The Contribution of the Individual Rib to Thoracic Response Under Dynamic Loading Conditions: A Preliminary Hierarchical Approach

THESIS

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Abstract

Rib fractures are still prevalent in motor vehicle crashes and a leading cause of morbidity and mortality. A large body of work has been undertaken to obtain thoracic and individual rib properties, but such testing has primarily focused on 50th percentile males and relies heavily on scaling to apply findings to other populations. Component level testing (i.e., individual bones) has the advantage of capturing large amounts of variation in subject level characteristics (sex, age, stature, etc.). To this end, 318 individual mid-level ribs from 168 post-mortem human subjects (4-108 years) were tested in a dynamic bending scenario simulating a frontal impact to the thorax. Although these data have allowed for an extensive exploration of variation in response of the rib, a gap remains in the ability to understand these findings in the context of the intact thorax. To address this, a series of non-injurious frontal impacts (<20% chest compression) were conducted on three post-mortem human subjects. Each subject was tested in four sequential tissue states: intact, intact with upper limbs removed, denuded (superficial tissue removed), and eviscerated (superficial tissue and viscera removed). Force and deflection data were used to evaluate differences between the tissue conditions. Mid-level ribs were removed and tested to failure in the
dynamic bending scenario previously described. Preliminary data presented here reveal that denuded thoraces retain 70% of the intact peak force and 73% of the intact stiffness, and the eviscerated thoraces retain 54% of peak force and 59% of the intact stiffness. Furthermore, the application of a model in which each rib is treated as a spring acting in parallel was developed in order to use individual rib response data to predict thoracic response. Initial analyses show the model has potential to predict eviscerated peak force and stiffness from cumulative rib response data. The ultimate goal is to develop a transfer function which utilizes the response of the individual rib to predict the response of the thorax from which it came, allowing for the generation of estimated thoracic response data for all populations. These data could be used to improve thoracic response targets and help assess the biofidelity of current anthropomorphic test devices.
Dedication

This thesis is dedicated to my mom for teaching me how to read and playing the alphabet game with me every night before I went to bed, and my dad for teaching me about every tool in his toolbox, especially the screwdriver.
Acknowledgements

I would like to thank Dr. Amanda Agnew for being my advisor and mentor for the past four years. Dr. Agnew’s direction and leadership have helped me to become the best researcher I can be and instilled me a perpetual thirst for knowledge. Her instruction and guidance for testing with post-mortem human subjects has been essential to this research. Her quest to understand human variation and provide car crash protection to all demographics has provided a tremendous amount of opportunities. The skills and knowledge I have gained in my time in the Injury Biomechanics Research Center are invaluable and will stay with me throughout my professional career.

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Table of Contents

Abstract .................................................................................................................................................. ii
Dedication ........................................................................................................................................ iv
Acknowledgments ............................................................................................................................ v
Vita ...................................................................................................................................................... viii
List of Tables ...................................................................................................................................... xiii
List of Figures ...................................................................................................................................... xiv
Chapter 1: Introduction .................................................................................................................... 1
  1.1 Thoracic Injury Risk .................................................................................................................. 1
    1.1.1 Thorax Anatomy ..................................................................................................................... 1
    1.1.2 Thorax Injuries ...................................................................................................................... 3
    1.1.3 Thorax Response ..................................................................................................................... 5
  1.2 Component Level Testing .......................................................................................................... 8
  1.3 Relationship Between the Rib and Thorax .............................................................................. 10
  1.4 Project Goals ............................................................................................................................ 11
<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.2.1 Displacement Time Histories</td>
<td>32</td>
</tr>
<tr>
<td>3.2.2 Force Time Histories</td>
<td>35</td>
</tr>
<tr>
<td>3.2.3 Force-Deflection Curves and Stiffness</td>
<td>37</td>
</tr>
<tr>
<td>3.3 Cumulative Rib Model</td>
<td>40</td>
</tr>
<tr>
<td>Chapter 4: Discussion &amp; Conclusions</td>
<td>46</td>
</tr>
<tr>
<td>4.1 Thoracic Hierarchy Testing</td>
<td>46</td>
</tr>
<tr>
<td>4.2 Individual Rib Testing</td>
<td>49</td>
</tr>
<tr>
<td>4.3 Cumulative Rib Model</td>
<td>51</td>
</tr>
<tr>
<td>4.4 Conclusions</td>
<td>53</td>
</tr>
<tr>
<td>References</td>
<td>54</td>
</tr>
</tbody>
</table>
List of Tables

Table 1. Subject Demographics.................................................................16
Table 2. Equations for equivalent deflection, force, and stiffness of a parallel spring model.................................................................22
Table 3. Peak displacement values from thoracic hierarchy testing.......................26
Table 4. Peak force values and force fractions from thoracic hierarchy testing......28
Table 5. Stiffness values and stiffness fractions from thoracic hierarchy testing....31
Table 6: Peak displacement values from individual rib testing........................34
Table 7: Initial span length and peak percent displacement values from individual rib testing.................................................................34
Table 8: Peak force values from individual rib testing....................................37
Table 9: Stiffness values from individual rib testing........................................39
Table 10. Peak displacement of the cumulative rib models................................42
Table 11. Peak force and force fractions of the cumulative rib models..............42
Table 12. Stiffness values and stiffness fractions of the cumulative rib models.....45
List of Figures

Figure 1. Experimental set-up for thoracic hierarchy testing..............................13

Figure 2. Experimental set-up for thoracic hierarchy testing
   in the four tissue states: intact (gray), intact without
   upper limbs (blue), denuded (green) and eviscerated
   (pink). Each image is at initial contact of the
   impactor face to the thorax.............................................................................15

Figure 3. Subject A (left), Subject B (middle), and
   Subject C (right) seated in the experimental test set-up.................................17

Figure 4. Individual rib testing fixture with a model rib........................................20

Figure 5. Mechanical spring models of a series and parallel system.......................21

Figure 6. Displacement time histories from thoracic hierarchy
   testing of Subject A (top), B (center), and C (bottom).................................25

Figure 7. Force time histories from thoracic hierarchy testing
   of Subject A (top), B (center), and C (bottom)..............................................27

Figure 8. Force-displacement curves with the linear stiffness
   (dashed lines) from thoracic hierarchy testing of
   Subject A (top), B (center), and C (bottom).................................................30
Figure 9. Displacement time histories from individual rib testing for Subject A (top), B (center), and C (bottom). Data are cut at time of fracture, determined by strain gage output. 33

Figure 10. Force time histories from individual rib testing for Subject A (top), B (center), and C (bottom). Data are cut at time of fracture, determined by strain gage output. 36

Figure 11. Force-displacement curves from individual rib testing for Subject A (top), B (center), and C (bottom). Data are cut at time of fracture, determined by strain gage output. 38

Figure 12. Force-displacement curves from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom) with the cumulative rib models included. 41

Figure 13. Force-displacement curves from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom) with the cumulative rib models and linear fits for stiffness included. 42
1.1 Thoracic Injury Risk

The human thorax is commonly injured in motor vehicle crashes (MVCs) due to the injurious, high-rate loads that can be placed on the chest (Lien et al. 2009; Pattimore et al. 1992; Kent et al. 2005). These loads often result in rib fractures, which are considered indicators of trauma, because the greater the number of fractured ribs, the higher the morbidity and mortality rates. Although great strides have been made in improving frontal crash protection of the thorax, rib fractures are still prevalent and a leading cause of death in MVCs (Lee et al. 2015).

