Finite Element Modelling in Bones: A Review

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ABSTRACT

This chapter explores the utilisation of Finite Element Method (FEM) modelling in the field of bone biomechanics. It emphasises the crucial significance of bone biomechanics within the field of medical research for therapeutic applications. The versatility of finite element method (FEM) is demonstrated in its application to biomechanical investigations, clinical applications, and disease modelling. Nevertheless, this study also acknowledges and discusses the obstacles and potential avenues for further exploration within the discipline. The discussion revolves around the careful management of model complexity and computing efficiency, the need for accurate material characterization, the importance of interdisciplinary collaboration, and the ethical considerations involved. In its whole, the chapter illustrates the significant role of finite element modelling (FEM) in advancement of bone biomechanics research and its practical applications..

Keywords—FEM, bone biomechanics, numerical methods, UMAT, abaqus, medical research

I. INTRODUCTION

The term biomechanics involves application of laws of mechanics on the biological systems to understand the motion of the body. Unlike other bodies, biological systems are continuously evolving which often gives rise to problems in applying the general mechanical principles to such living systems [1]. Bone Biomechanics emcompasses developing a deeper understanding of the adaptation of the bone in response to mechanical forces for fracture repair or bone growth [2]. It helps in understanding the quality and fragility of bone which play a significant role in various disease progression such as osteoporosis thereby, playing a crucial role in medical research and engineering to curate practical solutions for various diseases [3], [4]. Since 1972, finite element modelling (FEM) has been used to understand the mechanical behaviour of the skeletal system. It provides a way to represent the entire irregular bone structure and analyze the complete stress field [5]. FEM aids in understanding the fracture and disease progression in the bone as a response to mechanical stimuli to develop treatment or rehabilitation modalities for the patient [6].

This chapter focuses on the biomechanical properties of bone and their utilization as an input for developing a finite element analysis (FEA) model. It aims to establish a framework for model creation and discuss the applications of such models in the field of medical research along with the challenges of utilizing FEA.

II. BONE STRUCTURE AND PROPERTIES

The structural framework of a human adult comprises 206 bones containing majority calcium in the body (99%) [7]. The bone architecture includes a hierarchical organization of differently arranged materials which orchestrate to perform mechanical and biochemical functions. The microscopic bone structure is composed of mineralized collagen fibres which are arranged in sheets to form lamellae. These lamellae are concentrically arranged around the Haversian canal to form osteon. However, in some bones, an absence of osteons is observed with a tangential lamellae arrangement forming a stack of layers with woven bone forming the lamellar bone while the 50-300μm rod like interconnecting trabeculae make up the cancellous bone [8].

A. Cortical Bone

It is a transversely isotropic material with a 5% - 15% to porosity whereby changes in porosity are responsible for 75% strength variability of the bone. Loading rate plays a significant role in the microcrack formation in the cortical bone, and its brittleness increases at high strain rates, with more brittleness in the transverse direction than longitudinal. Tension ends with an abrupt fracture strain of less than 3%. It exhibits a bilinear stress-strain response during tension in contrast to compressive longitudinal loading, during which simultaneous hardening and softening occur before failure at 1.5% strain showcasing the highest strength during longitudinal compressive loading.

B. Cancellous Bone

It is also known as trabecular bone and posses An anisotropic material with high porosity. Like cortical bone, it has the highest strength in compression and the lowest in shear. The spongy bone yields approximately 0.7% during compression while maintaining its load-bearing capacity at 50% compressive strains. Compressing the cancellous bone beyond yield (up to 5%) results in loss of stiffness and strength, depicting the occurrence of fractures due to cumulative, permanent deformations [9].

