HIGH RATE IMPACT TO THE HUMAN CALCANEUS: A MICROMECHANICAL ANALYSIS

ABSTRACT

An “underbody blast” (UBB) is the detonation of a mine or improvised explosive device (IED) underneath a vehicle. In recent military conflicts, the incidence of UBBs has led to severe injuries, specifically in the lower extremities. The foot and ankle complex, particularly the calcaneus bone, may sustain significant damage. Despite the prevalence of calcaneal injuries, this bone’s unique properties and the progression of fracture and failure have not been adequately studied under high strain rate loading. This research discusses early efforts at creating a high-resolution computational model of the human calcaneus, with primary focus on modeling the fracture network through the complex microstructure of the bone and creating micromechanically-based constitutive models that can be used within full human body models. The ultimate goal is to develop a micromechanics-based simulation of calcaneus fracture and fragmentation due to impact loading. With the goal of determining the basic mechanisms of fracture propagation through the internal structure of the calcaneus, a two-dimensional model was employed for preliminary simulations with a plane-strain approximation. In this effort, a cadaveric calcaneus was scanned to a resolution of 55 μm using an industrial micro-computed tomography (microCT) scanner. A mid-sagittal plane slice of the scan was selected and post-processed to generate a 2D finite element mesh of the calcaneus that, in situ, would be restrained by the talus. A displacement of 1.25 mm was applied to the heel of the calcaneus over 5 ms. In a typical result, following impact, the strain and stress propagated throughout the cortical shell and then began to radiate into the bone into the bone along the trabeculae. Local stress concentrations can be observed in the trabecular structure in the posterior region of the bone following impact. Upon impact, cortical and trabecular bone show different stresses of 13MPa and 1 MPa, respectively, and exhibit complex high frequency responses. Observed results may offer insight into the wave interactions between the different materials comprising the calcaneus, such as impedance mismatch and refraction. Pore pressure in the marrow may be another important factor to consider in understanding stress propagation in the calcaneus.

INTRODUCTION

An “underbody blast” (UBB) is the detonation of a mine or improvised explosive device (IED) underneath a vehicle. In recent military conflicts, the incidence of UBBs has led to severe injuries, specifically in the lower extremities [1]. The explosion results in both “local” and “global” effects. Local effects include inelastic deformations and the rapid acceleration of the vehicle floor following the blast. Global effects pertain to the gross motion of the vehicle, particularly in the vertical direction [2]. In an idealized scenario, the soldier in the vehicle is the 90-90-90 seated position in which the foot and ankle complex, specifically the calcaneus bone, may sustain significant damage [1]. According to information collected regarding blast victims among UK service personnel, 30 individuals presented 40 calcaneal fractures [3]. Despite the importance of calcaneal injuries, this bone’s unique properties and the progression of fracture and failure have not been adequately studied under high strain rate loading [4].

It is common when modeling the human body for the trabecular bone to be represented as a homogenous mass within
a shell of cortical bone [4][5]. However, based on micro-CT imaging data obtained during this study, this method is likely inaccurate, at least in the calcaneus, and may compromise the biofidelity of the model. Within a hierarchical modeling framework adopted for biological systems, as shown in Figure 1, the components and sub-assemblies are critical to the system-level response. As full (system-level) human body finite element models become more widely used in the military design process, the need for region-specific constitutive models and robust fracture modeling methods at the component and sub-assembly level will increase. This research discusses early efforts at creating a high-resolution computational model of the human calcaneus, with primary focus on modeling the fracture network through the complex microstructure of the bone and creating micromechanically-based constitutive models that can be used within full human body models. The ultimate goal is to develop a micromechanics-based simulation of calcaneus fracture and fragmentation due to impact loading.

The calcaneus is an irregular bone, comprised mainly of trabecular bone and encased in a thin layer of cortical bone [13][6]. The structural pattern of the calcaneus is developed according to the needs of erect, weight bearing human motion. The major load-bearing column consists of principal compressive trabeculae, which are situated in a path between the posterior articular surface of the talus and the heel. Secondary compression occurs in a trabecular group between the talus and the posterior calcaneal surface. The primary tensile load is carried by trabeculae running between the Achilles tendon towards the front of the bone and the secondary tensile load is borne in the arch of the foot. The area lacking significant trabecular structure is known as the “neutral triangle”. A simplified representation of these structural patterns is shown in Figure 2[13].