1.1.1 Thorax Anatomy

The thoracic skeleton is an anatomical structure comprised of the sternum anteriorly, ribs, and the thoracic vertebral column posteriorly. The 12 thoracic vertebrae support the sternum through 7 bilateral pairs of true ribs that articulate directly with the sternum via their costal cartilage and 3 pairs of false ribs that articulate with the sternum through the costal cartilage of the ribs above them. The final 2 pairs of ribs are floating ribs that do not connect to the sternum. The bony thorax has been studied extensively due to its role in the protection of vital
organs, with the mid-level ribs being of particular interest as they are typically the most commonly injured during MVCs (Lee et al. 2015).

The vital organs of the thorax, commonly referred to as thoracic viscera, consist of the right and left lungs laterally, with the central space being occupied by the mediastinum. The mediastinum extends anteriorly from the sternum to the thoracic spine posteriorly and is surrounded by connective tissue. Inside the mediastinum are the remaining thoracic viscera, which includes the heart, great vessels of the heart, esophagus, trachea, thymus, cardiac nerves, and phrenic nerves. The thoracic viscera function together to allow humans to breathe and circulate blood, which shows why it is important to protect these structures during a MVC.

Superficial to the bony thorax are several layers of muscles, which include, pectoralis major, pectoralis minor, serratus anterior, latissimus dorsi, trapezius, rhomboid major, rhomboid minor, and the erector spinae muscles. Just superficial to the muscles is the subcutaneous tissue, which lies deep to the skin. The skin, subcutaneous tissue, musculature, bony thorax, and viscera each provide a certain amount of resistance to loading that combine to create the intact thorax’s overall resistance to loading. Understanding the contribution of these components will allow for improvements to anthropomorphic test devices (ATDs) and finite element (FE) models of the human thorax.
1.1.2 Thorax Injuries

Despite the many improvements that have been made to injury countermeasures, injuries to the thorax still occur frequently in MVCs. For example, a study conducted by Arajarvi & Santavirta in 1989 reported 93.5% of severely or fatally injured adults utilizing a seatbelt sustained rib fractures. Other safety features such as airbags and force limiting seatbelts work to decrease the number of rib fractures that occur during an MVC (Crandall et al. 1996), but rib fractures have still not been eradicated. Crandall et al. in 2000 found that airbags themselves can cause rib fractures due to stress concentrations in the rib.

In one of the most recent studies on rib fractures seen epidemiologically, Lee et al. found that in a sample of 158 belted and 44 unbelted occupants, a total of 1,371 fractures were documented, resulting in an average of 6.5 fractures per occupant. These data were obtained using the CIREN database, and only focused on occupants that experienced a frontal crash in which the frontal airbag was deployed. Not surprisingly, the authors found that rib fractures in the belted scenarios typically followed the path of the belt and in the unbelted scenarios fractures patterns were more symmetrical due to the distributed loading of the airbag. Although it may seem concerning that the authors found that an injury countermeasure was causing damage, it is important to consider the alternative that would happen without it, which is often fatal.
Another important conclusion from Lee et al. 2015 was that occupant age affected fracture patterns, with fractures occurring more laterally for younger occupants and then moving more anterior as occupant age increased. Several other authors have found that thoracic injury risk is related to increases in age (Kent et al. 2008; Sirmali et al. 2003; Stitzel et al. 2010; Lien et al. 2009; Morgan et al. 1994; Kent & Patrie 2005). However, most of these studies were limited in the number of subjects that could be tested, thus not allowing for an in-depth understanding of variation with age. The epidemiologic studies do capture more variation in age, but lack the controlled laboratory experiments from which strong conclusions can be drawn.

On the other end of the age spectrum is the pediatric population which is a major concern for MVCs. A study conducted by Cooper et al. in 1994 found that 83% of the cases reported to the National Pediatric Trauma Registry were injuries cause by blunt thoracic trauma, with nearly 75% of that being due to MVCs. Thoracic trauma came in second only to head injuries in terms of cause of death. Due to the elasticity of pediatric ribs, rib fractures in children are viewed as hallmarks of severe trauma. Additionally, this elasticity allows for large amounts of chest deflection, which can lead to injuries to the lungs, diaphragm, heart, an even major thoracic blood vessels (Roux & Fisher 1992).
1.1.3 Thorax Response

Despite these populations being identified as vulnerable, most previous research on thoracic response and injury risk has been conducted on 50\textsuperscript{th} percentile males (for height and weight) (Nahum et al. 1971; Nahum et al. 1975; Ramet & Cesari 1979; Sacreste et al. 1982; Cesari & Bouquet 1990; Cavanaugh et al. 1993; Morgan et al. 1994; Yoganandan et al. 1995; Kent et al. 2003; Kent & Patrie 2005; Duma et al. 2006; Kemper et al. 2011). One of the first studies to obtain thoracic response was conducted by Nahum et al. in 1975. This study, in combination with the previous work by Kroell et al. in 1974, served as the basis for the development of the Hybrid III ATD by providing biofidelity corridors for thoracic response created by Neathery in 1974. Additionally, these studies provided a threshold for injury of the thorax, which is considered to be 20\% chest deflection. These studies impacted post mortem human subjects (PMHS) using blunt hub loading, which is useful for obtaining thoracic response, but is potentially not representative of the loading pattern by modern restraints.

To address this, Cesari & Bouquet in 1990 conducted thoracic testing using belt loading on 13 PMHS in the supine position with a fixed back. Two major findings came from this work. The first was that the injuries sustained by the cadavers were similar to those seen in crash scenarios, indicating that the flat back support does not lead to unrealistic loading conditions. Secondly, this study
tested one pediatric subject, a 17-year-old male, that did not sustain any rib fractures, whereas the 12 other, older PMHS all did. This highlights the difference between these populations and the need to establish pediatric specific response corridors for the thorax.

