I, Vineet Chaturvedi, hereby submit this original work as part of the requirements for the degree of Master of Science in Mechanical Engineering.

It is entitled:
Mechanical Testing and Modeling of the Human Index Finger Distal Pad

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Mechanical Testing and Modeling of the
Human Index Finger Distal Pad

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Abstract

The mechanics of contact between human hand and the external objects is very important in understanding the response of skin and the underlying tissue during dynamic loading. In the current study, the distal pad of the human finger was modeled to study the response to different loading conditions. Experimental data was obtained by conducting indentation, confined and unconfined compression on human cadaver fingers. A material model was developed for the finger tissue based on these experiments. Abaqus analysis software was used to study the response of distal pad of human finger during indentation due to a cylindrical indentor.
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Chapter 1

Introduction

1.1 Background

When it comes to daily activities, the hand is one of the most important parts of a human body. In the consumer packaging goods industry, the interaction between the hand and the product is an important consideration and many researchers have studied the mechanical behavior of the skin and tissue while performing various activities.

1.2 Objectives

The purpose of this research was to study and characterize the behavior of human hand, index finger and thumb, in compression. A finite element model of the human index finger and thumb was developed to compare and analyze the response with respect to the results of mechanical testing on human cadaver fingers. Using the results from the unconfined compression of actual human skin and tissue, a material model was developed and used in finite element simulation. The mechanical response of the actual indentation experiment was compared to the simulated response.
1.3 Thesis Outline

This thesis is comprised of six chapters. Chapter 1 gives the overview and introduction to the research and this paper. Chapter 2 describes the background research work that has been done in this field along with the material models that have been used for characterizing the behavior of human soft tissue and skin. This chapter also refers to the research that was used as motivation for this thesis. Following Chapter 2, the materials and methodology used in this research are described in Chapter 3. Chapter 4 discusses the material and the finite element models generated for the purpose of simulating the response of the finger. Chapter 5 discusses the results from the simulation and compares it with the actual indentation results on human cadaver fingers. We conclude the thesis in Chapter 6 with a discussion of potential future work.
Chapter 2

Background

2.1 Anatomy: Index Finger and Thumb

The human hand is comprised of five digits (four fingers and a thumb). Figure 2.1 shows the gross anatomy of the human hand including the description of the finger joints. The figure focuses on the index finger and thumb joints to be specific. Bone (metacarpals) segments called phalanges are connected to each other by ligaments, and tendons connect the hand muscles together. Fasciae connect the muscles to each other. Together they help provide the structure and strength to perform different physical activities on a daily basis.

Each finger is comprised of three phalanges which extend from the second metacarpal of the human hand. The proximal phalanx acts as the base of each finger, which connects to the metacarpals through the metacarpophalangeal (MCP) joint. At the end of the finger, the distal phalanx provides support to the soft tissue (finger pads) of the fingertip and connects to the rest of the finger through the distal interphalangeal (DIP) joint. The index, middle, ring, and fifth fingers have proximal, middle, and distal phalanges and three hinged joints: distal interphalangeal (DIP), proximal interphalangeal (PIP), and
metacarpophalangeal (MCP).

The thumb has a distal and proximal phalanx as well as an interphalangeal and MCP joint. The thumb plays a crucial part in grabbing and holding objects in day-to-day life. The range of motion of a human thumb varies greatly from bending at its joint to acting as a support for other fingers while holding an object. Anatomically, it is comprised of the metacarpal, which is connected to the wrist through the carpometacarpal (basiliar)
joint (Fig. 2.1). The metacarpal is connected to the proximal phalanx of the thumb through the MCP joint. The thumb’s distal phalanx is connected to the proximal phalanx through the interphalangeal joint. The distal phalanx supports the soft tissue which is the finger pad of the thumb. The distal phalanx-soft tissue area are together referred as the fingertip of the thumb.

2.2 Literature Review

Skin and soft tissue in human bodies are comprised of complex collagen fiber and tissue matrix. Collagen matrix contributes towards the mechanical nature of the skin and the underlying tissue matrix provides support. Load in tension is carried by the collagen fibers whereas the tissue matrix carries the majority of the load in compression. It behaves like a composite nonlinear material [28].

Throughout the literature, various tests have been conducted in tension, compression and shear deformation on soft tissue to assess and study the mechanical behavior of the skin. A study of the human finger soft tissue in compression might be the most accurate to observe the mechanical response since human finger pads are under compression during physiological loading conditions. When the hand is in contact with an object, the deformation in soft tissue occurs due to compression.

Ming-li et al [18] measured the maximum voluntary isometric contractions (MVIC) at various points along the middle of the proximal phalanx of the index finger. Isometric contractions means the contractions resulting from pushing and pulling against an object producing force. During the procedure, the force was produced on the fingers by pulling on a plastic ring through a cable connected to a force transducer. The data was recorded in 16 directions distributed evenly within 360 degrees. The data was collected to analyze the directional dependence of the force produced on the finger tip. This research mainly analyzes the motor functions related to the hand. The testing was done on the index
finger, however the data recorded was not used to characterize the mechanical behavior of the fingertip.