Osseointegration (OI) plays a vital role in achievement of such success rates. It refers to mechanical and biological anchorage resulting in a structural and functional connection between the implant surface and the living bone with no progressive micromotion [10], [11], [12]. A continuous disturbance in the equilibrium between bone resorption and apposition could result in loss of bone-to-implant contact (BIC), compromising osseointegration (Bosshardt et al., 2016). Unsuccessful healing can be caused by micromotion exceeding 150μm resulting in poor OI, causing compromised primary and secondary stability [13].
Bone healing is comprised of three partially overlapping steps:

1. Inflammation Phase
2. Repair Phase
3. Remodeling Phase

The inflammation phase marks the beginning of the healing process, which lasts up to seven days. It involves the release of pro-inflammatory factors such as cytokines, growth factors, and chemokines resulting in the release of polymorphonuclear neutrophils (PMNs) as a response to tissue injury byproducts [14]. This phase is also known as the hematoma phase due to hematoma formation, which is critical for the successful resolution of fractures [15] Post hematoma resolution, a cartilaginous callus formation occurs (endochondral ossification), ending with the formation of a hard callus osseous tissue peripheral to the soft callus region [15], [16]. Lastly, mechanobiological regulations result in bone remodeling, a dynamic physiological process involving bone apposition and resorption [15], [17]. It begins with osteoclast activation and bone resorption by them, followed by a deposition of a new organic bone matrix by osteoblasts and its subsequent mineralization [18]. Various mechanical forces drive the remodeling process. These mechanical forces act as principal signals, which are transduced into biochemical signals via mechanosensitive signaling pathways and promote bone adaptation to the environment guiding bone healing and growth [19],[20].

The surface topography and design characteristics guide the implant-bone response along with various other factors such as loading parameters, material properties, quality and quantity of the surrounding bone, and quantity and quality of the BIC, specifically during the bone remodeling phase [21], [22] Siegele et al. showcased significant differences in stress dissipation in the bone when subjected to vertical loads, whereby implants with a small radius of curvature or geometric discontinuities exhibited more significant stresses. They also demonstrated maximum stress concentration in the crestal region for screw implants and in the region of the soft tissue layer and direct-implant bone contact during lateral loading. Another study by Holmgren et al. suggested the application of oblique loads for FE analysis, stating that they mimic realistic occlusion directions generating the highest localized stresses in the cortical bone [23].

III. Finite Element Method

The finite element analysis (FEA) is a propitious tool for non-invasive analysis and prediction of complex in-vivo biomechanical processes [24] and evaluation of mechanical forces applied to biological systems. It is a numerical method for analyzing stress, strain, and damage/deformation. FEA involves the discretization of the object in the form of finite elements connected through nodes. The type and arrangement influence the evaluation, and the total number of elements, producing physically repeatable results. FEA provides deeper

insight into the different stress patterns in the implant and the peri-implant bone, providing a biomimetic approach to improve the implant design [25], [26]. FEM involves meshing of the part using softwares such as Abaqus, Ansys or Hypermesh etc for the object discretization. It is immensely important to maintain mesh quality and obtain an aspect ratio of less than one for each element since it is a key determinant of the effectiveness and accuracy of the developed model and simulations [27]. This is especially pertinent to maintain in case of studies on bone due to their inhomogeneous structures whereby, sometimes surprisingly large deformations might occur giving rise to inaccurate data [28]. Two types of nodal elements namely hexahedral and tetrahedral are used in meshing of skeletal parts.

IV. Model Creation

A 3D part of the skeletal system is generated from the computed tomography (CT) scan or three-dimensional (3D) scan post extraction of the external curvature from obtained slices using an algorithm for border tracing. The extracted geometry is meshed using softwares with an automatic meshing routine such as Hypermesh. A recent development in automatic mesh generation (AMG) utilising voxel meshes instead of hexahedral or tetrahedral meshes as it allows the user to bypass the geometry extraction procedure [29]. Moreover, a study conducted by Austman et al. demonstrated comparatively better accuracy when a second-order tetrahedral mesh was used instead of triangular surface meshes [30]. Kluess et al reported use of hexahedral mesh for better accuracy and faster computational time in case of contact definition problems [31]. It is pertinent to perform a mesh sensitivity analysis for model optimization after the assignment of the relevant material properties and interactions. Bone is an anisotropic material but has been modelled as an isotropic or transversely isotropic material as well. For instance, a study carried out on the femur bone of humans assumed isotropic material properties for bone due to the tedious process of assigning the entire bone anisotropic properties [32]. While developing a model for simulation of the human body parts, it is necessary to take into consideration the various human factors which vary in a population and individual. Hence, equation 1 describes the relationship between the Young's modulus, strain rate and bone density to simulate bones of different quality [33].

