Study of the interaction of various forces with failure crack propagation during typical orthopedic bone fracture

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Abstract

Bone is a naturally occurring composite substance with remarkable strength and toughness due to its internal hierarchical structure. Despite its unique characteristics, bone also has a semi-brittle behavior and can still experience fractures caused by accidents and surgeries. This paper will delve into the study of failure mechanisms in bone influenced by standard procedures in orthopedic surgeries and accidents. The understanding of bone fracture patterns caused by osteotomy (bone cutting) and bone damage during accidents is critical to properly treating patients during postoperative recovery. Researchers have been involved in understanding bone failure for decades through in-house lab experiments and simulation. In this work, we will summarize a branch of literature survey on different failure understanding of the bone that has already been reported. Our primary focus will center on the damage to healthy bones due to various loading conditions, particularly affecting the bones in the legs. Finally, we will attempt to describe experimental techniques and numerical models for bone failure using standard and controlled loading conditions, which replicate typical bone damage during typical orthopedic fracture.

Introduction

Bones are one of the most critical parts of the human body's skeletal system, fulfilling various necessary functions. They provide structural support to bear our body weight, protect our organs, and enable the fluidity and efficiency of a person's everyday movements. On average, an adult possesses 206 bones, each of which has its distinct size, form, and density, depending on one's physical condition and features. Bones take up about 15-20 % of the total body weight, and they have unique structural characteristics that allow them to adapt to any changes in the external environment. The bone's structure adapts to the magnitude, frequency, and distribution of loads and the type of deformation that exposes it. [1]

Throughout life, bones can grow, repair, and change their structure, size, and strength qualities to comply with mechanical exposure around and within the body. However, this ability can encounter certain restrictions due to the semi-brittle behavior of the bone. Moreover, the fragility of our bone structure is directly related to the risk of injury and mortality rate. Bone fragility, which is frequently connected with aging, medical disorders such as osteoporosis, and lifestyle choices, contributes to the likelihood of fractures and other orthopedic complications. Fractures resulting from falling, traumatic experiences, and accidents can have profound implications, as they are accompanied by temporary suspension from work or permanent disability. Considering the wide variety of bone studies available, we will channel our focus to the study of cortical bones since they are prone to injuries and bone deterioration side effects. [2, 3]

Orthopedic Cortical bone (Leg Bone) is a composite material naturally occurring. It primarily comprises minerals (hydroxyapatite) and organic components (bone cells and collagen). These components are arranged in a complex, hierarchical, anisotropic microstructure. The osteons, which are cylindrical structures made up of 10–30 concentric lamellae with a thickness of 3–7 µm each, are microstructural characteristics. The outer limit of these lamellae structures, which have an inner diameter of 30 µm to 50 µm, divides the osteons from the cement line, an interstitial matrix. The interstitial bone, an incoherent tissue that makes up the bone matrix, fills the area between the osteons. Several microcracks are observed inside osteons. They usually grow and combine inside the interstitial bone, extending to the cement lines. Cement lines and cracks interacting can cause Osteonal delamination, resulting in various fracture mechanisms. Since cortical bone uses multiple methods to raise the energy needed to fracture, the start of cracks is much less significant than their propagation. [4, 5]

During the life cycles, these micro-cracks in Leg Bone undergo several levels of static and dynamic loading conditions. Hence, it is crucial to understand the behavior of cracks under different circumstances to understand and predict the behavior of their propagation nature and following bone fractures. Therefore, a complete understanding of fracture's nature and its behavior's mechanics fracture nature and its behavior under physical load is necessary. In this paper, we will review several articles on different fracture mechanics understanding and other computational and experimental techniques to study bone failure under several types of dynamic loading.