In the general population, the calcaneus is represented in 60% of fractures to the foot and ankle complex [6]. One fracture mechanism occurs due to excessive axial loading in the angle of Gissane, or the angle of the subtalar surface. This generates a fracture line through the neutral triangle with common secondary fracture lines through the upper or posterior surfaces of the bone [13][6]. Intraarticular fractures, representing 75% of adult calcaneal fractures, are commonly classified by the Sanders system. In this system, type I represents non-displaced fractures. Types II-IV refer to the number of articular pieces resulting from the fracture. An A, B, or C following the number further classifies the fracture types. A, B, and C denote whether the fracture line is located laterally, centrally, or medially, respectively. Extraarticular fractures are represented as Type A, B, or C. Type A refers to an anterior fracture, B to a mid-body fracture, and C to a posterior fracture [6].

![Figure 1: Range of modeling length scales: (a) high level lower extremity model, (b) calcaneus highlighted within the greater structure, (c) raw microCT image data](image1)

![Figure 2: Simplified representation of load bearing trabecular structures in the calcaneus as described by Dhillon [13](image2)

**METHODS**

**Finite Element Analysis**

In using a finite element analysis, dynamic equilibrium is enforced by means of the weak form of the principle of virtual work:

\[
\int_{V_0} \nabla \eta dV_0 + \int_{\Gamma_0} \rho_g b \cdot \eta dV_0 + \int_{\Gamma_0} \bar{f} \cdot \eta dS_0 - \int_{\Gamma_1} \rho_s \alpha \cdot \eta dV_0 = 0
\]  (1)
The internal forces in the body are represented by the first term of this equation. External work is represented by the second and third terms. The final term is representative of virtual work. In this equation, \( P \) refers to the first Piola-Kirchhoff stress tensor, \( \nabla \) represents the material gradient, \( \eta \) is a virtual displacement which satisfies homogenous boundary conditions over \( \partial B_0 \), the \( : \) symbol indicates an inner product of second-order tensors, \( \rho_0 \) is the density, \( b \) represents body forces, \( f \) represents tractions applied to \( \partial B_0 \), and \( a \) refers to acceleration [7].

Anatomic Representation

In this effort, a cadaveric calcaneus was scanned to a resolution of 55 µm using an industrial micro-computed tomography (microCT) scanner. The scans were post-processed and used to generate a finite element mesh of the calcaneus. The raw CT data was imported into image processing software and a mid-sagittal plane slice was selected for analysis. A label field was created based on a density threshold of the CT slice to delineate the bone within the slice. It is interesting to note that cortical shell thickness measurements were taken on both the raw CT slice and the label field. An average thickness of 0.408 mm was found for the cortical shell on the raw slice. However, when the measurements were repeated on the label field, an average thickness of 1.31 mm was found. This discrepancy is attributed to the fineness and complexity of the internal geometry of the bone, especially close to the cortical shell. The detail was difficult to adequately capture when segmenting the bone, and the process should be further refined in future study.

A second label field was created to identify the marrow. The label fields were used to generate surfaces for the bone and the marrow, each of which was exported and further processed using in order to create two-dimensional surface meshes, which were converted to spline curves using and meshed. The calcaneus was modeled with two dimensional plane strain elements. This process is illustrated in Figure 3.

Concurrent modeling of the full lower extremity is being conducted and validated against simulation results from Shin and Untaroiu published in their paper, “Biomechanical and Injury Response of Human Foot and Ankle Under Complex Loading” [8]. In this paper, the lower leg was bolstered at the knee and loading was applied via a plate in contact with the foot. A peak velocity of about 5 m/s was reached about 20 ms into the simulation [8]. This simulation was meant to replicate the experimental tests of Funk et al [9]. Accordingly, validation simulations were conducted with the leg fixed above the knee with vertical translation allowed. A linearly increasing velocity was applied, reaching its peak of 5 m/s at 20 ms and linearly decreasing back to 0 m/s at 40 ms. Sampling of a node at the bottom of the calcaneus showed increasing stresses during the first 5 ms of the simulation, and the displacement during this time was 1.25 mm. Figure 4 shows 5 ms of the macroscale simulation and the stress and displacement over that time in the calcaneus.