Tada et al [25] used MR (magnetic resonance) compression test data to generate a finite element model of the human fingertip and visualize 3-D deformation of the subcutaneous tissue. The contraction force was measured by using a magnetic resonance compatible force sensor which records the deformation and force in the soft tissue during compression under MRI. The data collected from the compression of the fingertip using MR technique was then used to estimate material constants for the skin (epidermis) and the soft tissue.

Jindrich et al [17] used voluntary dynamic tapping from subjects across several different motor control tapping conditions. The force versus displacement data collected in this case was used to calculate the viscoelastic model parameters (non-linear stiffness and time dependent relaxation coefficients) for the human index finger distal pad. Serina et al [23] used repeated tapping (dynamic compressive loading) of the fingertip on a contact plate with a force transducer at various angles and tapping rates to identify factors that influence the dynamics of the finger pads. In the literature, many studies have also been conducted on analyzing the grip force and strength to predict the behavior of fingertips to different physiological loading conditions [11, 8]. These studies, however, are not directly relevant to our study of the mechanical response of finger tissues.

Many studies have been conducted to study the behavior of distal pads of human fingers using compression testing. Wu et al [28] used pig skin and soft tissue to study the mechanical behavior of skin/soft tissue in compression. Uniaxial compression tests were performed on sample at four different loading speeds of (0.5, 1, 40, 400 µm/s). Stress relaxation tests were performed at a fast rate of 1mm/s with samples being compressed to a specified strain (5% for confined and 8% for unconfined compression). Wu et al [32] conducted a similar type of testing in 2003 in which a flat indenting probe was used on the index finger distal pad to study the dynamic interaction between the flat surface
and the finger pads. Stress relaxation tests were conducted on the distal pads. Two series of displacement controlled tests (2 mm and 3 mm), at variable loading speeds, were conducted and the displacement was held constant for 30 seconds to observe the relaxation in the given sample.

Wu et al [31] also studied the tactile performance of human fingertip by developing a multi-layered two-dimensional finite element model for the fingertip. They performed two series of numerical tests on the fingertip. The first series of tests simulated the behavior of the fingertip under line load. A finite element model was used to compute the reaction force, displacement, fluid pressure, stress and strain within the soft tissue due to the load. The results obtained from the simulation were compared with the experimental results from Srinivasan et al [24] who conducted experiments to record deflections of primate fingertip under line loads. The second series of tests done by Wu et al [31] used one point and two point discrimination tests on the fingertip to study the mechanics of tactile sensation of the fingertip. The one point test used one steel bar to indent the surface of the fingertip and two point test used two steel bars to deform the skin on the fingertip at the same time. The displacement, stress and strain were recorded on the soft tissue for a maximum depression of 1 mm. This data was used to predict the material parameter for hyperelasticity of the skin, subcutaneous tissue and also the time-dependent viscoelastic response of the skin under mechanical loading.

Another study done by Wu et al [32], predicted the nonlinear and time dependent force response of the fingertip during interaction with a flat indentor. They used experimental and theoretical analysis to predict and analyze the behavior of the fingertip. The experiments were done on actual human subject who agreed to be a part of the study (2 male and 2 female). The index finger of the right and the left hand of each subject was compressed using a flat indentor for a specified displacement. A physical model was suggested to simulate the nonlinear and time-dependent behavior of the fingertip. The force relaxation test data at a fast loading rate was used to identify the material
parameters for the soft tissue in the fingertip. The nonlinear and time dependent behavior of the fingertip was studied from the response of the force versus displacement plots obtained from the experiments.

Table 2.1 summarizes a few of the studies that have been done to study the behavior of human skin and soft tissue.

<table>
<thead>
<tr>
<th>Group</th>
<th>Type of test</th>
<th>Method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ming-li et al [18]</td>
<td>Tension</td>
<td>Voluntary isometric contraction</td>
</tr>
<tr>
<td>Tada et al [25]</td>
<td>Compression</td>
<td>Magnetic resonance</td>
</tr>
<tr>
<td>Jindrich et al [17], Serina et al [23],</td>
<td>Compression</td>
<td>Dynamic compressive loading</td>
</tr>
<tr>
<td>Wu et al [28]</td>
<td>Compression</td>
<td>Uniaxial compression</td>
</tr>
<tr>
<td>Wu et al [31], Srinivasan et al [24]</td>
<td>Compression</td>
<td>Finite element analysis</td>
</tr>
<tr>
<td>Wu et al [32]</td>
<td>Compression</td>
<td>Experimental and theoretical analysis</td>
</tr>
</tbody>
</table>

Table 2.1: Mechanical characterization of skin-soft tissue in literature

### 2.3 Constitutive Modeling Review

Modeling soft tissue and skin to predict the behavior during different physiological loading conditions has been a topic of great interest in the field of biomechanics. The presence of tissue matrix, collagen fibers and multiple layers of skin makes the process very detailed and tedious. Accurately representing the behavior through constitutive modeling is very critical in the study of soft tissue and skin behavior in human fingers.