Stress associated with the bone

Everyday activities exert various pressures on our femur bones. Understanding the stresses that affect the femur bone is important because they are closely linked to the risk of mechanical fractures. Walking, running, jumping, and lifting objects all subject our femur bones to stress. Fractures can happen when the femur is subjected to powerful or unexpected pressures, like those from falls or impacts. For instance, if a fall or collision applies too much force to the femur, it can fracture because the compressive, tensile, or bending forces exceed its strength. Similarly, twisting the femur too much can also cause fractures. By understanding the different stresses, the femur bone faces, we can better grasp what makes it vulnerable to mechanical fractures. [1, 3]

Type of Injuries

There are different types of femur injuries, and depending on the location of the injury and how serious it is, every type of femur injury requires a different treatment approach. Fractures and dislocations are common types of femur injuries. Penetration injuries are also types of femur injuries, which include gunshot and stab wounds. Loading forces cause fracture injuries that include transverse fractures and spiral fractures. Transverse fracture might develop from compressive forces, while spiral fracture might develop from torsional or twisting forces. Avulsion fractures and greenstick fractures are types of fractures that involve stress concentration. Localized failure could arise from this type of stress concentration. Fracture mechanics concepts make it easier to measure stress concentrations and predict the fracture initiation. [3]

Theory of Bone Failure

Bones are matrix-like structures due to multi-dimensional hierarchical structure; one might gain an insight into the complexity of bone fracture. Bone has different biological parts that can act as chemical barriers (like cell membranes), ion reservoirs (like bone), catalytic properties (like enzymes), convert chemical energy into kinetic energy (like muscle), etc. At the macroscopic structural level, bones can take on various morphologies based on their role. In our extremities, long bones like the tibia and femur offer support against buckling and bending. In other situations, the imposed load is primarily compressive, such as the vertebra or the head of the femur. In these situations, trabecular or cancellous bone—a "spongy" substance—may fill the bone shell. Cortical bone refers to the walls encircle trabecular bone regions and the walls of long, tube-like bones. The cortical bone shell, located on the outside of every bone, can vary in thickness, as shown in Figure 1, from a few tenths of a millimeter in vertebrates to several millimeters or even centimeters in long bones' midshafts. [6]

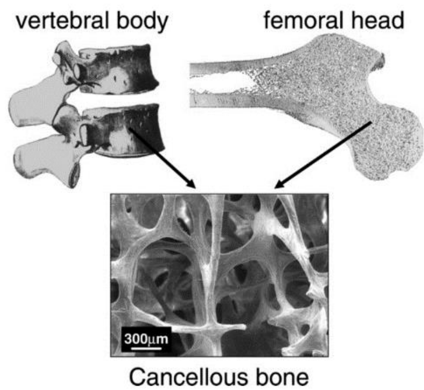

Fig 1. The rib cage is a spongy substance that fills some bones (or portions of bones), such as the femoral head and vertebra. The struts, or trabeculae, are a few hundred micrometers thick[6].

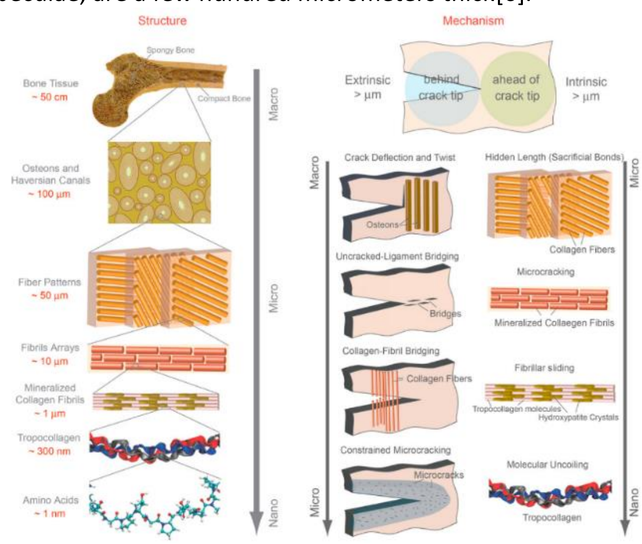

Figure 2: Different layers of bones and their corresponding crack behavior[4]

In figure 2, Launey et al. Described intermolecular behavior and characteristics of bones and their corresponding molecular interaction. The slightest degree of (intrinsic) toughening, or plasticity, is produced at the scale of tropocollagen molecules and mineralized collagen fibrils by molecular uncoiling and intermolecular sliding of molecules. Microcracking and fibrillar sliding act as plasticity mechanisms at the finer scale level. It improves the intrinsic toughness of fibril arrays. Energy dissipation at micrometer scales is enhanced by the breakdown of sacrificial connections at the interfaces of fibril arrays and crack bridging by collagen fibrils. Significant fracture deflection and crack bridging by uncracked ligaments are the primary extrinsic causes of toughening at the largest length scales in the millimeter range. [3, 5]