$$E = 3790\,\rho^{3}\dot{\varepsilon}^{0.06} \tag{1}$$

A simulation expert often needs to develop a simplistic model to minimize the model running times but also take into account that the developed model is a close approximation of the real-world scenario. This involves applying boundary conditions in consideration with the assumptions made for the model. Nonetheless, it is to be noted that the forces, pressure or applied displacements should mimic the physiological conditions [34].
Recent developments in the field of medical devices have prompted varied studies to simulate the dynamic modelling and remodelling process of bone using VUMAT or UMAT subroutines. Humbli and Thurner expressed the bone remodelling process using a Continuum Damage Mechanics (CDM) approach demonstrated in equation 2.

$$\sigma_{ij} = (1 - D^{fat})a_{ijkl}\varepsilon_{kl}\ , \tag{2}$$

where,

$$D^{fat} = 1 - [1 - (\frac{N}{N_f})^{\frac{1}{1-\gamma}}]^{\frac{1}{1+\beta}} \tag{3}$$

Here, γ and β are material parameters and N_f is the cycle of failure for compressive or tensile loading. Further, the Young's modulus for the mineralized or damaged bone can be calculated as,

$$E = C\,(1 - D^{fat})\rho^{p}\alpha^{q} \tag{4}$$

Where, ρ is the calculated density using Komarova's model and α is the ash function denoting mineralization of bone. C is experimentally derived and according to Hernandez et al. q = 2.74 meanwhile $3 \geq p \geq 2$.
[35]. Weinans et al., describes the rate of change of bone density as a function of mechanical stimulus S and the threshold stimulus k as described in equation 5.

$$\frac{d\rho}{dt} = B(S - k) \qquad \rho_{cb} \geq \rho > 0\,, \tag{5}$$

where, B represents a constant and ρ_{cb} the maximum density of bone. Furthermore, if the threshold value (k) is greater than the loading, resorption will occur; conversely, if the loading is greater than 'k', bone growth will take place. However, Weinan's model fails to address the overload resorption occurring due to large loadings.

Validation and verification of the model through experimentation is crucial for model calibration. Statistical analysis such as analysis of variance (ANOVA) should be performed to validate the FEA model whereby, the standard deviation (SD) between the results from the simulation analysis and experimentally analyzed average data shouldn't exceed 1 with an $\alpha > 0.05$ to reject the null's hypothesis in ANOVA [36], [37].