In order to have one-to-one comparisons between hub loading and belt loading, Kent utilized a test matrix consisting of loading with a circular hub, a rectangular hub, a single diagonal belt, and a double diagonal belt. From this testing, it was found that the belted loading scenarios resulted in a stiffer thoracic response, mostly due to the belt interacting with the stiffer anatomical structures of the upper portion of the thorax, i.e. the clavicle and the upper level ribs. The second facet of Kent’s study was that each PMHS was tested in three sequential tissue states: intact, denuded, and eviscerated. This allowed the authors to understand the contribution of each thoracic component to the fully intact thoracic response, which is useful information for FE models, which are being increasingly utilized in the field of injury biomechanics and car crash safety.

Using the vast amount of thoracic data collected, many comparisons have been made between PMHS and ATD response. With respect to the 50th percentile male, several authors have found that the Hybrid III ATD is much stiffer than PMHS, meaning that the ATD’s thorax deflects much less during impacts (Cesari & Bouquet 2010; Kent et al. 2003). This is concerning because ATD designs for
other populations, in particular the pediatric, rely on scaled data since the availability of pediatric response data is limited for ethical reasons (Mertz 1984; Maltese et al. 2010; Parent et al. 2010; Moorhouse 2013; Yoganandan et al. 2014). However, it has been well established that the geometry and material properties of mature bones differ greatly from pediatric, indicating that a child is not simply a scaled down adult (Franklyn et al. 2007).
1.2 Component Level Testing

Although there has been an extensive amount of thoracic testing conducted, the work still does not capture a large amount of human variation, with a specific focus on age, sex, height, and weight. These parameters essentially bin the population into groups for which ATDs are designed and the thoracic response data for these population bins is generated using scaling techniques. While full PMHS testing is limited, component level testing (i.e. testing individual bones) can capture a large amount of human variation and allow for robust investigations of the variation in component response with respect to subject level parameters (age, sex, height, weight, etc.).

With respect to the thorax, the primary load bearing component is the individual ribs, which work in parallel to protect the viscera. Structural properties such as forces, deflections, and stiffness have been obtained by conducting 3-point bending tests on rib segments (Granik & Stein 1973; Stein & Granik 1976; Yoganandan & Pintar 1998; Stitzel et al. 2003; Cormier et al. 2005; Agnew et al. 2013; Kalra et al. 2015) as well as whole rib bending tests (Charpail et al. 2005; Kindig et al. 2011; Agnew et al. 2015; Schafman et al. 2016).

For example, in 1967 Stein & Granik were able to conduct 3-point bending tests on mid-level ribs from 79 PMHS ranging from 27 to 83 years of age. The authors found a weak relationship between the mean failure load with
respect to subject age ($R^2=0.285$), highlighting that the majority of variation in response can be attributed to other sources. Other 3-point bending studies have found that these other sources include, but are not limited to, anatomical location of the specimen, cross-sectional geometry, and subject stature (Yoganandan & Pintar 1998; Stitzel et al. 2003; Cormier et al. 2005; Agnew et al. 2013; Kalra et al. 2015).

The most expansive testing to date was conducted by Schafman et al. in 2016. The authors tested 184 whole mid-level ribs from 93 PMHS ranging from 4 to 99 years of age in a dynamic loading scenario. Similar to the results of Stein & Granik in 1976, the authors found that age, as well as sex, are insufficient in explaining the large amount of variation seen in rib structural response. In a study by Agnew et al. in 2017, which utilized the same testing scenario and a similar sample to Schafman et al., the authors found that stature and weight were also not able to capture the majority of variation seen in individual rib response. However, Murach et al. found that rib geometry was able to explain a large amount of the variation in rib structural response ($R^2$ values ranged from 35-75%). These studies indicate the importance of understanding the source of the variation in rib response, as it is assumed that if thoracic testing were to be conducted on an analogous sample, the thoracic response would reflect the same amount of variation.
1.3 Relationship Between the Rib and Thorax

Although the individual rib testing has provided a vast amount of data, the challenge remains to put these results into the context of the intact thorax under dynamic loading conditions. A study in 2010 by Kindig et al. attempted to solve this problem by applying loads to the individual ribs of an eviscerated thorax. They found that ribs were more coupled bilaterally than by level, with the upper ribs being more coupled than the lower ribs. However, this study was limited to one thorax and the loads were applied quasi-statically, making the results difficult to extrapolate to the dynamic loading environment that is encountered during MVCs. Regardless, the authors proposed a parallel spring model of the human thorax in order to analyze their results, which is a useful tool for understanding how each rib contributes to eviscerated thoracic response (i.e. just the thoracic skeleton).

When utilizing the parallel spring model, the relationship can only be established between the individual rib and bony thorax. In order to continue up the thoracic hierarchy, it is important to understand the contribution of the viscera, superficial tissue, and upper limbs (due to their muscular attachments being in the thoracic region) to the overall intact thoracic response. Although the data from Kent 2008 provide useful information to help solve this problem, the testing of those PMHS did not continue down to the individual rib level.
1.4 Project Goals

The over-arching goal of this project is to be able to use the 318 individual rib tests (from 168 PMHS) that have been conducted in the Injury Biomechanics Research Center at The Ohio State University to generate thoracic response data for populations not commonly tested in full cadaver tests. Specifically, the first objective of this project was to establish a testing methodology in order to understand the contribution of the thoracic components to the intact thoracic response. The second objective of this project was to develop a transfer function which utilizes the response of the individual rib to predict the response of the thorax from which it came.
Chapter 2: Materials & Methods

2.1 Thoracic Hierarchy Testing

2.1.1 Testing Parameters and Instrumentation

In order to understand the contribution of each thoracic component, a series of non-injurious frontal impacts were conducted utilizing a fixed back scenario (Figure 1). Prior to impact, the subject was instrumented with strain gages at 30% and 60% of the rib curve length (measured from the vertebral end) at levels three through eight, which were used to calculate strain rate and detect any fractures that might occur. All tests were conducted at a rate of 3m/s, which corresponds to a strain rate of approximately 0.5 ε/s. This strain rate is analogous to both the strain rates for individual rib testing as well as strain rates of ribs during belt loading, which are the closest simulation of a car crash environment. The fixed back scenario was utilized in order to create a loading environment analogous to that of the individual rib tests (i.e. the vertebral end is fixed translationally, but free to rotate).
Figure 1. Experimental set-up for thoracic hierarchy testing.
Impacts were delivered using a pneumatic ram with a 23kg impactor. The impactor face was a 6” high x 12” wide x 0.5” thick rectangle and was centered vertically and horizontally on the sternum. Chest deflection was measured anteriorly using a linear displacement potentiometer (Celesco CLWG-600-MC4, TE Connectivity Co., Berwyn, PA) attached to the impactor face. Chest depth was measured prior to each impact and the stroke of the ram was limited to <20% chest compression, as this is below the currently accepted threshold for thoracic injury (Kent 2008; Duma et al. 2006; Kemper et al. 2011).