Nevertheless, most measurements have only included tensile (mode I) loading with linear-elastic fracture mechanics, despite bones always fracturing under complex loading conditions. However, as recent research has demonstrated, the fracture toughness in shear is much lower than in tension. This is pertinent because, in a physiological sense, bones are rarely loaded uniaxially; instead, they are frequently subjected to extremely mixed-mode combinations of tension, compression, and shear, contingent upon the nature of the applied forces they encounter, the geometry of the bone, and, most importantly, the orientation of the crack in relation to the applied loads. Consequently, a mixed-mode fracture toughness includes contributions from mode I (tensile), mode II (shear), and/or mode III (anti-plane shear) crack displacements. Figure 3 shows the observed lower critical strain energy release rate along the transverse direction compared to the longitudinal direction. For a healthy human, the leg bone suffers most loads along mode-I axial direction throughout his life. [4]

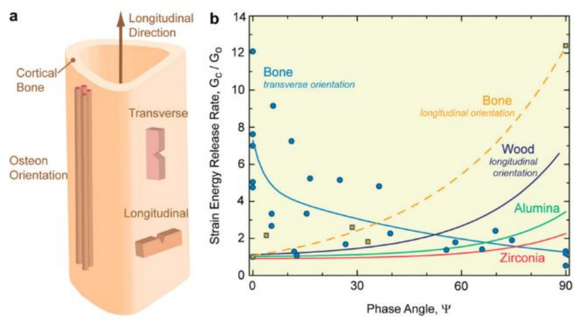

Figure 3: Strain energy release rate for bones (Femur)[4]

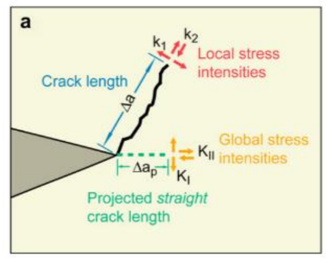

Fig. 3. (a) The local stress intensities (k_1 and k_2) that emerge at the crack tip when the crack deflects regarding its initial plane can be seen under a magnified image of the crack tip. [7]

Strain Energy Release Rate under mix model loading,

$$G = \frac{{K_I}^2}{E'} + \frac{{K_{II}}^2}{E'} + \frac{{K_{III}}^2}{2\mu}$$

Where, μ=shear modulus, E'= E in plane stress or E/ (1- μ^2) plane strain case, and

Phase angle, $tan(\varphi) = \frac{K_I}{K_{II}}$ [2, 7]

Recent studies on human cortical bone have used several methods to assess strain-energy release rate critical values. However, because of their intricate nature, single-value LEFM toughness parameters that are based on crack initiation, like K_{Ic} and G_c, are unable to adequately capture or even accurately depict the numerous extrinsic and intrinsic length-scale toughening mechanisms at work in cortical bone, where most of the toughness is generated during crack growth rather than crack initiation. Hence, different experimental and numerical techniques have been used by other research groups to understand and characterize the failure behavior in human bones. [4, 8]

Experimental Techniques to Analyze Bone Failure

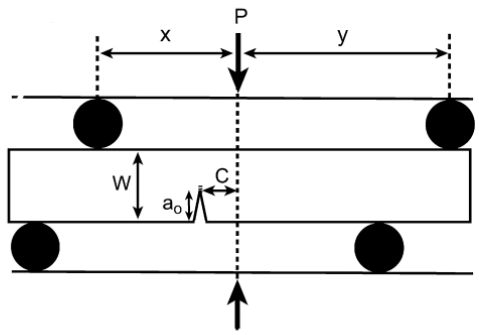

Figure 4: Experimental setup for 4-point bending test. [4]