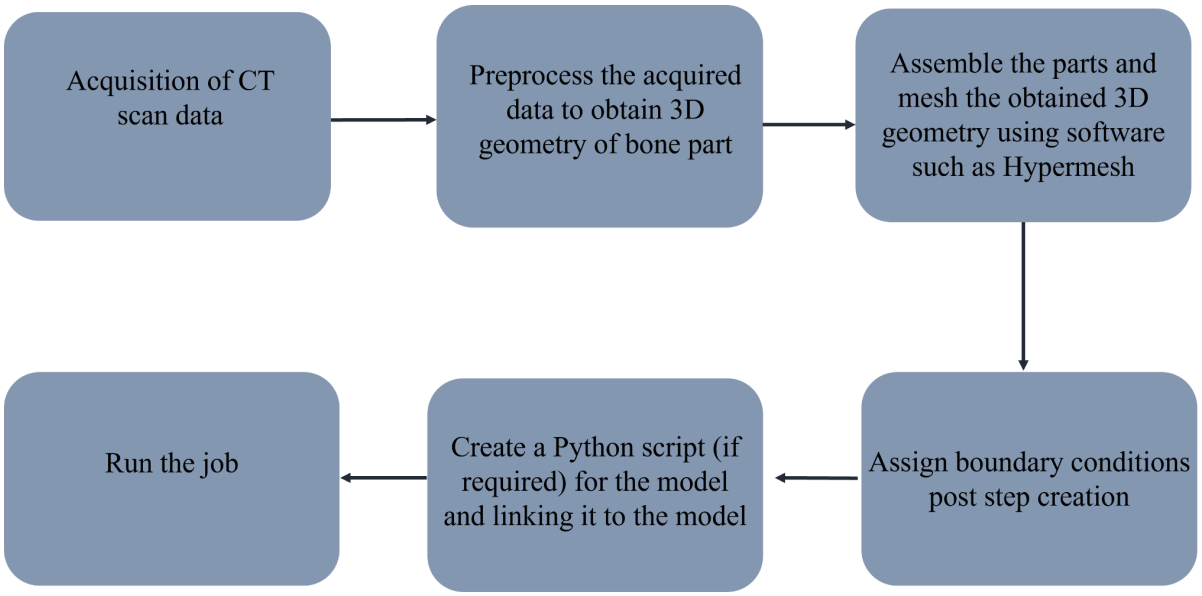

Figure 1: Schematic representation of the process involved in model creation

V. Applications

Advancements in healthcare have led to an increase in awareness about movement disorders which often result in death or fracture due to inappropriate gait. Gait disorders can stem from varied clinical disorders such as osteoarthritis, parkinsons, stroke or due to old age [38]. FEA is used to predict loading conditions while standing or moving.in the parts of the foot which is useful for improvement of gait and development of rehabilitation devices [39]. Additionally industries and research organizations perform FEA for developing and optimizing implant designs for various parts of the body. For instance Ferdinand et al. developed an anterior-cruciate ligament (ACL) compatible design for knee implant using a Mooney-Rivlin model with transverse isotropy and X,Y translation and Z rotation connectors for the knee joint motion.. This led to the development of an implant design for bi-cruciate knee arthroplasty [39]. FEA has also been used for management of fracture fixation sometimes utilizing dynamic solver with complex contacts and subroutines predicting fracture healing or progression of diseases such as osteoporosis resulting in bone porosity change and efficiency of the implant for fracture fixation [40], [41].

VI. Challenges and Future Directions

Currently, there have been very few studies incorporating the piezoelectricity induced mechanoadaptation in bone with Bansod et al. performing the first study utilising a python script for femur bone remodelling with piezoelectric effect. However, hexahedral meshing is not embedded in the software used for this study so, a ten-noded tetrahedral mesh was utilised for the domains and the study was performed for 100 days with a step time interval (t) of 0.1 day. However, experimental validation of the simulated data in case of human bones is not always possible due to the unavailability of patient specific human models [42] resulting in hampered accuracy [43]. Additionally, bone remodelling is a complex process involving complex theoretical equations which are sometimes oversimplified for simulations thereby, not remaining relevant in the clinical practice. Prior utilization of FEA in dentistry involved simplifying parameters such as boundary conditions, material properties, geometry, and load [44], [45]. In future, these models can be optimized to improve the validation of the bony

structures and studying implant fixation [46]. Furthermore, characterization of bone material properties is a tedious process due to the bone microstructure for instance, orientation of stress is largely dependent on the the trabeculae orientation in bone which is changing and challenging to incorporate in a model [47] thereby, there is a requirement for studies with improved models incorporating algorithms to closely model the complexities of bone. Brekelmans et al. describe difficulties in mimicking the real world boundary conditions associated with bone biomechanics and identifying the appropriate data to be incorporated into the model; however, they performed studies on a two-dimensional (2D) model instead of a 3D model of the skeletal part [48]. A recent study carried out by Cronin et al. described a CDM model with stress triaxiality, previously ignored in myriad studies due to lack of experiments demonstrating relationship between stress triaxiality. They developed an orthotropic model with the primary material axis in the direction of osteons and the remaining axis as radial and circumferential. Nonetheless, experimental verification to understand the reliability of such models still needs to be validated [49], [50].
Navigating these challenges and pursuing the future directions can lead to more accurate, reliable, and clinically applicable bone FEM models, ultimately advancing our understanding of bone biomechanics and its implications for healthcare and medical interventions.