Hu et al [16] characterized the human skin and soft tissue as an incompressible, homogenous and isotropic elastic material. They studied the behavior of the skin and soft tissue using the Mooney-Rivlin and Ogden models and compared their results to those in the literature. Wu et al [32, 33, 31] studied the behavior of the tissue characterizing it as a hyperelastic and linearly viscoelastic material exhibiting rate-dependence and
non-linear force-displacement relationship. Much of the research done in the field of soft tissue and skin biomechanics model the components as hyperelastic or viscoelastic materials.

The work of John Wu et al [32, 33, 31] uses a third order Ogden model to characterize the materials used in the experimental testing and compare them to the behavior of the skin and soft tissue during real physiological loading conditions. Their model also incorporates the viscoelastic time dependence of the material in the model along with the hyper elastic and non-linear force and displacement behavior of the stress-strain. The implementation of Ogden model follows the assumption that the mechanical behavior of the skin tissue is considered to be nonlinear and viscoelastic.
Chapter 3

Materials and Methods

3.1 Materials

Mechanical testing was carried out on two materials: a neoprene foam, and index fingers and thumbs from human cadavers. Neoprene foam (McMaster-Carr, Elmhurst, Illinois) was used to verify the testing technique and to calibrate the testing experiment. Cadaver index fingers and thumbs were used for the actual testing. The fingers were extracted from human cadavers (two males and two females) from the medical anatomic donations lab at the University of Cincinnati. A signed approval from the Institutional Review Board (IRB) was not required since the research did not involve any risk to any living subject.

3.2 Testing Fixture and Equipment

Two different fixtures were designed and machined to conduct experiments. The fixtures for testing were designed in Solidworks [7]. The fixtures were then machined at the College of Engineering and Applied Science machine shop to the desired specifications.
3.2.1 Indentation

Figures 3.1 and 3.2 show the design of the indentation fixture. The design has two adjustable L-shaped angles on the sides and one L-shaped angle on the top used to hold the testing finger sample in a fixed position. Additional adjustable screw slots were machined in the base of the fixture to prevent any movement of the fixture during the course of the experiment. Figure 3.2 shows the side view of the fixture with the adjustable screw slots in the clamps and the base of the design. The slots were adjustable because the width of the cadaver index fingers and thumb varies from subject to subject.

Figure 3.1: The front view of the fixture design for indentation
3.2.2 Confined Compression

Figures 3.3 and 3.4 show the design of the fixture for confined compression. A cylindrical fixture with the diameter of 76 mm was designed and machined. A cylindrical chamber of diameter 5 mm was made through the middle of the fixture for the sample to be placed. A filter with hole density of $40 \mu\text{m}$ was placed at the bottom of the fixture in a circular slot (1.5 mm in thickness) for the tissue fluid to escape through during confined compression testing. This fixture was then placed on the base with circular cross section and the probe used for indentation was used for confined compression as well. Cylindrical samples with approximate diameter of 5 mm were extracted from the distal pad to be placed in the cylindrical chamber. Match marking on the fixture and the probe was used to align the indentor and the hole of the cylindrical fixture. In order to avoid and minimize the effects of friction between the indentor sides and the inside walls of the cylindrical chamber, motor oil was used as lubricant on the indentor. The inside of the chamber in the fixture was also coated with teflon to minimize the frictional
effect to some extent.

Figure 3.3: Design of the confined compression fixture

Figure 3.4: Bottom view of the confined compression fixture

3.3 Methods

Three different testing methods were employed to test the mechanical behavior of soft tissue: indentation, confined compression and unconfined compression. No tension tests
are performed because it is assumed that the finger tissue is primarily under compression during physiological loading.

3.3.1 Test Sample Preparation

After extraction the cadaver fingers were stored in a $-80^\circ$C freezer in bio-hazard bags until testing. The day before the testing, the samples were moved into a $-20^\circ$C freezer. The samples were then placed in phosphate buffered saline (PBS) solution to condition and mimic the biochemical environment in human body. The fingers are stored in PBS for three hours before the actual testing to thaw for use in testing.

Indentation samples were simply the cadaver fingers placed in the custom made fixture introduced in Section 3.2. For unconfined compression testing, rectangular samples were extracted from the index and thumb finger samples using a scalpel. Each sample was extracted carefully to include the skin and subcutaneous tissue. The height of the sample varied between fingers. Cylindrical specimens for confined compression were cut using a steel biopsy punch at a temperature of $-20^\circ$C. The punch had a sharp edge, tapered outside wall and a straight inside wall with a diameter of 5 mm.

Figure 3.5 shows a typical sample used for confined compression testing.
3.3.2 Indentation Testing

Figure 3.6 shows the setup of an index finger indentation testing.