Dr. Robert's group used a 4-point bending setup per Figure 4 to determine the Resistance curve (R-Curve) and critical strain energy release rate under mix-mode loading conditions for a notched femur bone. Since stable (subcritical) cracking occurs before outright fracture; they conceptualized that rising resistance-curve (R-curve) behavior—where the fracture resistance increases with crack extension—better captures the fracture toughness than outright fracture. They deployed load profiles to generate pure mode-I loading (ψ=0), mixed-mode (mode I + II), and pure mode II (ψ=90) circumstances were measured using the asymmetric four-point bend geometry. They deployed the LEFM and K-dominance field due to a smaller plastic zone (one order smaller) than the characteristic length of the specimen. [4]

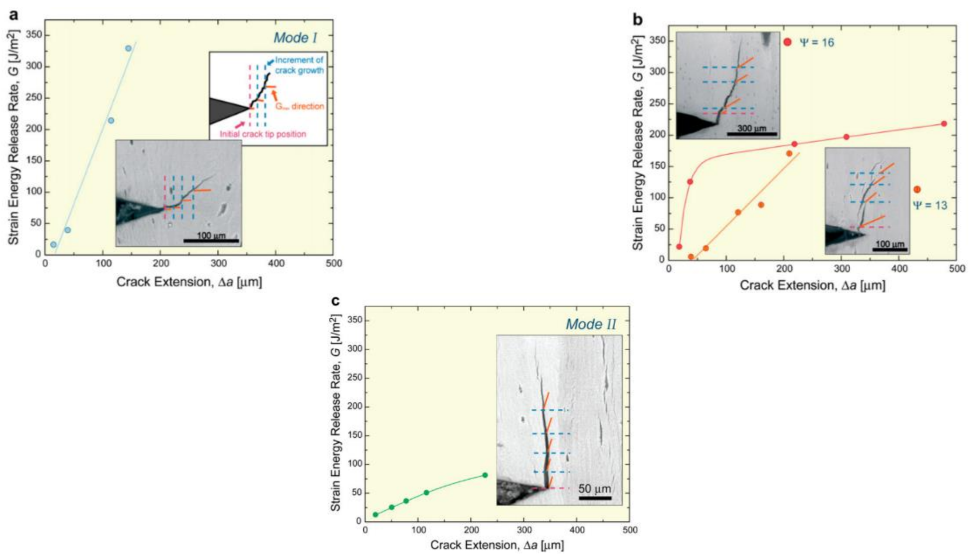

Fig. 5. 2D R-curves showing an image of the crack extension and the various failure modes of the lateral-oriented human cortical bone (femur) samples (a) mode I, (b) mixed-mode, and (c) mode II. The dotted red line indicates the original crack position, the dotted blue lines mark the crack growth increments, and the orange line indicates the G_{max} direction at that place. When the driving force is perpendicular to the weak microstructural path, higher toughness values are obtained when comparing the G_{max} direction with the actual fracture path. [4]

Ritche's observation shows that bone fractures always follow a mixed-mode pattern. This work's main hypothesis was that the change in cortical bone toughness with applied loading direction can be studied using a fracture path. The energy minimization direction between the preferred mechanical and microstructural paths determines the cracking direction. The preferred mechanical crack path is the path of the maximum driving force, or the direction of maximum G, which varies between 0° and 74° with respect to the original crack plane for modes I and II, respectively. The preferred microstructural crack path is along the cement lines, or along the long axis of the bone (Fig. 3 and 3a), which in the transverse orientation is perpendicular to the original crack plane. [4]

In Figure 5a for the mode-I tension case, the preferred mechanical (G_{max}) path stays orthogonal to the preferred microstructural path along the long axis of the bone. Hence, the transverse toughness of human cortical bone increases with fracture extension causing more driving power to spread. This is demonstrated by the steepest R-curves for any mode-mixity, which are in line with the fracture surfaces' roughness, and it strongly deflected crack courses. But the opposite is observed in mode II (in-plane) shear; the preferred microstructural path and the driving force's direction are almost similar (Fig. 6c). Because the driving force is in line with the desired microstructural direction, the fracture can theoretically follow it. The predominant source of toughening is crack bridging, with linear crack courses and planar fracture surfaces, as microcracks now prefer to form ahead and parallel to the advancing crack.