Two 6-axis load cells (Denton 2944JFL, Humanetics, Plymouth, MI) were used to measure forces at the level of impact, with one attached to the impactor face (anterior) and one located behind the back plate (posterior). Both load cells were used in order to eventually compare these data to previous frontal thoracic work i.e., the force in an ATD frontal thorax impact test is typically measured anteriorly and the force in individual rib testing is measured posteriorly. To quantify the effect of all thoracic components, each subject was tested in four sequential tissue states: intact, intact without upper limbs, denuded (superficial tissue removed), and eviscerated (superficial tissue and viscera removed) (Figure 2).
Figure 2. Experimental set-up for thoracic hierarchy testing in the four tissue states: intact (gray), intact without upper limbs (blue), denuded (green) and eviscerated (pink). Each image is at initial contact of the impactor face to the thorax.

Removal of the upper limbs began with disarticulating at the acromioclavicular joint and then the muscular attachments at the scapula. The clavicles were left intact and attached at the sternoclavicular joint. The denuding process involved removing all skin, subcutaneous tissue, pectoralis major, pectoralis minor, serratus anterior, trapezius, latissimus dorsi, rhomboid major, rhomboid minor, and all superficial back muscles. The intercostal muscles and deep back muscles were left intact. Lastly, the eviscerating process required the removal of all thoracic and abdominal viscera. Special care was taken to ensure that the intercostal muscles remained intact during the evisceration process.
2.1.2 Subject Demographics

Three male subjects were tested for this study, the demographics from which can be found in Table 1. Images of each subject in the intact condition in the experimental test set-up can be found in Figure 3. The subjects ages ranged from 55 to 73 years, and their mass and stature were similar to that of the 50th percentile male. All subject aBMD t-scores fell within the normal range.

Table 1. Subject Demographics

<table>
<thead>
<tr>
<th></th>
<th>Subject A</th>
<th>Subject B</th>
<th>Subject C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sex</td>
<td>M</td>
<td>M</td>
<td>M</td>
</tr>
<tr>
<td>Age (years)</td>
<td>73</td>
<td>62</td>
<td>55</td>
</tr>
<tr>
<td>Stature (cm)</td>
<td>170</td>
<td>173</td>
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<tr>
<td>Mass (kg)</td>
<td>62</td>
<td>84</td>
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</tr>
<tr>
<td>BMI (kg/cm$^2$)</td>
<td>21.5</td>
<td>28.3</td>
<td>22.4</td>
</tr>
<tr>
<td>aBMD t-score (lumbar)</td>
<td>0.4</td>
<td>1.6</td>
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</tr>
<tr>
<td>Chest Depth (cm)</td>
<td>20</td>
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</tr>
<tr>
<td>Chest Breadth (cm)</td>
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</tr>
<tr>
<td>Chest Circumference (cm)</td>
<td>90</td>
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</tr>
<tr>
<td>Thoracic Index</td>
<td>0.61</td>
<td>0.71</td>
<td>0.65</td>
</tr>
</tbody>
</table>

*BMI = Body Mass Index (mass/stature$^2$), aBMD = areal bone mineral density as measured by dual-energy X-ray absorptiometry (DXA)
Figure 3. Subject A (left), Subject B (middle), and Subject C (right) seated in the experimental test set-up.

2.1.3 Data Analysis

All data were filtered using a CFC180 filter (SAE 2007). Posterior force data were shifted in order to have the peak of the posterior force occur at the same time as peak displacement. Only the posterior force data were analyzed in detail because the forces are measured posteriorly during the individual rib test. Posterior force data were plotted against displacement data and stiffness was calculated by fitting a line to the linear portion of the curve.

Peak forces and stiffness values were compared back to the intact condition by dividing the value of the property from the test of interest by the property calculated for the intact condition. These values are referred to as the force fraction (FF) and stiffness fraction (SF) throughout this manuscript. The
resulting values can be interpreted as the amount of force or stiffness that each tissue condition retained when compared to the intact condition. The closer these values are to one, the more similar the responses.

2.2 Individual Rib Testing

2.2.1 Testing Parameters and Instrumentation

Following eviscerated testing, bilateral pairs of rib levels 4-7 were removed and tested to failure in a custom-built pendulum fixture (Figure 4).
Details regarding the preparation of specimens, experimental testing, and data analysis can be found in Agnew et al. 2015 and Schafman et al. 2016. The experiment simulated a frontal impact to the thorax in which the sternal (anterior) end of the rib was linearly translated toward the vertebral (posterior) end until the point of fracture. A 54.4kg pendulum impacted ribs at 2m/s, resulting in an approximate strain rate of 0.5/s. Displacement of the sternal end of the rib was measured by a linear string potentiometer (Rayelco P-20A, AMETEK, Inc. Berwyn, PA), and forces were recorded by a 6-axis load cell (CRABI neck load cell, IF-954, Humanetics, Plymouth, MI) located posterior to the rib. The strain gages applied for thoracic testing remained in place for rib testing.

Figure 4. Individual rib testing fixture with a model rib.
2.2.2 Data Analysis

All data were filtered using a CFC180 filter (SAE 2007). Force-displacement curves were generated by plotting the displacement of each individual rib test against the force up to the point of fracture. Linear structural stiffness of each rib was defined as the slope of the force-displacement curve for 20-80% of yield. A cumulative rolling mean and standard deviation method was used to determine the point of yield, the details of which can be found in Agnew et al. 2015.

2.3 Cumulative Rib Model

2.3.1 Theory

In mechanics, springs can be used to model systems when information is known about the components in order to determine the response of the overall system. Two or more springs are said to be acting in series when they are connected from end-to-end and in parallel when they are connected side-by-side (Figure 5).
The bony thorax can be thought of as a parallel spring system, as all the ribs are connected side-by-side, especially when the thorax is loaded anteriorly (Kindig et al. 2010). Table 2 contains the equations for calculating the equivalent deflection, force, and stiffness (i.e. the spring constant) for a parallel spring system.