Figure 3.6: Index finger testing setup during indentation

Slot A was placed at the top of the finger and slots B and C were placed on the sides to hold the finger in place. The cadaver finger sample was placed in between the
slot B and slot C. Each finger was placed between the slots and slots were tightened to get accurate reading from the same point on the distal pad. The indentation probe is cylindrical and made of hardened steel (McMaster-Carr, Elmhurst, Illinois). The probe was attached to a 10 N load cell with a resolution of 0.1%. After establishing contact, the indentor was lowered to achieve a 2 mm indentation, and the reaction force was recorded. This was repeated for each of the thumb and index fingers from each donor.

### 3.3.3 Unconfined Compression

The second testing method employed was unconfined compression. Figure 3.7 shows the setup for unconfined compression. Tissue samples of measured dimensions were extracted from distal pads of the cadaver index and thumb fingers as described earlier. Each rectangular sample was then placed on top of a plate and an aluminum probe with a rectangular cross-section was used to compress the sample. The sample was placed within the confines of the boundaries of the probe. The testing was again displacement controlled and the sample was compressed to 50% of its original thickness. The force-displacement data was recorded and used to determine the stress-strain response of the sample.

![Experimental setup for unconfined compression](image)

Figure 3.7: Experimental setup for unconfined compression
3.3.4 Confined Compression

Figure 3.8 shows the experimental setup for confined compression. The indentor position was zeroed right above the sample. The sample was compressed to 50% of its thickness and force-displacement data was collected.

![Confined Compression Experimental Setup](image)

Figure 3.8: Top view of confined compression experimental setup
Figure 3.9: Full experimental setup inside PBS bath
Chapter 4

Computational Methods

In this chapter, we discuss the computational methods used in the course of this research. First, we discuss the development of constitutive models for finger tissue. Next, we discuss the finite element models developed to carry out various simulations.

4.1 Material Modeling

The constitutive model of a given material is developed using the classical theory of continuum mechanics. The reference configuration is the initial state of the material particles of the body under consideration. This is also referred to as the undeformed configuration. The current configuration refers to the deformed state or new position of the material particles after undergoing deformation. Let $\mathbf{X}$ denote the initial position and $\mathbf{x}$ denote the current position of the particle in the body. The deformation gradient $\mathbf{F}$ is

$$
\mathbf{F} = \frac{\partial \mathbf{x}}{\partial \mathbf{X}}
$$

(4.1)

The right Cauchy-Green deformation tensor is
The left Cauchy-Green deformation tensor is

\[ \mathbf{C} = \mathbf{F}^T \mathbf{F} \]  \hfill (4.2)

The Green-Lagrange strain tensor is

\[ \mathbf{E} = \frac{1}{2} (\mathbf{C} - \mathbf{I}) \]  \hfill (4.4)

where \( \mathbf{I} \) is the second order identity tensor. The principal invariants of \( \mathbf{C} \) are denoted as \( I_k \), and are given by

\[ I_1 = tr(\mathbf{C}) \]  \hfill (4.5)

\[ I_2 = \frac{1}{2} \left( (tr(\mathbf{C})^2 - tr(\mathbf{C}^2)) \right) \]  \hfill (4.6)

\[ I_3 = det(\mathbf{C}) \]  \hfill (4.7)

### 4.1.1 Hyperelasticity

The behavior of a hyperelastic material can be completely characterized using a strain energy density function \( W \). Assuming material isotropy, the second Piola-Kirchoff stress tensor is given by

\[ \mathbf{S} = \frac{\partial W}{\partial \mathbf{E}} = 2 \frac{\partial W}{\partial \mathbf{C}} \]  \hfill (4.8)

where \( \mathbf{C} \) is the Cauchy-Green deformation tensor defined in equation (4.2). Then Cauchy stress \( \mathbf{T} \) is given by

\[ \mathbf{T} = J^{-1} \mathbf{F} \mathbf{S} \mathbf{F}^T = 2J^{-1} \mathbf{F} \frac{\partial W}{\partial \mathbf{C}} \mathbf{F}^T \]  \hfill (4.9)
The quantity $J$ is a measure of volumetric change and is given by

$$J = \det(F) \quad (4.10)$$

In isotropic hyperelastic materials, the cauchy stress $T$ can be defined in terms of the invariants of the right Cauchy-Green deformation tensor, $C$ and principal stretches, $\lambda_k$.

$$T = 2J^{-1}F \left( \frac{\partial W}{\partial I_1} \frac{\partial I_1}{\partial C} + \frac{\partial W}{\partial I_2} \frac{\partial I_2}{\partial C} + \frac{\partial W}{\partial I_3} \frac{\partial I_3}{\partial C} \right) F^T \quad (4.11)$$

$$T = 2J^{-1}F \left( \frac{\partial W}{\partial \lambda_1^2} \frac{\partial \lambda_1^2}{\partial C} + \frac{\partial W}{\partial \lambda_2^2} \frac{\partial \lambda_2^2}{\partial C} + \frac{\partial W}{\partial \lambda_3^2} \frac{\partial \lambda_3^2}{\partial C} \right) F^T \quad (4.12)$$

For incompressible, isotropic hyperelastic materials, $J=1$. Hence

$$I_3 = \det(C) = 1 \quad (4.13)$$

The Cauchy stress can now be written as

$$T = -pI + 2F \frac{\partial W}{\partial C} F^T \quad (4.14)$$

where $p$ is a Lagrange multiplier term which is determined from the boundary conditions of the given problem.