When a fracture propagates in mode II (in-plane) shear, the preferred microstructural path and the driving force's direction are almost similar (Fig. 7c). Because the driving force is in line with the desired microstructural direction, the fracture can theoretically follow it. Furthermore, in this mode, the R-curves tend to be the shallowest (Fig. 7c). Hence, the bone is less resistant to fracture in shear forces. It appears that cracks under mixed-mode tension and shear loads (Figure 7b) travel in a direction that causes the preferred mechanical and microstructural routes to diverge more and more. R-curves, however, are steeper than in mode II but less so than in mode I because this divergence is always less than in mode I. Hence, nature allows the bone to become more resistant to fracture along the mode-I loading direction. [2, 4]

Computational Techniques

The experimental approach discussed above involves physical experiments, whereas the computational approach deals with computer simulation and data generated through mathematical computations, such as Finite Element Analysis. Dr. Anna's group conducted the first study to simulate complete fracture paths in subject-specific FE femur models. This study demonstrated how discrete damage models can provide a more comprehensive view of fracture risk by taking both bone strength and fracture toughness into account in a subject-specific manner. [9]

The methodology in Anna's paper describes a process for simulating complete fracture paths in subject-specific models by including both the PUFEM framework (Partitioned Unifying Framework for Explicit Dynamics and Mechanics) and the femur finite element (FE) models. This paper used validated femur FE model from a presvious study and combined it with PUFEM framework. [9]

In the PUFEM framework, a crack is embeded in a continuum body using enhanced of freedom to describe discontinuous displacement field equation. This equation is expressed as the sum of compatable and enhanced parts. The displacement field equation leads to two variational statements and facilitates the formulation of consitutive models for the bulk material and the failure zone. The failure zone, described by the cohesive zone model, represents the crack. the PUFEM framework includes a two-step algorithim to track the crack path (crack propagation) in 3D. Modifications to PUFEM framework were conducted, including implementing a heterogeneous material description in order to allow for subject-specific distribution of material properties. This relationship $E(\rho) = 6850\rho^{1.49}$

describes Young's modulus as a function of bone density. It involves converting Hounsfield Units (HUs) in the CT images to Young's moduli and assigning specific values within finite elements. [9]

Experimental date set and full details of the mechanical testing of the bone samples were reported from a previous study. The bone samples used in this study, that describe two male cadaveric femurs (bone #1, bone #2), were CT scanned (Definition AS64, Siemens AG, 0.4×0.4×0.6 mm voxel size) prior to mechanical testing. The bone samples were resected 5.5 cm below the minor trochanter, and the shafts were then embedded in epoxy. After the mechanical testing, the bone samples were loaded to failure in a single-leg-stance configuration at a speed of 15 mm/s. They used a testing machine actuator to acquire the force-displacement data. The fractured samples were scanned again using a cone beam CT (CBCT, Planmed Verity, 0.2×0.2×0.2 mm voxel size). The second round of imaging was performed to visualize the fracture pattern. [9]

Model adaptation of subject-specific FE models from a previous study were made toward the application of PUFEM framework. Bone #1 was used for all steps of model development, including testing the effect of various assumptions. Bone #2 was used for the final approach. After validating both bone samples against the experimental data, the mesh genearation occured, that meshes from previous study were used. The study evaluated the affect of using homogenous and heterogeneous material distributions on crack propagation (Fig. 6). In the homogeneous models (Fig. 6a and b), a Young's modulus of 10 GPa was assigned to all bone elements. In the heterogeneous models (Fig. 2c and b), a Young's modulus was calculated in the relationship discussed above that describes the young's modulus as a function of bone density. A minimum of Young's modulus of 5 GPa was assigned to cortical bone elements at the surface. Viscoelastic properties were accounted in this paper. In the homogenous models, the fracture initiation was allowed either throughout the entire bone (Fig. 6a) or was limited to the femoral neck (Fig. 6b). In the heterogeneous models, the fracture initiation was allowed within the entire bone (Fig. 6c and d). A sensitivity analysis was then performed to explore the influence of a parameter in an equation, resulting in different fracture types. [9]