Accordingly, simulations for the two-dimensional calcaneus model were conducted with a 1.25 mm displacement over 5 ms. A fixed boundary condition was applied to the upper right corner of the calcaneus to represent the talus. Boundary conditions are shown in Figure 5.

![Figure 3](image1.png)

Figure 3: From left to right are shown the three stages of image processing for the calcaneus scan. (a): Unprocessed mid-sagittal CT slice. (b): Label field delineating the bone. (c): The three regions, cortical bone (blue), trabecular bone (green), and marrow (red) that were meshed.

![Figure 4](image2.png)

Figure 4: Simulations conducted on the full lower extremity show stresses in the calcaneus increasing for the first 5 ms of the simulation with a displacement of ~1.25 mm during that time.

![Figure 5](image3.png)

Figure 5: (a) Displacement over time of the heel of the calcaneus. (b) boundary conditions on the calcaneus in Abaqus simulations. There is a displacement of 50 µm impacting the heel over 1 ms and a fixed boundary condition representing the talus.
Material properties for the trabecular bone, cortical bone, and bone marrow are given in Table 1. Currently, the bone is modeled as elastic perfectly plastic. In future studies, more sophisticated fracture modeling techniques will be employed as well as a mesh dependency study to ensure a converged solution. Results of these simulations are described in the following section.

### Table 1: Material properties for the biological materials comprising the calcaneus

<table>
<thead>
<tr>
<th>Material Properties</th>
<th>Cortical Bone</th>
<th>Trabecular Bone</th>
<th>Marrow</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\nu$</td>
<td>0.325</td>
<td>0.45</td>
<td>0.167</td>
</tr>
<tr>
<td>$\sigma_c$ (MPa)</td>
<td>132 [4]</td>
<td>1.8 [12]</td>
<td>N/A</td>
</tr>
</tbody>
</table>

### Plane strain approximation

With the goal of determining the basic mechanisms of fracture propagation through the internal structure of the calcaneus, a two-dimensional model was employed for preliminary simulations. In the absence of rigorous experimental validation, and given the complexity of developing a 3D model of the calcaneus, it was considered a more practical approach to begin with a plane-strain approximation for early investigation. Similar approaches for two-dimensional approximations have been followed elsewhere for the brain [13][14] and the lower extremity [15][16]. As a primary subject of interest is the wave propagation through the calcaneus following dynamic loading, a plane-strain approximation model should provide reasonable preliminary data. However, one important limitation to consider is that, in two dimensions, some of the trabeculae are represented only as islands which, considering marrow as a fluid, may be free to move in the plane, unlike in a three-dimensional scenario. This problem may be overcome by the viscous nature of the bone marrow and the time scales associated with the simulation, preventing this nature of motion. Though the fracture pattern may be altered, this approximation will give some indication of dynamic wave propagation.

### RESULTS

A typical result for the is shown in Figure 6, which plots the von Mises equivalent stress over the time frame. Following impact, the stress propagates throughout the cortical shell and then begins to radiate into the bone along the trabeculae. This can be seen in the image as most of the cortical shell has reached a stress of greater than .5 MPa before much of the interior structure has reacted to the impact at all. One can note from the images at 1 ms and 2 ms that the stress seems to travel close to the bottom of the calcaneus along the trabeculae towards the front of the calcaneus prior to spreading towards the top of the calcaneus. It is possible that this pattern of motion may derive from the loading paths illustrated in Figure 2. The stress contours seen in the image raise questions about the interesting and unknown relationship between the bone marrow that fills the pores within the trabecular bone and the way in which stresses are transferred through the trabecular structures.