For the purpose of this study, third order Ogden model is used to predict the hyperelastic response of soft tissue and skin. There are different strain energy density potentials used to predict the behavior of different hyperelastic material. The Ogden model is used frequently for biological soft tissue. Abaqus [14] uses the following form of Ogden model, where the strain energy density is in terms of the principal stretches $\lambda_k$

$$W = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^2} \left[ \lambda_1^{-\alpha_i} + \lambda_2^{-\alpha_i} + \lambda_3^{-\alpha_i} \right] + \sum_{i=1}^{N} \frac{1}{D_i} (J - 1)^{2i} \quad (4.15)$$
where \( \lambda_k \) are the principal stretches, \( J \) is the volumetric ratio, and \( N \) is the order of the Ogden model. The parameters \( \mu_i, \alpha_i, D_i \) are the material-dependent values. The term \( \sum_{i=1}^{N} \mu_i \) is the shear modulus of the material.

### 4.1.2 Viscoelasticity

Material models discussed in Section 4.1.1 are used to predict the response of materials where there is no time-dependent behavior. In the case of biological soft tissues, the stress-strain response of the material changes with time. Therefore, the material model has to include the time dependent effects. Viscoelastic material models predict both the viscous and the elastic behavior of the soft tissue.

Stress relaxation testing is usually conducted to obtain the viscoelastic properties of the material. In stress relaxation testing, the material is instantaneously loaded to a certain strain and held there for a long period of time. This deformation results in sudden and rapid increase in stress during the loading part and then the material relaxes eventually to an equilibrium stress. The experimental data obtained in this manner is used to estimate the viscoelastic material model parameters in this study. In a viscoelastic material, the Cauchy stress as a function of time is obtained using

\[
T(t) = G(t, \epsilon) \quad (4.16)
\]

where \( G(t, \epsilon) \) is called the stress relaxation function, \( t \) is the time, and \( \epsilon \) is the strain. In this study, we assume that the stress relaxation function can be separated into material response and time-dependent response and given by

\[
G(t, \epsilon) = G_R(t)T(\epsilon) \quad (4.17)
\]

The history-dependent stress response is obtained using
\[
T(\epsilon, t) = \int_{-\infty}^{t} G_R(t - \tau) \frac{\partial T(\epsilon)}{\partial \epsilon} \dot{\epsilon} dt
\]  \hspace{1cm} (4.18)

where \( \dot{\epsilon} \) is the strain rate, \( \tau \) is the time constant, \( \frac{\partial T(\epsilon)}{\partial \epsilon} \) is obtained from the hyperelastic response of the material discussed in the previous section, and \( G_R(t) \) is the stress relaxation function. The stress relaxation function is expressed as a Prony series in this study. It is given by

\[
G_R(t) = E_0 \left( 1 - \sum_{i=1}^{N} p_i (1 - e^{-\frac{t}{\tau_i}}) \right)
\]  \hspace{1cm} (4.19)

where \( E_0 \) is the instantaneous modulus of the material (units of stress), \( p_i \) are the Prony constants (dimensionless) and \( \tau_i \) are the time constants (units of time). The instantaneous and the long term response of the relaxation function is

\[
G_R(t) = \begin{cases} 
E_0, & \text{if } t = 0, \\
E_0 \left( 1 - \sum_{i=1}^{N} p_i \right), & \text{if } t = \infty.
\end{cases}
\]  \hspace{1cm} (4.20)

4.2 Finite Element Model

4.2.1 Finger Model Assembly

Geometry files were obtained from a CT scan of human hand as .stl files. Commercial meshing software HyperMesh [3] was used to import the input geometry file. The geometry file was then exported to SolidWorks [7] software to convert them into solid files. The distal pad and distal bone of the index finger were then extracted. The extracted components were then taken back to HyperMesh and meshed using solid tetrahedral elements for tissue (C3D10H) and rigid shell element for bone (S4). Each component was meshed separately. The skin surface was generated by projecting a solid surface
from the outer layer of the tissue. Skin was then modeled as a two-layered solid with a thickness of 1.3 mm. The skin surface was generated from the outer surface of the tissue to maintain the consistency of the geometry. The element size for tissue and skin was kept at 0.15 mm. Figures 4.1, 4.2 and 4.3 show the individual meshed components of the distal pad of human finger. Figure 4.4 shows the assembly of the components set up for indentation simulation.

