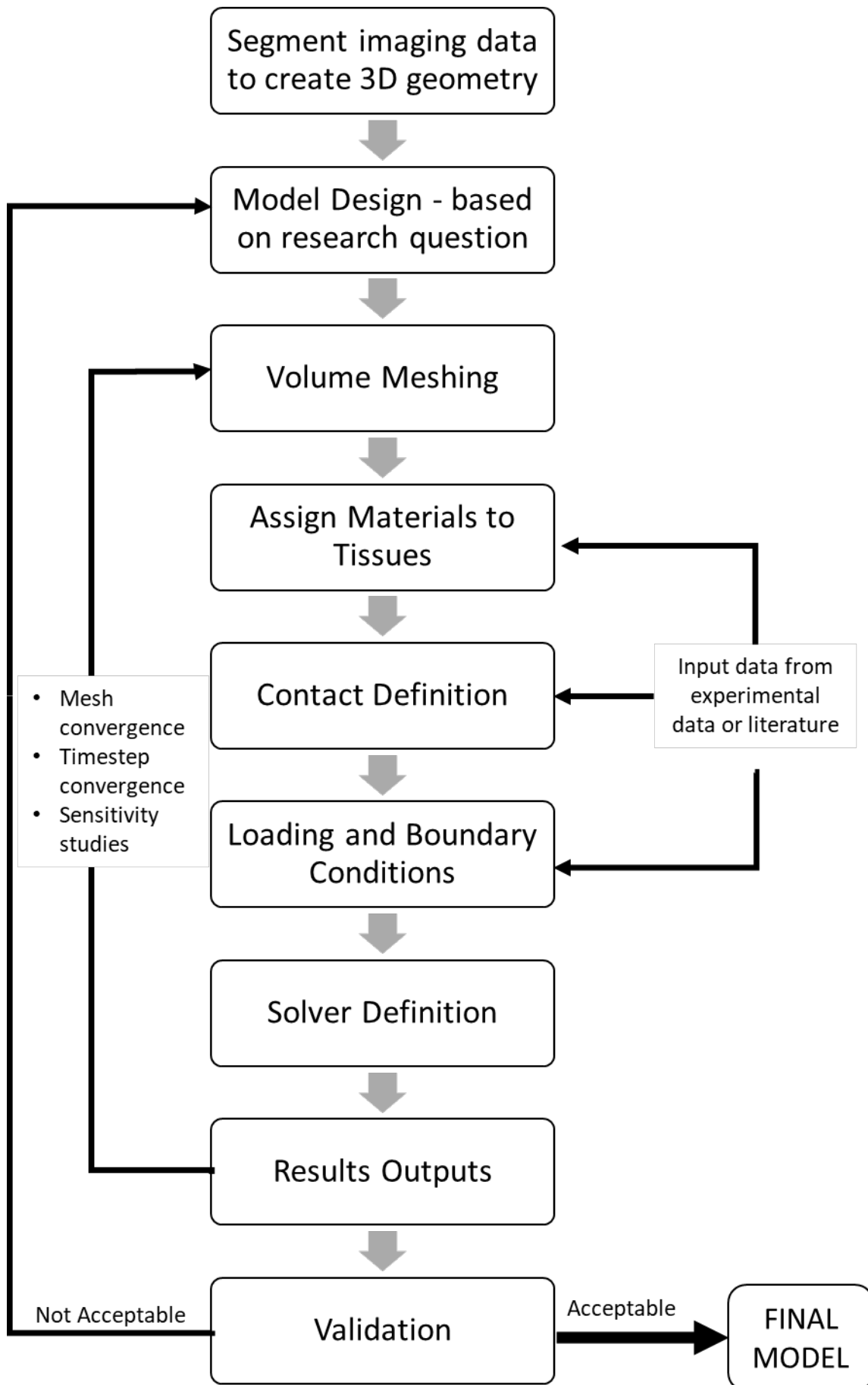
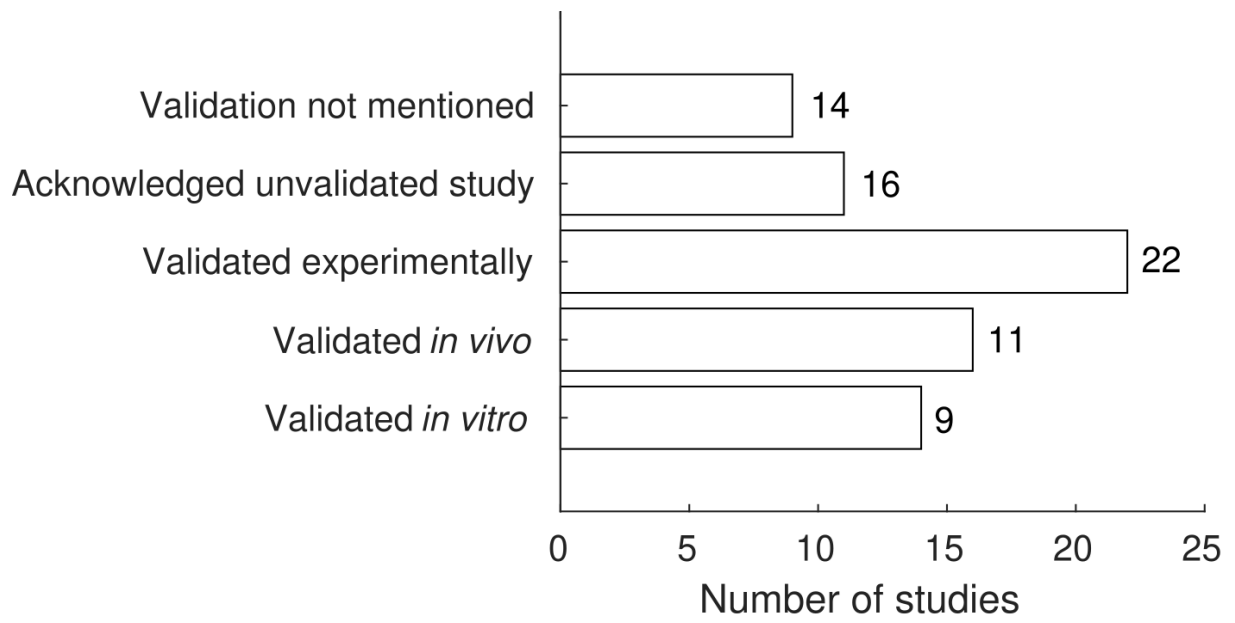


Figure 3 Proportion of finite element software used in foot and ankle literature, between 2001 and June 2022

705



706 Figure 4 Flowchart of processes to generate a validated finite element model



707

Figure 5 Number of studies in literature reporting methods of validating foot and ankle models