As can be seen in Figure 7, the cortical bone elements show the highest overall stresses of the three materials, due to the higher Young’s Modulus of cortical bone. The marrow elements show much lower stress relative to the other two materials. In all three materials, the stress gradually increases over the timeframe, though fluctuations are seen in each plot, which may be caused by the varying wavespeed through the differing materials.

![Figure 6: The progression of von Mises stress throughout the calcaneus at 1 ms intervals over the course of the 5 ms simulation.](image-url)
DISCUSSION

Examination of the stress patterns over the course of the simulation show local stress concentrations at a variety of points. Figure 8 shows an example of this towards the front of the calcaneus, but additional instances have occurred at other locations in the trabeculae, and generally higher stresses are seen throughout the cortical shell.

The differing material properties between cortical bone and trabecular bone can account for the higher stresses in the cortical shell. However, the reason for the distribution of local stress concentrations in the trabecular structure is less clear. It may be related to the loading paths described in the introduction and illustrated in Figure 7. Another possibility is based on the relative wave speeds in the different materials. Table 2 provides the longitudinal and transverse wave speeds for each material comprising the calcaneus. Additionally, the time necessary for a wave to propagate across the length of the calcaneus (~0.0625 m) and the height of the calcaneus (~0.0376 m) is given for each material. These values are averaged for each material to determine the approximate time for a wave to propagate across the whole calcaneus. These values are then averaged to find the approximate wave speed for the model used. As is evident, the wave speeds for the materials composing the calcaneus are very different. A longitudinal wave travels more than 5 times faster in cortical bone than in trabecular bone, and almost 76 times faster than in marrow. Similarly, a transverse wave traveling through cortical bone travels 9 times faster or 61 times faster through cortical bone than through trabecular bone or marrow, respectively. These wave speed discrepancies could account for the peaks and troughs seen in the mid and back cortical bone elements. Interface impedance could also be a contributing factor to the local stress concentrations seen in Figure 8.

Figure 7: A graph of stress versus time for elements in the front, middle, and back of the calcaneus for cortical bone, trabecular bone, and bone marrow.
Developed a high-dimensional approximation of a human calcaneus. In order to be able to implement high-resolution findings in full body models, a three-dimensional calcaneus model must be developed. A high-resolution constitutive model of human bone structure is critical to quantifying the biomechanical response of the lower extremities under high impact loading. However, the microarchitecture of trabecular bone is extremely complex, and thus enormously expensive to evaluate computationally. It is therefore highly desirable to produce an efficient, but representative model.

Extensive studies in bone histomorphometry have demonstrated that trabecular bone is composed of an anisotropic network of trabeculae that take the shape of either rods or sheets [17]. This construction generates more than ten times the surface area of cortical bone [18]. It is this extensive surface that makes computational evaluations so expensive. Consequently, approximations must be applied to perform efficient calculations. This research attempts to perform the approximations by reducing the trabecular microarchitecture to elementary geometry. This is achieved by reducing the bone to a solid volume element and then reininserting artificial porosity through the subtraction of basic volume types.

The process was begun with a stereo-lithographic model embracing all internal and external surfaces of the bone. To start the procedure, the model must be converted into a shell. Fortunately, a method for converting point clouds into external surfaces has already been developed. Using the shrink-wrapped boundary face (SWBF) algorithm developed by Bon Ki Koo et al., it is fairly straightforward to convert the points that comprise the bone model into a singular surface [19]. Once the external surface has been created, the porosity must be generated. Each hole that composes the porosity will be based on an elementary geometric shape, so its center must be defined. There are a number of ways to achieve this, but the most efficient method generates a distribution of points based on three parameters derived from the original microstructure: shell thickness, porosity, and anisotropy weighting. The first two are scalars, but the anisotropy weighting is a vector that describes the axis of higher density (which results from Wolff’s Law).

