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

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

Interpretation

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

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

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1 Introduction

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

1.1 Review methodology

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

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

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

1.2 Clinical Need

FE models have been utilised extensively in the knee to evaluate the biomechanics of the cruciate ligaments [3] and there is certainly a role for models in the ankle to similarly research the pathomechanics of ankle instability and ligament injury. The effect of each sequential ligament injury (lateral, syndesmotic and medial) in multi-ligament patterns, as well as repair of each in turn, could be examined to determine the effect on joint displacement and contact stresses, providing information on optimal repair strategy, prognosis, and how aggressively such injuries should be treated. The timing of surgery after ligament injury of the ankle has been identified as a priority for UK Foot & Ankle research [4]. Furthermore, FE models could be utilised to optimise techniques (e.g. number of bone anchors and bone anchor positions in lateral ligament reconstruction, effect of fibretape or hamstring augmentation of lateral ligament reconstruction, screws versus dynamic stabilisation of the syndesmosis) as well as quantify the effect of bony alignment (e.g. heel varus) and the impact of correction. FE could also be a useful tool in examining fixation at the enthesis – with previous groups using this method to evaluate and compare rotator cuff repair methods [5]. In this way FE could also examine treatment of insertional Achilles tendinopathy and re-attachment strategies.

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

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

One of the critical barriers to the translation of FE models into clinical practice is trust; clinicians need to be sure that the data from a simulation can be relied upon to be correct before any treatment decisions are made. A first step towards this is to improve clinical understanding of how such models work, what they can (and cannot) represent, and what questions to ask in order to ascertain how trustworthy a model and its results are.

1.3 Biomechanics Simulation Overview

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

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

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

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

2 Model Generation

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

There have been no studies published contrasting models generated entirely from CT or MRI on the same foot and ankle geometry to understand as to what extent this influences model outputs. Other imaging parameters, such as slice thickness, slice gap, and voxel size, will also influence model accuracy. However, this information is rarely cited in literature and is not standardised due to different scanners set-ups; this is a current knowledge gap in the literature which needs to be filled. A recent literature review by Barkaoui *et al.* explored several different imaging modalities used to create orthopaedic finite element model and emphasised that each modality has its strengths and weaknesses [68]. MRI scans are much more suited to representing soft tissues as these are rarely visible within CT scans, but CT is more suited to represent hard tissues such as bone. In relation to the foot-ankle complex they discuss an unusual study by Wong *et al.* [69] who created an MRI-based FE model to look at talus and calcaneus impact fracture risk; due to the modality chosen the authors could not represent the heterogeneous material properties of the bones so they simplified this to homogeneous orthotropic properties for the cortical and trabecular regions. Interestingly the model was successfully validated against experimental data from a cadaveric experiment, though the cadaveric results had large variability.

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

2.1 Discretisation

FE models discretise the structure into small elements with simple shapes. In orthopaedics, this complex geometry is commonly discretised into tetrahedral or hexahedral elements which have robust meshing algorithms. These can be described as first order, or second order, which relates to the number of nodes present in an element; as the nodes are the locations at which the partial differential equations are solved, an increased number of nodes would suggest an increased model accuracy. There is a limit to this, and this is known as the point at which the mesh converges; it is important to show that the meshing parameters do not influence the model outputs, while keeping the computational expense as low as possible.

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

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

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

2.2 Material Properties

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

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

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

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

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

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

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

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

Another method of modelling ligaments is using linear spring elements; these are defined with stiffness values (Table 3). Some of the stiffness values have been derived from mechanical test data, but for ligaments with no associated mechanical testing data a stiffness of 70 N/mm has been assumed. There is some variation between studies with different sources of original values [18], but it is not possible to identify the impact of this variation on the model outputs due to other differences in the study designs.

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

2.3 Contact Definitions

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

Some models have also included normal contact behaviours, which have predominantly been defined as “hard” [15, 38, 41, 51, 52] using either the Penalty or Augmented Lagrange algorithm. Some models also use a non-linear normal contact definition [33, 47, 49]. These contacts tend to be defined using a surface-to-surface discretisation rather than node-to-surface, as it produces lower errors for contact pressures, however, it is more computationally expensive.