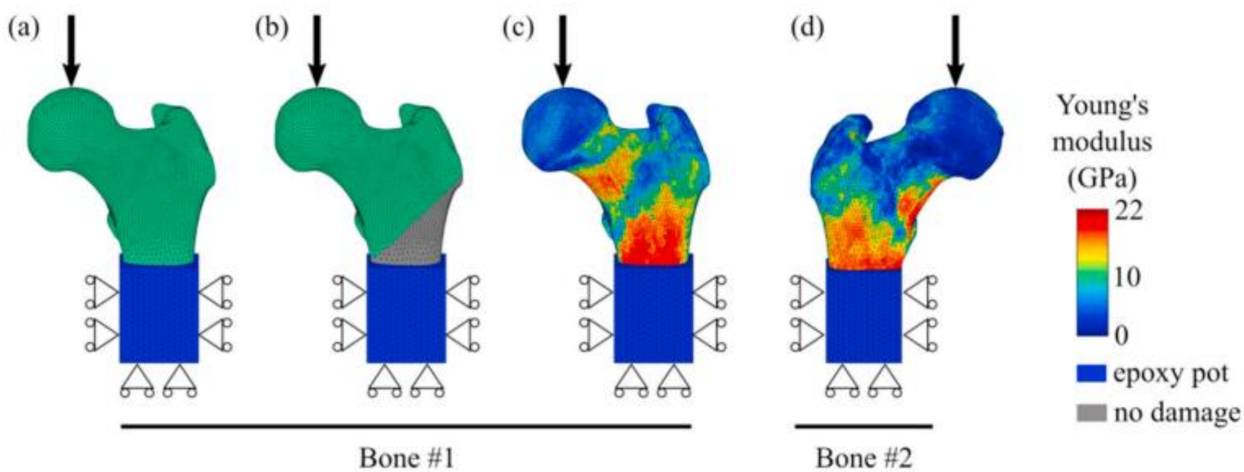

Fig. 6 Material distribution and boundary condition for analyzed FE models. Bone #1 was used for all steps of model development, including testing the effect of various assumptions. Bone #2 was used for the final approach. [9]

After the simulation setup, a solution was introduced; a displacement-controlled arc-length method was used to solve the nonlinear problem in FEAP (Finite Element Analysis Program), with simulations ran until the bone failure condition was met. Then, a sensitivity analysis was conducted to determine the appropriate load-step for each model. [9]

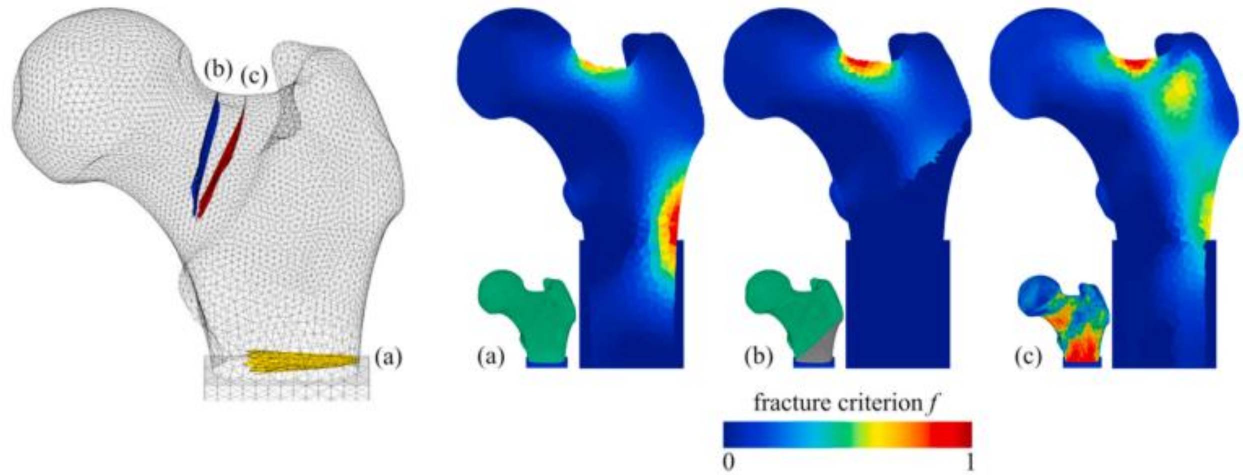

Fig. 7 Predicted crack paths for the three different models representing bone #1. The fracture criterion f, and the figures represent this parameter color-coded at the cross section where the crack has initialized. The sub-figures illustrate the material distributions from Fig. 6. [9]