Table 2. Equations for equivalent deflection, force, and stiffness of a parallel spring model

<table>
<thead>
<tr>
<th>Quantity</th>
<th>Equation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deflection (d)</td>
<td>$d_{tot} = d_1 = d_2 = \ldots = d_n$</td>
</tr>
<tr>
<td>Force (F)</td>
<td>$F_{tot} = F_1 + F_2 + \ldots + F_n$</td>
</tr>
<tr>
<td>Stiffness (k)</td>
<td>$k_{tot} = k_1 + k_2 + \ldots + k_n$</td>
</tr>
</tbody>
</table>

2.3.2 Application of Parallel Spring Model to Individual Rib Data

The parallel spring model was applied to individual rib data in order to calculate a cumulative response, called the cumulative rib model. To ensure the
data from each rib test were summed and averaged appropriately, a generic displacement vector was determined, and then force data were obtained via interpolation. Because of the nature of the model, all individual rib data had to be truncated according to the rib with the shortest time to failure. The cumulative deflection was then calculated by averaging the displacement across the individual rib tests, because the deflection in a parallel spring model should be equivalent across all springs (Table 2). The cumulative force was calculated by summing the forces in the primary loading direction (X) from the eight individual rib tests from each subject.

Additionally, the angle of the rib during thoracic hierarchy testing was taken into consideration. Rib angle was measured by estimating a line along the rib length on a 2D lateral image of the seated subject in the eviscerated condition, and then measuring the angle of that line with respect to horizontal. The cumulative displacement and force data were then multiplied by the cosine of the average rib angle of each subject. The cumulative rib model with and without the rib angle incorporated will each be analyzed.

Similar to the thoracic hierarchy data analysis, the peak force of the cumulative rib model was defined at the maximum force. Stiffness of the cumulative rib model was calculated by fitting a line to the linear portion of the curve. However, peak forces and stiffness values were compared back to the
eviscerated condition by dividing the value of the property from the cumulative rib model by the property calculated for the eviscerated condition. These values can be interpreted as the ability of the cumulative rib model to represent the bony thorax. The closer these values are to one, the more similar the responses.
Chapter 3: Results

3.1 Thoracic Hierarchy

3.1.1 Displacement Time Histories

All plots of data from thoracic hierarchy testing will have results from the intact condition in gray, intact without upper limbs (UL) condition in blue, denuded condition in green, and eviscerated condition in pink. Time histories for displacements recorded at impact location for all tissue states of Subjects A-C can be found in Figure 6, and peak displacement values are shown in Table 3. The slight variation in peak displacement is due to 20% chest deflection being calculated using the chest depth of each tissue state as well as a small amount of error (approximately 1mm i.e. 2%) in the system used to stop the pneumatic ram.
Figure 6. Displacement time histories from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom).
Table 3. Peak displacement values from thoracic hierarchy testing

<table>
<thead>
<tr>
<th>Thorax Condition</th>
<th>Peak Displacement (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Subject A</td>
</tr>
<tr>
<td>Intact</td>
<td>51.2</td>
</tr>
<tr>
<td>Intact without Upper Limbs</td>
<td>50.0</td>
</tr>
<tr>
<td>Denuded</td>
<td>49.2</td>
</tr>
<tr>
<td>Eviscerated</td>
<td>49.6</td>
</tr>
</tbody>
</table>

3.1.2 Force Time Histories

Time histories for forces recorded by the posterior load cells for all tissue states of Subjects A-C can be found in Figure 7, and peak force values are shown in Table 4 along with the force fractions for each condition.

It should be noted that Subject B was found to have incomplete (i.e. not through both cortices) rib fractures located near the costochondral junction at autopsy. Strain gage data revealed that these fractures occurred during the intact impact (baseline). Since these fractures occurred in the first impact, subsequent responses and relative differences with respect to that initial impact configuration were still compared within Subject B. Subject A and C did not experience any rib fractures.
Figure 7. Force time histories from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom).
Table 4. Peak force values and force fractions from thoracic hierarchy testing

<table>
<thead>
<tr>
<th>Subject</th>
<th>Posterior Load Cell Peak Force (N)</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Intact Force (baseline)</td>
<td>Intact without UL</td>
<td>Denuded</td>
<td>Eviscerated</td>
</tr>
<tr>
<td></td>
<td>Force</td>
<td>Force Fraction</td>
<td>Force</td>
<td>Force Fraction</td>
</tr>
<tr>
<td>A</td>
<td>1267.2</td>
<td>1069.4</td>
<td>0.84</td>
<td>1035.7</td>
</tr>
<tr>
<td>B</td>
<td>1030.2</td>
<td>800.9</td>
<td>0.73</td>
<td>557.1</td>
</tr>
<tr>
<td>C</td>
<td>1221.3</td>
<td>1191.9</td>
<td>0.95</td>
<td>962.1</td>
</tr>
<tr>
<td>Average</td>
<td>1172.9</td>
<td>1020.7</td>
<td>0.84</td>
<td>851.6</td>
</tr>
<tr>
<td>St. Dev.</td>
<td>102.6</td>
<td>163.3</td>
<td>0.09</td>
<td>210.5</td>
</tr>
</tbody>
</table>

Peak forces decrease within each subject as tissue is removed. It appears that the removal of the upper limbs did influence the posterior force data (average force fraction = 0.84), likely due to the removal of mass that was going into the fixed back plate. On average the denuding procedure resulted in the thorax retaining 70% of its intact force response and the eviscerating procedure resulted in the thorax retaining only 54% of its intact force response.
3.1.3 Force-Deflection Curves and Stiffness

Force-deflection curves were generated by plotting the displacement data against the posterior force data and can be found in Figure 8. Each plot in Figure 8 also shows the linear fit used to calculate stiffness (dashed lines). The stiffness values can be found in Table 3 along with the stiffness fractions of each condition. For stiffness, the denuded thoraces retained 73% and the eviscerated thoraces retained 59% of their intact stiffness.
Figure 8. Force-displacement curves with the linear stiffness (dashed lines) from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom).
<table>
<thead>
<tr>
<th>Subject</th>
<th>Stiffness (N/mm)</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Intact</td>
<td>Intact without UL</td>
<td>Denuded</td>
<td>Eviscerated</td>
</tr>
<tr>
<td></td>
<td>Stiffness (baseline)</td>
<td>Stiffness</td>
<td>Stiffness</td>
<td>Stiffness</td>
</tr>
<tr>
<td>A</td>
<td>45.6</td>
<td>39.2</td>
<td>0.86</td>
<td>38.8</td>
</tr>
<tr>
<td>B</td>
<td>32.1</td>
<td>23.9</td>
<td>0.75</td>
<td>18.4</td>
</tr>
<tr>
<td>C</td>
<td>38.3</td>
<td>37.2</td>
<td>0.97</td>
<td>29.4</td>
</tr>
<tr>
<td>Average</td>
<td>38.6</td>
<td>33.4</td>
<td>0.86</td>
<td>28.8</td>
</tr>
<tr>
<td>St. Dev.</td>
<td>5.5</td>
<td>6.8</td>
<td>0.09</td>
<td>8.3</td>
</tr>
</tbody>
</table>