2.4 Loading and Boundary Conditions

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

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

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

Some models, loaded proximally, have also taken into consideration that the load transfer through the tibia, and through the fibula is not equal. For example, one study simulating a participant bodyweight of 98 kg used the assumption that 84.3 % BW is transferred through the tibial column, and 15.7% through the fibula. Giving a tibial loading of 810.44 N, and a load of 150.94 N through the fibula [37]. A second study used slightly different proportions, with 93.59 %BW transferred through the tibia, and 6.41 %BW transferred through the fibula [40]. This is not a split that could be accounted for in the models loaded distally.

2.5 Solving

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

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

3 Results analysis (post-processing)

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

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

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

The FE method is best implemented when there is a clear failure mode, and therefore stresses or strains – once validated – can be directly contrasted with this, to suggest the likelihood of failure under a certain condition. For example, excessive strains in ligaments might highlight the potential for overstretching or rupture. Or in bone, changes in stresses and strains leading to bone remodelling can be implemented using bone adaptation laws which can highlight potential changes in bone mass after treatment[15, 72].

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

4 Validation

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

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

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

In those that do carry out *in vivo* or *in vitro* validation, this may be the main objective of the model generation [45], or be possible due to the data collected for input parameters; such as joint angles [9], loading [8] or GRF [58]. The standard for validation is to carry this out for the primary model output [90], such as; plantar pressures [46, 56, 94], peak strains [10, 25] or joint contact pressures [45, 71]. Of the studies which have been validated experimentally they can be split into five main groups:

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

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

Validations carried out against literature will be limited to studies considering the primary output; or may guide users to select a primary output based on what can be validated, rather than the research question. The simulation is also limited to replicating the model setup used in the previously reported study, as model parameters such as loading conditions and material properties should be kept in common for the purpose of validation. After the model is validated, it is then possible to ascertain the sensitivity to any parameter changed. For example, parametric tests could be used to examine the biomechanical effects of different sizes, shapes, and positions of osteochondral defects of the talus, which may assist our surgical decision-making as well as provide guidance on the likely natural history without intervention.

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

5 Clinical relevance

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

The Open Knee model [96] is an example of an orthopaedic FE model that has allowed for clinical translation; this open-source model has allowed investigators outside of the developers to utilise the model in research relating to the biomechanics and tissue structures of the knee joint. This has not only expedited the research capacity in the field but has standardised some aspects of model generation. An equivalent model, to the best of the authors' knowledge, is not available for the whole foot or ankle; therefore, the authors propose a similar strategy is required for this joint to improve translation into clinical practice. This could be either a single patient specific model, or several models to create an in-silico population. Alongside this a general template or checklist on good practice when creating a model of the ankle could improve the clinical usefulness of models; but as evidenced by Viceconti’s paper, such recommendations are not always followed. Our view is that provision of standard materials that make it easier to create good quality models of the ankle, alongside guidelines, is the way forward.

6 Conclusion

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

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

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References

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

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

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

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

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

Tables

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

	E (MPa)		***v***		**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 Isotropic trabecular bone	E1= E2= 17,500 E3=11,500	450	*v 12=0.31* *v13=0.43*	0.3	[43] G13= 3,500MPa
CT Greyscale Based					[15, 19, 20, 32, 39, 51]
Cortical, cancellous and Interface densities (ρ)	$3790 \cdot \rho^3$				[72]

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

	E (MPa)	***v***	**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, C10 = E/(4(1+n))	[83]
	Mooney-Rivlin: C01=0.41 MPa and C10=4.1 MPa	[60]
	Ogden: μ=2.43, α=12.45, D=0.176	[30]

Table 3 Spring stiffness, k (N/mm) reported in literature where ligaments were modelled as 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