![Figure 4.1: Solid skin component (C3D10H elements)](image1)

![Figure 4.2: Solid tissue component (C3D10H elements)](image2)
During the simulation of the distal pad in Abaqus, a number of issues were encountered, which caused the indentation simulation to fail. The outer and inner surface of the finger were uneven. This, and the geometric, material and contact nonlinearities inherent to the problem were identified as potential causes of the simulation failure. Due to the irregular geometry, the meshing of the tissue surface as well as the skin surface was complicated. Overall failure of the simulation process was narrowed down to three different causes

- Geometric irregularity
- Geometric and contact nonlinearity
- Material nonlinearity
The results in the next chapter (Section 5.1.4) discuss steps taken to address these issues.

4.2.2 Cylinder Model Assembly

A simpler geometry model was created and the finger structure was modeled as a simple cylinder with three components of skin, tissue and bone. Bone was once again modeled as a rigid shell structure. Element sizes were kept consistent throughout the model at 0.15 mm. Skin and tissue were modeled using eight noded hexahedral elements (C3D8H). These elements are, in general, considered more accurate than tetrahedral elements. With the geometry simplified, the analysis was successfully completed and results were obtained which will be discussed in the next section. Figures 4.5 to 4.7 show the different components of the cylindrical model and the assembly.

![Cylindrical model](image)

**Figure 4.5:** Solid skin cylindrical component (C3D8H elements)
Figure 4.6: Solid tissue cylindrical component (C3D8H elements)

Figure 4.7: Bone cylindrical component (S4 elements)

Figure 4.8: Cylindrical model assembly for indentation
Chapter 5

Results and Discussion

Following the discussion of the experimental methods and procedures, the results are discussed in this section. The order of the results is according to the order in which the materials were tested. The experimental setup is first calibrated and optimized using neoprene foam material. The equipment is then used for testing the samples obtained from human cadavers.

5.1 Foam

5.1.1 Indentation

The setup of the indentation experiment is shown in Fig. 3.6. The force-displacement response from the indentation testing on the foam sample is shown in the Fig. 5.1.

The calibration of the equipment is performed using four loading and unloading cycles of indentation done on neoprene foam material. The behavior shows that each of the four loading and unloading cycles follows and traces the previous cycle with very slight loss of strain energy.
5.1.2 Unconfined Compression

Figure. 3.7 shows the experimental setup for unconfined compression using a flat indentor with a rectangular cross section. The samples were compressed to 50% of their thickness at a loading rate of 0.1 mm/s. This process was repeated for four cycles.

Figure 5.2 shows the force-displacement response from uniaxial unconfined compression of neoprene foam sample.

Figure 5.2: Load-displacement response of neoprene foam from unconfined compression

\[ E = 0.7 \text{ MPa} \]
The force-displacement response of the neoprene foam shown in Fig. 5.2 was used to obtain the stress-strain response. The area of the rectangular foam sample was measured and used to calculate the stress-strain response of the material. This response was then used to calculate the Young’s modulus. Neoprene foam’s calculated Young’s modulus of (0.7 MPa) falls within the specified range of 0.5 Mpa to 2.0 MPa [5].

5.1.3 Confined Compression

Figure 3.9 shows the experimental setup of confined compression. A cylindrical indentor was used to compress the sample placed in cylindrical chamber. Once again, the sample was compressed for four cycles of loading and unloading.

Figure 5.3 shows the response of the setup without any foam material sample to observe the effect of friction between the indentor and the inside wall of the fixture chamber. Figure 5.4 shows the load-displacement response recorded with a foam sample in the cylindrical chamber.

Figure 5.3: Confined compression testing without any sample
In the experimental setup shown in Fig. 3.9, the inside chamber of the cylindrical fixture was coated with teflon to minimize friction between the steel indentor and the inside wall of the fixture. The indentor was also coated with motor oil to reduce the noise in the experiment even further. The zero position of the indentor corresponds to the point just before the indentor touches the sample in the chamber. The loading rate at which the indentor compresses the sample is 0.1 mm/s. The procedure is again displacement controlled and the sample is compressed to 50% of its thickness.

5.2 Index Finger and Thumb

Once the equipment was calibrated, it was used to test the finger samples and the soft tissue-skin samples obtained from human cadavers.

5.2.1 Indentation

The procedure for indentation of tissue is the same as that mentioned in Section 5.1.1 for the foam sample. The force-displacement response was obtained as shown in Fig. 5.5.
5.2.2 Unconfined Compression

During unconfined compression, the same setup was used as mentioned in Section 5.1.2. The foam sample was replaced by a rectangular sample of skin-tissue from the distal pad of the finger (Fig. 5.7) and compressed to 50\% of the thickness of the sample. The
thickness of the skin-tissue samples varied from 1 mm to 4 mm.

Figure 5.7: Skin-tissue sample used during unconfined compression

Figure 5.8 shows the force-displacement behavior of the skin-tissue sample during unconfined compression extracted from the index finger.

Figure 5.8: Load-displacement response for index finger unconfined compression
The skin-tissue sample from index finger sample was then replaced with a sample extracted from the thumb. Figure 5.9 shows the behavior of the skin-tissue sample from the thumb.