Table 5. Stiffness values and stiffness fractions from thoracic hierarchy testing
3.2 Individual Rib Testing

3.2.1 Displacement Time Histories

Time histories for the displacement data from each individual rib test from Subjects A-C can be found in Figure 9. All individual rib data are cut at the time of fracture, determined by the strain gage data. Peak displacement, initial span length (i.e. the linear distance from sternal to vertebral end of the rib) and peak percent displacement values can be found in Table 6 and Table 7. The displacement time histories qualitatively reveal that the loading rate for individual rib testing is very consistent. However, there do not appear to be any trends for any of the displacement data with respect to rib level. Peak percent displacements of the individual ribs ranged from 12.0 to 27.3%.
Figure 9. Displacement time histories from individual rib testing for Subject A (top), B (center), and C (bottom). Data are cut at time of fracture, determined by strain gage output.
Table 6: Peak displacement values from individual rib testing

<table>
<thead>
<tr>
<th>Ride Side &amp; Level</th>
<th>Peak Displacement (mm)</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Subject A</td>
<td>Subject B</td>
<td>Subject C</td>
<td></td>
</tr>
<tr>
<td>L4</td>
<td>44.4</td>
<td>25.8</td>
<td>52.0</td>
<td></td>
</tr>
<tr>
<td>R4</td>
<td>42.3</td>
<td>21.7</td>
<td>56.8</td>
<td></td>
</tr>
<tr>
<td>L5</td>
<td>35.4</td>
<td>29.1</td>
<td>70.5</td>
<td></td>
</tr>
<tr>
<td>R5</td>
<td>29.4</td>
<td>23.1</td>
<td>53.1</td>
<td></td>
</tr>
<tr>
<td>L6</td>
<td>33.8</td>
<td>23.6</td>
<td>59.2</td>
<td></td>
</tr>
<tr>
<td>R6</td>
<td>47.9</td>
<td>25.8</td>
<td>47.2</td>
<td></td>
</tr>
<tr>
<td>L7</td>
<td>50.6</td>
<td>21.5</td>
<td>59.2</td>
<td></td>
</tr>
<tr>
<td>R7</td>
<td>39.5</td>
<td>26.4</td>
<td>65.9</td>
<td></td>
</tr>
</tbody>
</table>

Table 7: Initial span length and peak percent displacement values from individual rib testing

<table>
<thead>
<tr>
<th>Ride Side &amp; Level</th>
<th>Initial Span Length (mm)</th>
<th>Peak Percent Displacement (%)</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Subject A</td>
<td>Subject B</td>
<td>Subject C</td>
<td>Subject A</td>
<td>Subject B</td>
</tr>
<tr>
<td>L4</td>
<td>190.0</td>
<td>176.0</td>
<td>218.0</td>
<td>23.4</td>
<td>14.7</td>
</tr>
<tr>
<td>R4</td>
<td>182.0</td>
<td>177.0</td>
<td>208.0</td>
<td>23.2</td>
<td>12.3</td>
</tr>
<tr>
<td>L5</td>
<td>210.0</td>
<td>184.0</td>
<td>226.0</td>
<td>16.9</td>
<td>15.8</td>
</tr>
<tr>
<td>R5</td>
<td>206.0</td>
<td>191.0</td>
<td>220.0</td>
<td>14.3</td>
<td>12.1</td>
</tr>
<tr>
<td>L6</td>
<td>226.0</td>
<td>207.0</td>
<td>233.0</td>
<td>15.0</td>
<td>11.4</td>
</tr>
<tr>
<td>R6</td>
<td>222.0</td>
<td>203.0</td>
<td>239.0</td>
<td>21.6</td>
<td>12.7</td>
</tr>
<tr>
<td>L7</td>
<td>233.0</td>
<td>221.0</td>
<td>261.0</td>
<td>21.7</td>
<td>9.7</td>
</tr>
<tr>
<td>R7</td>
<td>238.0</td>
<td>220.0</td>
<td>256.0</td>
<td>16.6</td>
<td>12.0</td>
</tr>
</tbody>
</table>
3.2.2 Force Time Histories

Time histories for force from each individual rib test can be found in Figure 10, and peak force values are shown in Table 8. Both the plots and tables show that there is a fair amount of bilateral symmetry and peak forces typically increase as rib level increase (i.e. moving down the rib cage).
Figure 10. Force time histories from individual rib testing for Subject A (top), B (center), and C (bottom). Data are cut at time of fracture, determined by strain gage output.
### Table 8: Peak force values from individual rib testing

<table>
<thead>
<tr>
<th>Ride Side &amp; Level</th>
<th>Peak Force (N)</th>
<th>Subject A</th>
<th>Subject B</th>
<th>Subject C</th>
</tr>
</thead>
<tbody>
<tr>
<td>L4</td>
<td>100.6</td>
<td>53.9</td>
<td>81.1</td>
<td></td>
</tr>
<tr>
<td>R4</td>
<td>89.0</td>
<td>46.4</td>
<td>76.0</td>
<td></td>
</tr>
<tr>
<td>L5</td>
<td>116.0</td>
<td>74.7</td>
<td>84.7</td>
<td></td>
</tr>
<tr>
<td>R5</td>
<td>106.6</td>
<td>75.8</td>
<td>99.7</td>
<td></td>
</tr>
<tr>
<td>L6</td>
<td>121.6</td>
<td>118.3</td>
<td>158.6</td>
<td></td>
</tr>
<tr>
<td>R6</td>
<td>138.2</td>
<td>101.0</td>
<td>128.4</td>
<td></td>
</tr>
<tr>
<td>L7</td>
<td>122.5</td>
<td>111.3</td>
<td>123.2</td>
<td></td>
</tr>
<tr>
<td>R7</td>
<td>134.4</td>
<td>107.1</td>
<td>144.4</td>
<td></td>
</tr>
</tbody>
</table>