Figures

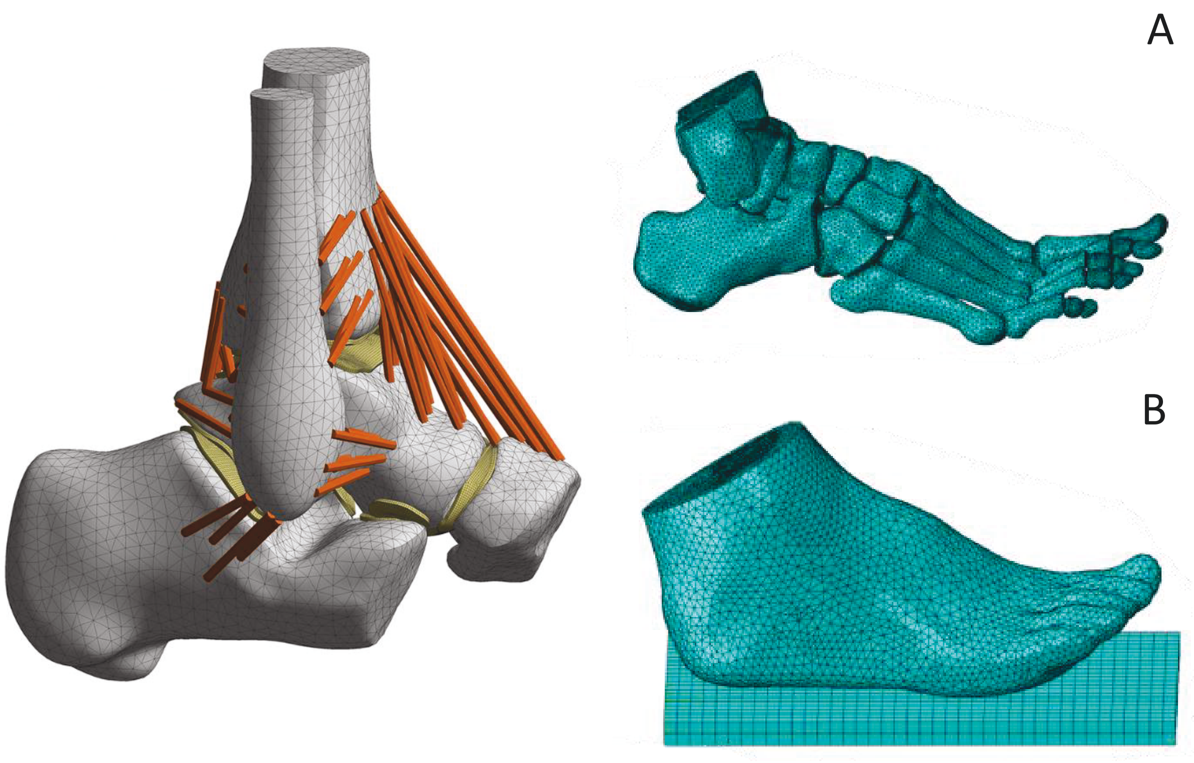

Figure 1 FE model of the bones of the ankle with 3D renderings of 2D truss elements used for ligaments (left) [1] , and of the whole foot (right) adapted from Wang et al. where A) shows the underlying bony structures of the foot and ankle modelled , and B) shows the encapsulated tissues and rigid plate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

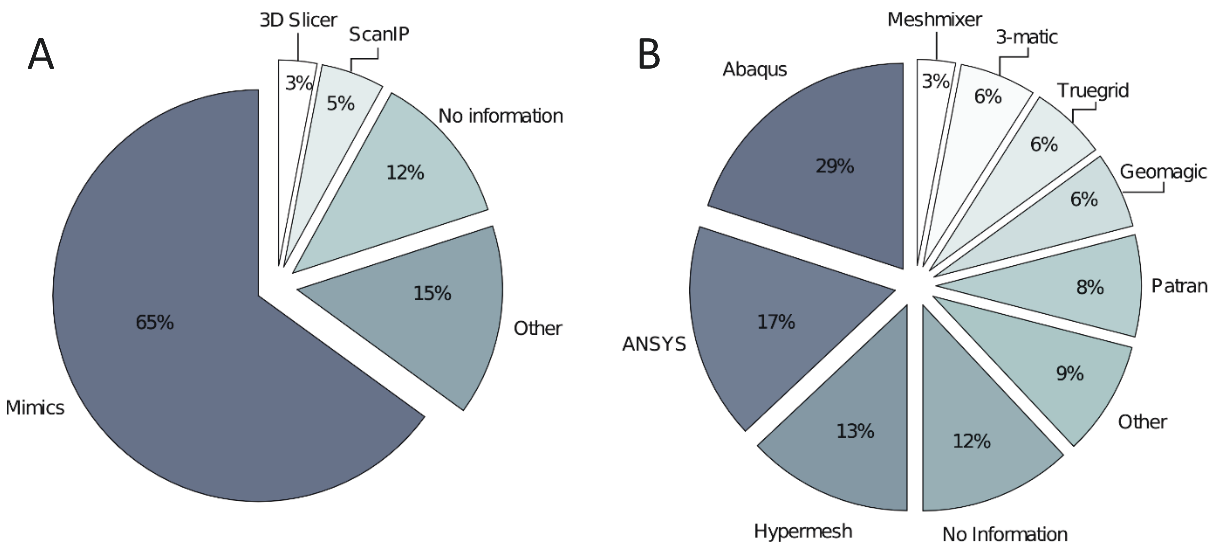

Figure 2 Proportion of software used for A) segmentation and B) meshing based on analysis of foot and ankle FE literature, between 2001 and June 2022