The paper stated that the density distribution in the proximal femur influenced the predicted fracture patterns. In fig. 7, three different models for bone #1 were analyzed. In the homogeneous models, fracture occurred in the lateral side of the femoral shaft, above the epoxy pot, due to strain concentration from bending deformation (Fig. 7a). Fig. 7b shows that the bone cracked in the middle of the femoral neck, the smallest cross-sectional area. Fig. 7c, heterogeneous model, predicted a neck fracture to initiate approximately 7 mm closer to the trochanter as compared to the homogeneous model. The predicted fracture patterns in the heterogeneous subject-specific FE models (Fig. 4c and d) align with the experimental results provided in the paper. The predicted crack initializations were provided in the paper, and they were approximately 7 and 12 mm from the experimental fracture line in the CBTC images in bone #1 and bone #2, respectively. Other figures in the paper showed the computed force-displacement curves, including peak force values of 13231 N and 7428 N for bone #1 and #2, respectively. The experimental peak forces were 13383 N and 7428 N for bone #1 and bone #2, respectively. The simulation predicted errors of 1% and 5% for the bone strength. [9]

Conclusion

In concluding our comprehensive study on the interaction of various forces with failure crack propagation during typical femur bone fractures, we consolidate our findings from theoretical perspectives and practical applications through experimental and computational techniques. As a composite material, the intricate nature of bone poses unique challenges in understanding its failure mechanisms, especially under the diverse conditions it is subjected to during orthopedic surgeries and accidents.

Theoretically, our exploration into orthopedic cortical bone's hierarchical and microscopic structure has shed light on its complex behavior under stress. The primary composition of minerals and organic components contributes significantly to the bone's unique strength and toughness, yet its semi-brittle nature makes it susceptible to fractures. Theories regarding bone's adaptability to mechanical exposure highlight its dynamic ability to remodel itself to improve the intrinsic toughness of the fibril arrays and the primary extrinsic causes of toughening. We also learned that bone fractures are subject to extreme mixed-mode fracture toughness, and single-value LEFM toughness parameters are unable to capture the fracture mechanism at the cortical bone

From the experimental study we explored, some scholars used sophisticated techniques, such as the 4-point bending setup, to replicate mixed-mode loading conditions and determine the relationship between cortical bone toughness and its fracture path. Through the controlled experiments, we could observe the toughness values increase as the driving force is perpendicular to the weak microstructural path. The preferred mechanical crack path and the microstructural crack path were found to be key determinants of the cracking direction. The Mode-I tension cases demonstrated the steepest R-curves for any mixed-mode condition, making the bone more resistant to fracture along the mode-I loading direction. In contrast, mode-II shear cases demonstrated shallower R-curves with linear crack courses and planar fracture surfaces, indicating less resistance to fracture in shear forces. Mixed-mode tension and shear loads caused the preferred mechanical and microstructural routes to diverge, making the bone less resistant to fracture. It is observed from the literatures that for femur i.e., legs bone, the shear loading direction in mode II have no significant toughening effect causing rapid growth of crack with no significant increase in driving force. But in legs bone, load is applied axially in mode-I direction. Hence, natural self-toughening effect by osteons fiber causes restricted crack growth in mode-I loading condition. Thus, this is a self-limiting phenomenon.

On the computational technique, scholars adopted numerical models to simulate bone failure under various conditions using finite element analysis tools such as XFEM and PUFEM framework. The simulated models accurately depict both fracture patterns and bone strength in comparison to laboratory experiments, and the proposed computational framework integrates a representation of subject-specific fracture energy. This feature is anticipated to be particularly relevant for analyzing osteoporotic patients. Moreover, the ability to predict fracture type could aid in identifying patients at risk of complex fractures.

In conclusion, this paper emphasizes that the understanding of bones in response to mechanical forces, both under normal and extreme conditions, is crucial for advancing orthopedic care. The convergence of experimental and computational techniques highlights the diverse approach required to tackle the complexities of bone failure. By leveraging a combination of theoretical insights, experimental techniques, and computational models, we can better predict, prevent, and treat bone fractures, ultimately improving patient outcomes in orthopedic practices.

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