#### 3.2.3 Force-Deflection Curves and Stiffness

Force-displacement curves of each individual rib test can be found in Figure 11 and the linear structural stiffness values can be found in Table 9. Similar to peak force and peak stiffness, there is a fair amount of bilateral symmetry in the stiffness values and stiffness tends to increase with increasing rib level. However, the 6th level ribs were typically the stiffest for each subject.
Figure 11. Force-displacement curves from individual rib testing for Subject A (top), B (center), and C (bottom). Data are cut at time of fracture, determined by strain gage output.
Table 9: Stiffness values from individual rib testing

<table>
<thead>
<tr>
<th>Side &amp; Level</th>
<th>Stiffness (N/mm)</th>
<th>Subject A</th>
<th>Subject B</th>
<th>Subject C</th>
</tr>
</thead>
<tbody>
<tr>
<td>L4</td>
<td>2.73</td>
<td>2.23</td>
<td>1.95</td>
<td></td>
</tr>
<tr>
<td>R4</td>
<td>3.01</td>
<td>1.66</td>
<td>2.17</td>
<td></td>
</tr>
<tr>
<td>L5</td>
<td>4.47</td>
<td>3.02</td>
<td>2.25</td>
<td></td>
</tr>
<tr>
<td>R5</td>
<td>4.02</td>
<td>2.71</td>
<td>2.42</td>
<td></td>
</tr>
<tr>
<td>L6</td>
<td>5.19</td>
<td>3.57</td>
<td>3.71</td>
<td></td>
</tr>
<tr>
<td>R6</td>
<td>5.32</td>
<td>4.62</td>
<td>4.43</td>
<td></td>
</tr>
<tr>
<td>L7</td>
<td>4.96</td>
<td>3.97</td>
<td>3.40</td>
<td></td>
</tr>
<tr>
<td>R7</td>
<td>4.77</td>
<td>5.18</td>
<td>3.12</td>
<td></td>
</tr>
</tbody>
</table>
3.3 Cumulative Rib Model

The force-deflection responses generated from the individual rib data using the cumulative rib model can be found in Figure 12. Both the baseline model (solid line) and the model with the rib angle incorporated (dashed line) are shown. The average angles of ribs 4-7 for Subject A, B, and C were 33.6°, 33.3°, and 46.3°, respectively. Due to the nature of the model, all individual rib data had to be truncated according to the rib with the shortest time to failure (i.e. smallest displacement). For the baseline model, the peak displacements ranges from 21.5 to 47.21mm (Table 10) and peak forces of 597.7 to 812.2N (Table 11). For the model including the rib angle, the peak displacements ranges from 18.0 to 32.6mm (Table 9) and peak forces of 499.6 to 676.5N (Table 10). Force fractions, which are calculated by comparing each rib model back to the eviscerated condition, can also be found in Table 10. The cumulative rib model that did incorporate the rib angle performed better than the model that incorporated the rib angle, with a force fraction of 0.93.
Figure 12. Force-displacement curves from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom) with the cumulative rib models included.
Table 10. Peak displacement of the cumulative rib models

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak Displacement (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Cumulative Rib</td>
</tr>
<tr>
<td>A</td>
<td>29.4</td>
</tr>
<tr>
<td>B</td>
<td>21.5</td>
</tr>
<tr>
<td>C</td>
<td>47.2</td>
</tr>
</tbody>
</table>

Table 11. Peak force and force fractions of the cumulative rib models

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Eviscerated</td>
</tr>
<tr>
<td></td>
<td>Force</td>
</tr>
<tr>
<td>A</td>
<td>844.5</td>
</tr>
<tr>
<td>B</td>
<td>473.5</td>
</tr>
<tr>
<td>C</td>
<td>602.5</td>
</tr>
<tr>
<td>Average</td>
<td>640.2</td>
</tr>
<tr>
<td>St. Dev.</td>
<td>153.8</td>
</tr>
</tbody>
</table>
The stiffness of the cumulative rib models was calculated in the same manner that it was calculated for the thoracic hierarchy data i.e. by fitting a line to the linear portion of the curve. The force-deflection responses generated from the individual rib data using the cumulative rib model with the linear fits used or stiffness can be found in Figure 13. The stiffness values and the stiffness fractions calculated by comparing back to the eviscerated condition can be found in Table 12.

Since the force and displacement data were both multiplied by the cosine of the rib angle, the stiffness values that are generated by the linear fit of the force-deflection curve does not differ from the cumulative rib model that does not incorporate the rib angle. Therefore, the stiffness values were multiplied by the cosine of the rib angle and the values can be found in Table 12.
Figure 13. Force-displacement curves from thoracic hierarchy testing of Subject A (top), B (center), and C (bottom) with the cumulative rib models and linear fits for stiffness included.
Table 12. Stiffness values and stiffness fractions of the cumulative rib models

<table>
<thead>
<tr>
<th>Subject</th>
<th>Eviscerated Stiffness</th>
<th>Cumulative Rib Stiffness</th>
<th>Cumulative Rib with Rib Angle Stiffness</th>
<th>Stiffness Fraction</th>
<th>Stiffness Fraction</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>35.54</td>
<td>33.58</td>
<td>27.97</td>
<td>0.94</td>
<td>0.79</td>
</tr>
<tr>
<td>B</td>
<td>14.68</td>
<td>26.28</td>
<td>21.97</td>
<td>1.79</td>
<td>1.50</td>
</tr>
<tr>
<td>C</td>
<td>20.39</td>
<td>23.90</td>
<td>16.51</td>
<td>1.17</td>
<td>0.81</td>
</tr>
<tr>
<td>Average</td>
<td>23.53</td>
<td>27.92</td>
<td>22.15</td>
<td>1.30</td>
<td>1.03</td>
</tr>
<tr>
<td>St. Dev.</td>
<td>8.80</td>
<td>4.12</td>
<td>4.68</td>
<td>0.36</td>
<td>0.33</td>
</tr>
</tbody>
</table>
Chapter 4: Discussion

4.1 Thoracic Hierarchy

Across all subjects, tissue removal resulted in a decrease in thoracic force and stiffness. On average, the denuded thoraces retained 70% of the intact peak force and the eviscerated thoraces retained 54%. Similarly, the denuded thoraces retained 73% of the intact stiffness and the eviscerated thoraces retained 59%. This is expected because this stiffness value was calculated from curves where the peak force and peak displacement were aligned, making the stiffness calculation similar to simply dividing peak force by peak displacement. Since displacement was controlled to a similar value within subjects, it is not surprising that the stiffness reductions parallel those of the peak force.

