DETERMINATION OF THE
MATERIAL PROPERTIES OF THE PEDIATRIC RIB

A Thesis

Presented in Partial Fulfillment of the Requirements for

the Degree Master of Science in the

Graduate School of The Ohio State University

By

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* * * * *

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Understanding the characteristics of the pediatric chest is of great importance for analyzing pediatric traumatic injury. As indicated by the increase in fatalities when thorax injury is accompanied with head and abdominal injuries, there is a correlation between the behavior of the thorax and the head. The initial motion of the thorax dictates the relative movement of the top of the spine, neck, and head. Also, the deflection of the pediatric rib cage may result in internal injuries to the abdomen and chest cavity without rib fracture. In order to realize the potential for internal injuries without visible rib fracture, the behavior of the pediatric thorax must be determined. The literature contains much data on the adult rib, but properties of the pediatric rib are lacking from previous studies. Due to the extreme differences in growth and development between children and adults, the scaling of adult properties to reflect those of children based strictly on size is not reasonable. Several other factors, including fracture mechanisms and bone development, must be taken into account.

The objective of this study is to determine the material properties of the excised pediatric rib as a function of age and to compare the results to the material properties of
the excised porcine rib. A total of 150 porcine rib samples from 25 porcine subjects, and 31 rib samples from 9 pediatric subjects were loaded in 3-point bending at a quasi-static rate. All data were analyzed for rotation during testing, axis of moment of inertia, stiffness, Young’s modulus, flexural rigidity, peak force, bending strength, yield force, and yield stress.

The results of the study indicate that the rib cross-section undergoes minimal rotation during testing. However, the orientation of the bending axis does not always correspond to the orientation of the principal axis of the minimum moment of inertia. For the porcine subjects, moment of inertia, stiffness, flexural rigidity, peak force, and yield force showed a linear increase with age. The variables of Young’s modulus, bending strength, and yield stress are believed to be constant over all ages of porcine subjects. All variables analyzed for 3-point bending of the pediatric rib showed increase in value as a function of pediatric age.

The major finding of this study is that porcine subjects do not present an acceptable animal model for the behavior of the pediatric rib. Development rates between humans and the pig are vastly different. The conclusion from this study is that additional pediatric rib specimens are needed in order to fully understand the changes in the material properties of the pediatric rib as a function of age.
Dedicated to my parents for always believing in me;

and to my loving fiancé, John, for his never-ending support and for his staying up many
late nights offering suggestions, advice, and comments on this project.
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CHAPTER 1

INTRODUCTION

1.1 Statement of the Problem

Traumatic injury is the leading cause of death among children. Although thoracic injury alone accounts for only 5% of the total number of deaths due to traumatic injury, chest injuries accompanied with head and abdominal injuries account for almost 40% of traumatic injury fatalities (Bliss and Silen, 2002). The increased flexibility of the pediatric ribcage leads to an increased compression and deflection of the pediatric chest compared with that of an adult. As a result, thoracic injuries to children often involve pulmonary contusions and other internal injuries as the rib cage is compressed, even without rib fractures (Bliss and Silen, 2002). Because of the potential for occult internal thoracic injury and because children are frequently incapable of communicating the extent of their injuries, it is pertinent to understand the biomechanical behavior of the pediatric chest. Also, as indicated by the increase in fatalities when thorax injury is accompanied with head and abdominal injuries, there is a correlation between the behavior of the thorax and the head. The initial motion of the thorax dictates the relative movement of the top of the spine, neck, and head. For example, with knowledge of how the pediatric thorax behaves and reacts to an impact to the chest, the location of the head
and the possibility of head contact with the interior of the motor vehicle in a crash can be better predicted.

1.2 Study Benefits

A study designed to determine the material properties of the pediatric rib would provide many benefits. The knowledge of rib material properties obtained from the study can be used to develop more biofidelic anthropomorphic test devices (crash test dummies) that better mimic the response of the pediatric chest to blunt impact. An improved dummy will lead to advancements in the research of automobile safety and development of safer vehicles and restraint systems to protect child occupants. Federal Motor Vehicle Safety Standards (FMVSS) No. 213 (§571.213) states regulations for child restraint systems. The regulations include testing of child restraint systems with child crash test dummies. Procedures for these tests, as well as the correct crash test dummy to use with each child restraint system are specified (FMVSS 213). The Transportation Recall Enhancement, Accountability, and Documentation (TREAD) Act mandates the development of more side-impact and rear-impact crash tests, as well as a greater range of sizes for crash test dummies. Anton’s Law, part of the TREAD Act, also mandates the development of a ten-year-old dummy. The data that will be obtained in this study will support these Federal regulations by providing much needed information on the pediatric thorax. A better characterization of the pediatric chest will promote the development of better child restraint systems.