Figure 5.9: Load-displacement response for thumb unconfined compression

5.2.3 Confined Compression

During confined compression, the same setup was used as the foam but the sample was replaced with a cylindrical sample of tissue in the cylindrical well. The sample was extracted from the distal pad of the finger using a biopsy punch with 5 mm diameter. Figure 3.9 shows the full experimental setup. Figures 5.10 and 5.11 show the force-
displacement response of the skin-tissue sample obtained from index finger and thumb under confined compression.

![Confined Compression Index Finger](image)

Figure 5.10: Load-displacement response for index finger under confined compression

![Confined Compression Thumb](image)

Figure 5.11: Load-displacement response for thumb under confined compression
5.3 Finite Element Simulation

As described in Section 4.2, the geometry file (.stl) of human hand was imported into HyperMesh [3] and the index finger and thumb were extracted using SolidWorks [7]. The distal pad of the index finger was extracted and was then imported back in HyperMesh and meshed using 0.15 mm size solid tetrahedral elements (C3D10H). Skin was generated by projecting a surface from the outside layer of the tissue in HyperMesh. Each component (skin, tissue and bone) was meshed separately and assembled together in Abaqus [14]. This assembly was then set up to run the indentation simulation in Abaqus. The numerical solution of the indentation diverged and failed after reaching 0.3 mm of the 1 mm total displacement. Possible reasons are discussed in Section 5.3.2.

5.3.1 Material Model Verification

Uniaxial unconfined compression testing on skin-tissue sample was conducted. Engineering stress and strain values were calculated from the collected data. These values obtained were then used in the material definition module in Abaqus and the stability of the model was analyzed. The results predicted second and third order Ogden model, coupled with time-dependent viscoelastic parameters, to be the most stable within the range of strain applied during the indentation testing. The third order Ogden model gave a better qualitative fit to the experimental data as shown in Fig. 5.12.
For further validation of the material model, a uniaxial unconfined compression simulation was setup in Abaqus to compare to experimental results obtained during testing. The geometry used in the unconfined compression simulation is shown in Fig. 5.13.

Figure 5.12: Material model calibration in Abaqus

Figure 5.13: Simulation of unconfined compression experiment
Figure 5.14 compares the simulation results to those from experimental testing.

![Unconfined compression results comparison with experimental results](image)

The two results compare very well. There is a slight deviation of the results towards the end of compression. This could be because of the irregular geometry of the surface being tested.

### 5.3.2 Indentation

#### 5.3.2.1 Finger model

As mentioned in Section 4.2, failure of the indentation simulation of the finger assembly could have been caused by the following.

- **Geometric irregularity**: The uneven profile of the human tissue and bone surface made the meshing of the components a very complex process. The irregular geometry caused the aspect ratio of the meshing elements to be inconsistent throughout the model. The severe aspect ratio of the elements and the change in element shapes due to contact deformation can cause the solution to diverge.
- **Geometric and contact nonlinearity**: The surface of the human finger is indented using a cylindrical indentor with flat contact surface. When contact with the skin is initiated by the indentor, the underlying surface of the tissue is deformed. The geometry of the components changes significantly due to contact. This is primarily caused during large deformations (5% or more of the total thickness of the surface). This coupled with the irregularity in geometry during contact may cause the solution to diverge and eventually fail.

- **Material nonlinearity**: The response of the material during the process of indentation is nonlinear. The behavior of the material depends on the current loading step as well as the history of loading in the previous time steps. This could exacerbate the convergence problems.

### 5.3.2.2 Cylindrical model assembly

A simpler cylindrical geometry replicating the human finger model and layers was designed to reduce the geometric irregularity of the distal pad of human finger model. This is shown in Fig. 5.15. This model was assembled in Abaqus using same size (0.15 mm) eight node hexahedral elements (C3D8H). The boundary and contact conditions were kept the same as in the distal pad indentation simulation setup. An indentation simulation using the same size indentor was then setup in Abaqus on the cylinder model. As expected, simplifying the geometry helped reduce the effects of geometric nonlinearity in the structure and the solution converged and ran successfully for the process of indentation. The force-displacement response from the indentation simulation is shown in Fig. 5.16.
The figure compares the results obtained from the simulation on cylindrical representation of the finger and experimental results from actual human cadaver fingers. The simulation starts diverging from experiment around 0.3 mm. The solution successfully runs to completion but there is a difference in the peak load obtained at maximum displacement.

Figure 5.16: Load-displacement response from indentation simulation of cylindrical model assembly and indentation testing

Changing the geometry to a simpler representation of a human finger distal pad in
Abaqus and by using smaller time increments, the simulation successfully ran to completion. Simplifying the geometry reduced the contact and geometric non-linearity effects as mentioned earlier. The results shown in the Fig. 5.16 did not match those from the indentation experiment conducted on human cadaver fingers. The response was less stiff than the response of the distal pad during indentation. The difference in the peak load obtained for 1 mm of indentation of the surface came to be approximately 52%.