Using the anisotropy weighting, a gradient can be created that favors areas further from the vector. This can then be combined with the porosity value to create the average distance between points for a given location. Once we have the distance between points for each location is determined, every point can be generated from a single seed. However, these conditions will not constrain the points inside the original surface, nor will they prevent the points from entering the cortical shell. This is why the shell thickness is required. Using least squares approximation, the center of the model can be found. Then a radial vector is defined for each point from the center. The shell thickness is then subtracted from each radial vector. This effectively shrinks the model to exclude the cortical shell. Once the model has been shrunk, every point outside the model must be excluded. Fortunately, there is open source software available to complete this. The final step of the process is to use these points to generate holes in the solid model using industrial software. Scripts can be written to read in the external surface and generated points, then subtract basic geometries, thereby creating a representative model.

Simulations have thus far been conducted for a two-dimensional approximation of a human calcaneus. In order to be able to implement high-resolution findings in full body models, a three-dimensional calcaneus model must be developed. A high-resolution constitutive model of human bone structure is critical to quantifying the biomechanical response of the lower extremities under high impact loading. However, the microarchitecture of trabecular bone is extremely complex, and thus enormously expensive to evaluate computationally. It is therefore highly desirable to produce an efficient, but representative model.

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Table 1: Longitudinal and transverse wave speeds in the various materials composing the calcaneus.

<table>
<thead>
<tr>
<th></th>
<th>Cortical Bone</th>
<th>Trabecular Bone</th>
<th>Marrow</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Longitudinal Waves</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Longitudinal Wavespeed (m/s)</td>
<td>3559</td>
<td>665.2</td>
<td>46.89</td>
</tr>
<tr>
<td>Time across length (s)</td>
<td>1.756E-05</td>
<td>9.395E-05</td>
<td>0.001333</td>
</tr>
<tr>
<td>Time across height (s)</td>
<td>1.055E-05</td>
<td>5.646E-05</td>
<td>0.0008011</td>
</tr>
<tr>
<td>Average time across (s)</td>
<td>1.406E-05</td>
<td>7.521E-05</td>
<td>0.001067</td>
</tr>
<tr>
<td>Average for all materials (s)</td>
<td>0.0003854</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Transverse Waves</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Transverse wavespeed (m/s)</td>
<td>1812</td>
<td>200.6</td>
<td>29.65</td>
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<tr>
<td>Time across length (s)</td>
<td>3.449E-05</td>
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<td>0.002108</td>
</tr>
<tr>
<td>Time across height (s)</td>
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<td>0.0001873</td>
<td>0.001267</td>
</tr>
<tr>
<td>Average Time across (s)</td>
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<td>0.0002494</td>
<td>0.001688</td>
</tr>
<tr>
<td>Average for all materials (s)</td>
<td>0.0006549</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

**ONGOING WORK**

Simulations have thus far been conducted for a two-dimensional approximation of a human calcaneus. In order to be able to implement high-resolution findings in full body models, a three-dimensional calcaneus model must be developed. A high-resolution constitutive model of human bone structure is critical to quantifying the biomechanical response of the lower extremities under high impact loading. However, the microarchitecture of trabecular bone is extremely complex, and thus enormously expensive to evaluate computationally. It is therefore highly desirable to produce an efficient, but representative model.

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such as ellipsoids, at those points. Current progress is shown in Figure 9.

![Figure 9: Comparison of 3D surface generated from raw CT data and current progress from mathematical approximation of porosity. As seen in the zoomed in view, spheres represent empty spaces created within a solid model.](image)

### CONCLUSION

As a concurrent research effort to three-dimensional modeling of the full lower extremity, research has been conducted into the impact response of the calcaneus in a high resolution, two-dimensional model. This paper has presented preliminary study of the dynamic response of the calcaneus bone subject to impact loading. Our initial results indicate a complex wave propagation problem due to the various materials (cortical bone, trabecular bone, and bone marrow) and their wide range of mechanical properties. Much of our analysis focused on the length and time scales associated with the dynamic wave propagation without considering the inelastic properties and damage. This is reserved for ongoing and future work. Additionally, in order to implement findings regarding the effect of porosity on the response of the calcaneus under high impact loading, a mathematical method for inserting porosity into the internal structure of a bone is being developed. As this method continues to be improved, it can be applied to other bones to improve the overall biofidelity of human computational models.

### REFERENCES