VII. Case Study

To understand the behaviour of tibia subjected to torque, Carbonell et al. performed a study to demonstrate patient-specific bone remodelling whereby CT images of the bone were obtained in digital imaging and communication in medicine (DICOM) format and reconstructed using MIMICS software. Post, these mechanical properties were ascribed to the obtained 3D geometry after importing the file in Abaqus 6.13 (SIMULIA, Dassaults Systems, RI, USA). The complex geometry was meshed using C3D4 tetrahedral mesh creating an unstructured mesh with an average shape factor of 0.656. Assuming a bone remodelling equilibrium, the used stress was defined using equation 6, whereby U and E describe the strain energy and elastic modulus respectively for continuum stress.

$$\sigma_c = \sqrt{2EU} \tag{6}$$

Furthermore, the bone density change was described as shown in equation 7. ρ_t represents remodelled bone tissue density and k is the actively remodelling local area fraction. Martin quantified the actively available surface area bone volume for remodelling using S_V.

$$\Delta\rho = k \cdot S_V \cdot \rho_t \tag{7}$$

The algorithm was further written using UMAT subroutine in FORTRAN to simulate bone remodelling process after application of boundary conditions on the model. During analysis, each iteration represented a day with a bone density of .5g/cm^3 during the first step and the job completion marked the evident visual of the newly remodelled bone structure with cortical bone density of tibia approximately 1.92g/cm^3 with the maximum density located near to the distal epiphysis[51], [52].

VIII. Conclusion

In the field of bone biomechanics, Finite Element Method modelling serves as a robust conduit that connects the disciplines of engineering, medicine, and biology. This chapter explored the intricate subject matter of FEM of bones, revealing its significant uses and associated difficulties, while providing insights into the future directions that this evolving area is poised to adopt. The field of bone biomechanics plays a crucial role in both medical research and engineering, highlighting the need of comprehending the behaviour of bones. It has emerged as a crucial tool in understanding the complex dynamics of bone reactions to mechanical stimuli, fractures, and illnesses. The chapter commences by delving into an examination of the hierarchical arrangement of bone structure, meticulously analysing the characteristics of cortical and cancellous bone. This foundation functions as a starting point to explore the core aspects of FEM, its fundamental principles, and its use in bone modelling. The versatility of FEM becomes evident as the chapter explores model creation. The process involves a detailed description of the systematic procedures involved in acquiring imaging data, generating a mesh, assigning material properties, and specifying boundary conditions. The careful procedure establishes the foundation for simulations that accurately replicate physiological conditions in the real world, shedding light on the complex stress patterns present in bones. It exhibits wide-ranging implications within the fields of biomechanics, clinical practice, and the development of medical implants. The utilisation of FEM in several

areas, including as the analysis of gait abnormalities and the optimisation of implant designs, has significantly contributed to the progression of knowledge and technologies aimed at improving human well-being. The chapter's focus on fracture fixation and disease progression models underscores its importance in the area of clinical practice. However, this chapter does not avoid discussing issues and suggesting potential future directions. The aforementioned statement explores the delicate balance between computing efficiency and model complexity, emphasising the continuous pursuit of precise material characterization, and promoting the importance of interdisciplinary cooperation. The incorporation of biological elements, personalised patient models, and sophisticated material representations is anticipated to facilitate the development of more authentic simulations that accurately replicate the dynamic characteristics of bones.

As we draw this chapter to a close, it becomes apparent that FEM modelling is positioned at the forefront of a paradigm-shifting period in the field of bone biomechanics. This robust methodology not only facilitates the interpretation of the mechanical complexities of bones but also drives us towards personalised medical therapies for patients, advanced disease modelling, and novel implant designs. The expedition has been replete with profound observations, difficulties, and opportunities, resonating with the cooperative ethos of the scientific and medical communities in moulding a healthier tomorrow.

A. Abbreviations

ACL	Anterior Cruciate Ligament
AMG	Automatic Mesh Generation
ANOVA	Analysis of Variance
CDM	Continuum Damage Mechanics
CT	Computed Tomography
DICOM	Digital Imaging and Communications in Medicine
FEA	Finite Element Analysis
FEM	Finite Element Method
SD	Standard Deviation
3D	Three-Dimensional
2D	Two-Dimensional

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