John Wu et al [28] conducted uniaxial compression to determine the hyperelastic parameter values for skin and soft tissue. They also conducted stress relaxation test on the skin and tissue components to determine the viscoelastic parameters. These values are shown in Tables 5.1 and 5.2. Table 5.3 shows the material parameters extracted using the experimental data obtained from our uniaxial unconfined compression testing. The values of relaxation parameter and time constant for the three term prony series function were obtained from a viscoelastic fitting template [12].

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Table 5.1: Material parameters obtained for skin from John Wu et al [28]
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Table 5.2: Material parameters obtained for subcutaneous tissue from John Wu et al [28]

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Table 5.3: Ogden model parameters determined from our unconfined compression experiments
Figure 5.17: Load-displacement response of indentation simulation: comparison of simulation results with results from the literature

An indentation simulation in Abaqus was then run with material parameters used in published research [28] as shown above. The results were compared and Fig. 5.17 shows the comparison using different material models.

As shown in Fig. 5.17, when using different material models from published research for indentation simulation in Abaqus, the results were not in good agreement with our experimental results. The peak load difference at 1 mm indentation was approximately 52%. Using material parameters from John Wu et al [28] published research for multi layered skin-tissue model in Abaqus and comparing the two simulations showed a difference of approximately 21% at peak load of 1 mm indentation. The nature of the response was very close for the two abaqus analysis results. The response from the published result was approximately 68% less than the peak load at 1 mm indentation experimental result.
5.3.2.3 Cylinder Model

- **Condition or nature of human finger:**

  The difference in the results obtained could be credited to the condition of the human cadaver finger obtained from the anatomic donations lab at the University of Cincinnati. The fingers obtained were preconditioned as mentioned in Chapter 3 by keeping it in PBS prior to conducting the experiments. The nature of the tissue and skin were deteriorated as compared to the geometry file (STL) files obtained through CT scan models. There was visible inconsistency in the nature of skin. The force recorded for displacement of the indentor from the human finger was relatively high as compared to the finite element model created in Abaqus software. The deterioration could result from loss of structural and mechanical integrity of the cadaver human fingers. The difference in the condition of two index fingers obtained from different cadavers can be clearly noticed in Fig. 5.18. Figure 5.19 shows the difference in response of two index fingers extracted from human cadaver bodies. It is clear that the peak load obtained for 1 mm of indentation of the two fingers vary considerably. The peak load result obtained from the index finger of a 42 year old male cadaver was almost 50% less than that obtained from a 47 year old male cadaver.
Figure 5.18: Finger obtained from 42 year old male (left) and 47 year old male (right)

Figure 5.19: Difference between the response of two different male index fingers for 1 mm indentation

The combination of above mentioned nonlinearities and the loss of structural and mechanical integrity in the finger could be responsible for the solution to be different from those obtained through Abaqus and the experimental results.
Chapter 6

Conclusions and Future Directions

In this work, experimental and computational studies were carried out to understand the mechanical response of human fingers. Three types of experiments were conducted: indentation, confined compression, and unconfined compression. Finite element models of the index finger and a representative cylindrical domain were also developed.

Material parameters used in Abaqus were extracted from the unconfined compression experimental results. Due to irregularity and non-linearity in the geometry and material of the model, the simulation analysis didn’t converge when using distal pad of a finger model in Abaqus. There were multiple factors influencing the simulation causing the failure. The contact nonlinearity between the skin-tissue and tissue-bone seemed to be a major contributing factor for the failure of the simulation. Material nonlinearity was also determined to be a factor affecting the simulation failure as the parameters extracted from the experiment did not translate to the simulation in Abaqus. The material parameters used for the simulating the indentation experiment in Abaqus did not accurately represent the behavior of human finger during actual indentation process. The condition of human cadaver fingers, as discussed previously, seemed to be one of the major reasons for the inconsistency and inaccuracy of the results as compared to the experimental re-
results of indentation testing.

Running the simulation with a simpler model reduced the geometric and contact nonlinearity issues and led to the convergence of the simulation. The inaccuracy of the results could be because of the deterioration of the tissue and skin material. As mentioned above, the mechanical strength of the tissue and structural integrity of the components from a cadaver body due to storage conditions also accounted for the variation in results obtained.

In the future, using more accurate material models to simulate the response of the tissue could give more reliable results and also eliminate the possibility of divergence of the simulation. Also, using a simpler cylindrical model with more accurate material parameters could help in predicting the accurate behavior of the human soft tissue under compression.

Furthermore individual experimental tests could be conducted on different components of human finger (skin, tissue and bone) to more accurately predict the behavior and the material parameters of the human finger assembly. Indentation testing could also be performed on actual human subject fingers to obtain more accurate prediction of the behavior of human soft tissue in distal pads of the index finger and thumb.

In order to move this research further, a full finite element model of human index finger and thumb and then the full human hand should be developed to characterize the behavior of full human hand used for various day-to-day applications.
Bibliography