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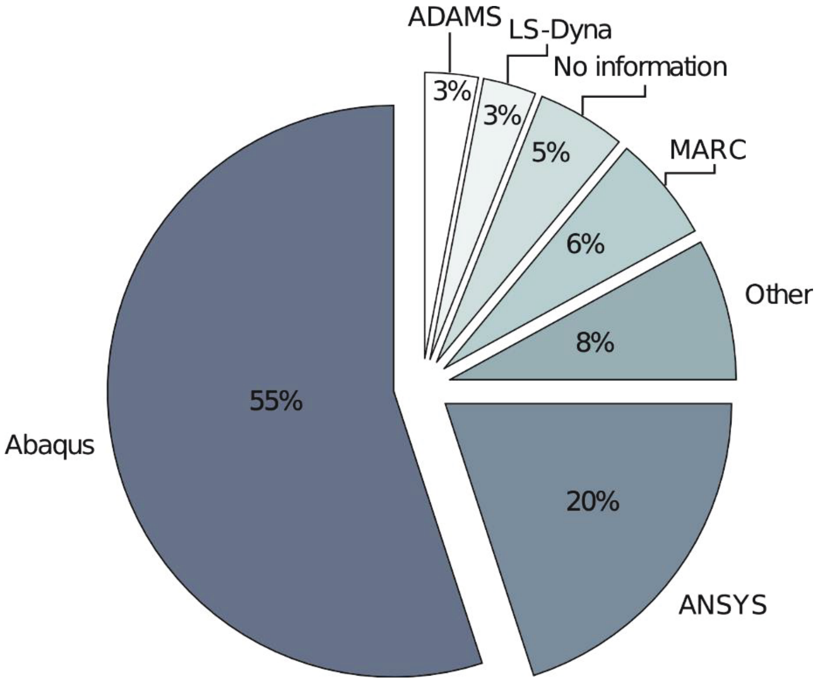
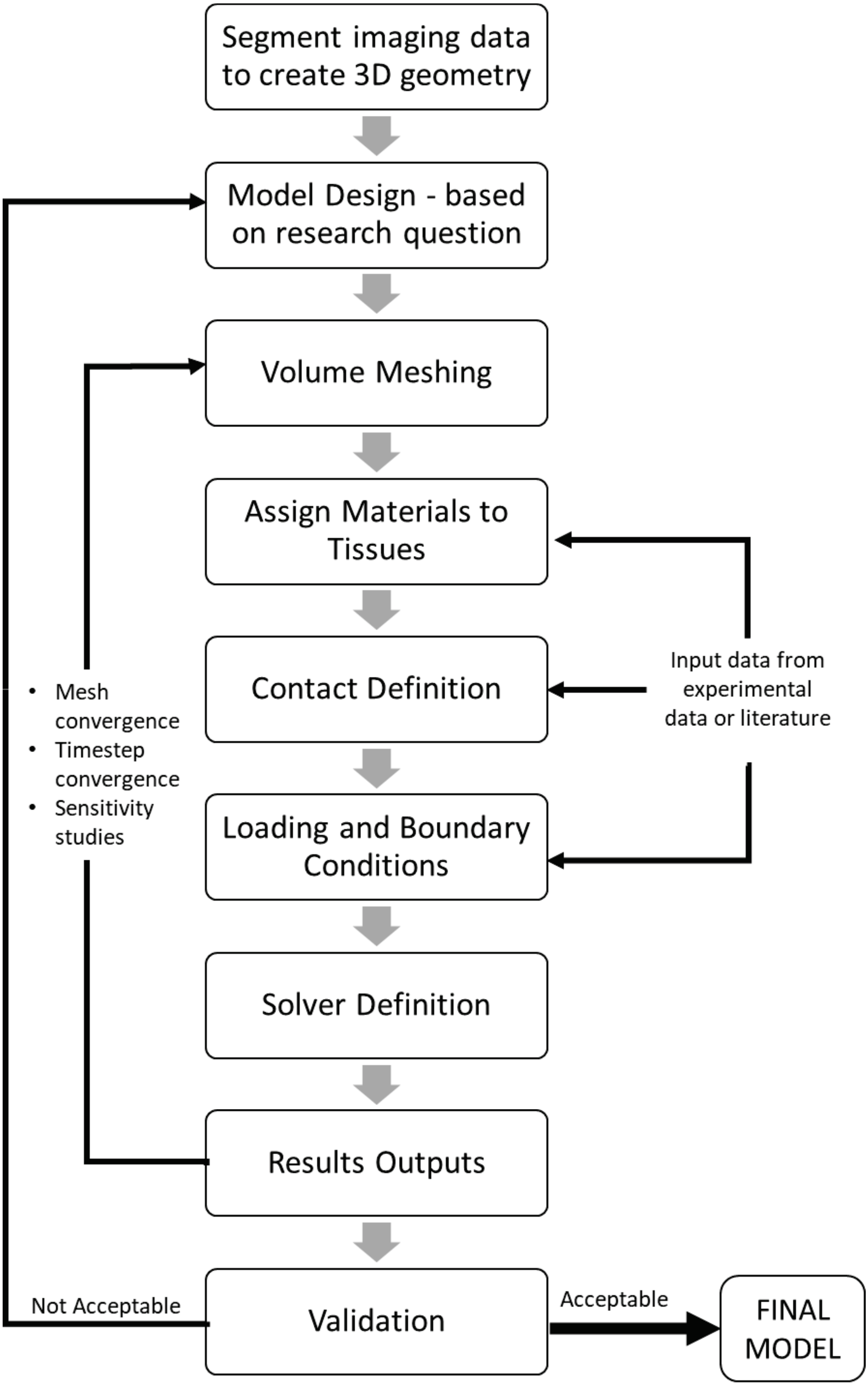