The reductions in stiffness are less than those found by Kent 2008, who found that denuded thoraces retained approximately 60% of their intact stiffness (and eviscerated thoraces retained approximately 30% of their intact stiffness by measuring force posteriorly. This could potentially be due to rate differences (Kent 2008 tested at ~1m/s), subject differences (Kent 2008 tested two female subjects), and testing configuration (Kent 2008 tested the subjects laying supine with a material testing system). Additionally, it should be noted that although the removal of the upper limbs did have an effect on the posterior force data (average
force fraction = 0.80), potentially due to the mass of the arms and the interaction of the scapulae with the back plate. Kent 2008 left the upper limbs on for all tissue states, which may have also contributed to the differences found between the studies.

Although the testing was designed to be non-injurious, Subject B experienced rib fractures in L4, L5, R5, and R6 in the intact impact (baseline), confirmed with strain gage data. Since these fractures occurred in the first impact, subsequent responses and relative differences with respect to that initial impact configuration may still be comparable within subject B, especially since fractures were incomplete (i.e., not through both cortices) meaning a connection of the bone remained; however, peak force magnitudes and stiffness values should be utilized with caution because progression of the fracture in each successive test is assumed to be stable throughout the test series. Surprisingly, Subject B had the highest aBMD t-score, which is the current metric for bone quality, indicating that this subject should not have experienced fractures.

The data presented here add to the large culmination of data with respect to the 50th percentile male thoracic response, with detailed information on the contribution of each component of the thorax to the overall response. However, several limitations to this testing methodology remain. Thoracic testing was limited to 20% chest compression to allow for repeated non-injurious impacts,
meaning the response data does not represent a full impact where the ram would
be allowed to displace until all input energy is absorbed. Despite limiting the
compression, fractures still occurred for one subject, which suggests that current
limits for thoracic injury may need to be improved. Interestingly, Kent 2008 used
the same limit on chest deflection and has subjects that sustained fractures.
Additionally, all tests were completed using a fixed back scenario, which is not a
realistic representation of a car crash scenario in which the forward inertia of
occupants plays an important role.

Future work will include more testing to better understand the decreases in
thoracic response that following the removal of tissue, as it is difficult to draw
strong conclusions based on three subjects. Additionally, parameters such as
tissue thickness and mass will be investigated in order to explain variation in the
force and stiffness fractions between subjects. The data collected by the anterior
load cell during thoracic testing will be analyzed and several different methods for
calculating thoracic stiffness will be investigated along with more complex
models of thorax that incorporate damping and inertial terms.
4.2 Individual Rib Testing

The peak percent displacement data range from 12.0 to 27.3% (Table 7). These data are within the range found by Schafman et al. in 2016, however they are on the lower end. Interestingly, several ribs broke at or lower than the 20% compression threshold established by cadaveric thoracic testing literature. Individual rib testing does not factor the angle of the rib in the thorax, which could be why the ribs fracture at these percent displacements. However, a study by Ali et al. in 2005 found that when anterior displacements are applied in a fixed back scenario, the primary deformation of the rib occurs along the length of the rib. Future work will continue to investigate the angle of the rib in the thorax during thoracic testing as an important factor to consider when attempting to place the rib in the context of the intact thorax.

The peak forces and stiffness ranged from 46.4 to 158.6N and 1.66 to 5.32N/mm. Again, both of these properties have values that fall within the range of values reported by Schafman et al. in 2016. Typically the force response increased as rib level increased (i.e. moving down the rib cage). Additionally, there is evidence of bilateral symmetry. Both of these trends are important because the cumulative rib model assumes that all ribs contribute equally. However, Kindig et al. 2010 found that due to the coupling of the ribs along the rib cage, not all ribs contribute equally to the response of the thorax.
Although Subject B did sustain rib fractures during thoracic testing, the fractures occurred far enough anteriorly that they were embedded in the potting material that is used on the sternal and vertebral ends of individual ribs to secure them into the rib testing fixture. Results of rib testing show that peak displacements and forces for subject B are all lower than subject A and C (Figures 8-9) and rib responses appear to lack a plastic response (Figure 10). However, it is difficult to say if these differences are due to rib fracture or simply individual variation. The results do need to be analyzed with a degree of caution because of the rib fractures, but it should be noted that previous individual rib tests have produced results similar to those of Subject B, indicating the results from Subject B could potentially be due to subject-to-subject variation.
4.3 Cumulative Rib Model

Typically, the cumulative rib model that did not incorporate the rib angle over-estimated the peak force of the eviscerated test (average force fraction = 1.18). The incorporation of the rib angle decreased the peak forces of the cumulative rib model and brought the average force fraction down to 0.93. This shows that incorporating the rib angle improved the ability to predict the eviscerated peak force. Nevertheless, both cumulative rib models worked well when compared to posterior load cell forces in thoracic hierarchy testing. This finding is not surprising, as forces are recorded by a posterior load cell during individual rib testing (Figure 1).

Similar to peak force, the cumulative rib model that did not incorporate the rib angle typically over-predicted the stiffness of the eviscerated condition (average stiffness fraction = 1.30). Multiplying each stiffness value by the cosine of the rib angle decreased the stiffness values of the cumulative rib model and resulted in stiffness values much closer to those of the eviscerated condition with an average stiffness fraction of 1.03. Again, this shows that the incorporation of the angle of the ribs during thoracic hierarchy testing improved the ability to predict the response of the eviscerated condition from individual rib testing data.

The cumulative rib models presented here assumed that ribs are fully coupled, meaning that if a displacement was applied to one rib while in the
thorax, the remaining ribs would displace the same amount. However, Kindig et al. 2010 found that although ribs were fully coupled bilaterally, the degree of coupling decreased with rib level (upper ribs were more strongly coupled than lower ribs), which is likely due to varying connections of the ribs to the sternum via costal cartilage. The models presented here do not account for off-axis loading or rotation of the ribs with respect to the spine during thoracic hierarchy testing, both of which are likely important for establishing an accurate model. Alternative cumulative rib models will be explored that incorporate individual rib force, displacement, and strain.
4.4 Conclusions

Preliminary data presented here reveal that denuded thoraces retain 70% of the intact peak force and the eviscerated thoraces retain 54% of peak force. When looking at stiffness, denuded thoraces retain 73% and eviscerated thoraces retain 59% of their intact stiffness. The cumulative rib model shows promise when utilized in context of the posterior load cell force in thoracic testing. These data will be useful for developing a transfer function to allow for the prediction of thoracic response from individual rib data, to allow for the generation of thoracic response data for populations for which full-thoracic impacts are not typically conducted (e.g., pediatric) but individual rib testing can be. These data could be used to improve thoracic response targets and help assess the biofidelity of ATDs of all sizes and demographics.
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