Figure 3 Proportion of finite element software used in foot and ankle literature, between 2001 and June 2022



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

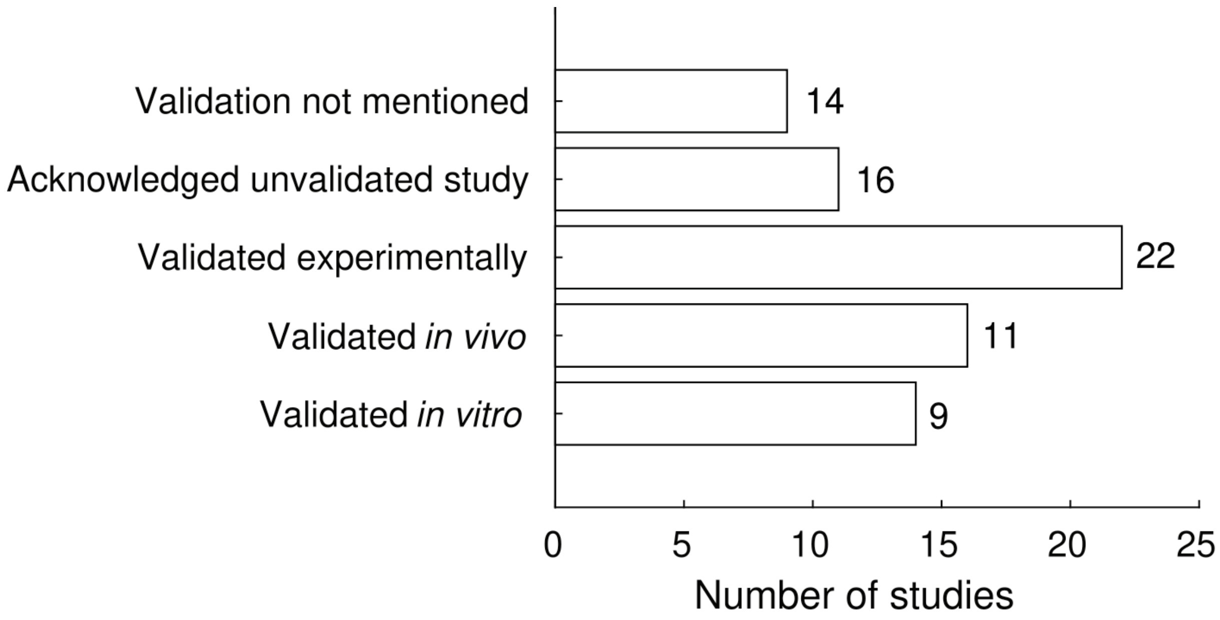


Figure 5 Number of studies in literature reporting methods of validating foot and ankle models