The victims of child abuse are another pediatric group that may benefit from this study. Although actual rib fractures in pediatric patients are uncommon in many thoracic
injuries, as many as 5-27% of the reported 1 million child abuse cases in the United States result in rib fractures (Bliss and Silen, 2002). By knowing the properties of pediatric rib bone, including strength and deflection properties, physicians will be able to better understand the degree of force to which a patient with chest trauma was exposed. Litigants of abuse cases will benefit with this knowledge by providing prosecutors with qualitative data to analyze and compare with evidence.

1.3 Previous Studies in the Literature

The amount of information available about the mechanical properties and behavior of the pediatric chest is relatively limited. Radiographic analysis of rib fractures are frequently used in child abuse studies in order to determine mechanisms of injury, but the mechanical properties are unavailable (Kleinman, et al., 2002). Sturtz cites the dissertation of Theis as a study in the four-point bending of pediatric rib bone. Theis found the absorption energy and the breaking loads of three pediatric samples. However, Theis’ sample size was small, and he analyzed all results for subjects of age 0 to 14 years as one similar group. The only valid conclusion from Theis’ study is that the bending strength of pediatric ribs is lower than that of specimens from subjects of age 15 to 64 years. No discussion of deflection or fracture patterns was included (Theis, 1975; Sturtz, 1980). Theis’ study will provide a starting point for the anticipated loads to be used in our study.

Because of the limited available data on material properties of the pediatric rib, an investigation into previous studies on the adult rib identified that the mechanical properties of adult thoracic ribs, as determined by bending tests, have been well
documented. Yoganandan and Pintar tested 30 samples of adult thoracic ribs in three-point bending tests to determine the biomechanics of the bone properties (Yoganandan and Pintar, 1998). Another study loaded adult ribs in three-point bending to determine the effects of different test parameters. Variations based on bending loading rate, rib location, asymmetry between samples, subject age, subject race, and disease were studied. However, the youngest subject was 27 years of age (Stein and Granik, 1976). Cormier, et al., investigated the effects of dynamic loading for the bending response of adult ribs. The study analyzed peak moments, strain, and geometric properties as a function of the region of the rib (Cormier, 2005). The deformation properties of the adult rib and direction of deformation with respect to the applied load were studied by Shultz, et al., in 1974. Geometrical studies have also been explored. Roberts and Chen examined the curvature and cross-sectional properties of the adult rib. The model that they developed for the anatomy of the rib is useful in determining orientation for test setup, as well as the distribution of the applied force for beam theory analysis (Roberts and Chen, 1972).

Although data available for the adult rib is much more extensive than that available for the pediatric rib, scaling the properties of the adult rib to a child is not practical because the differences are more extensive than just the size of the thorax. Figure 1.1 below shows a reconstructed computed tomography (CT) image of an adult thorax (a) and a reconstructed computed tomography (CT) image of an 8 year old thorax (b). The overall shape of the thorax and ribcage are different between the two ages. Also the heart occupies relatively more space in the pediatric thorax than in the adult thorax. The spacing of the ribs compared to the diameter of the rib varies over age as well.
The fracture mechanisms of pediatric bone also differ greatly from those of adult bone. Adult bone is significantly more brittle than that of a child, and it fractures through crack propagation as the energy from impact drives the crack through the bone. Pediatric bone does not usually experience the same crack propagation, and failure can occur by buckle fracture, bowing, or "greenstick fracture." Figure 1.2 below depicts common types of adult fractures as compared to a "greenstick fracture" of pediatric bone. A buckle failure occurs in compression much like a column will buckle under a large load. Extreme bowing of pediatric bone is a result of severe plastic deformation without an ultimate fracture. This type of fracture is common in long bones such as the ulna or fibula. A "greenstick" fracture is likened to the snapping of a "greenstick" from a new twig. The kinetic energy of the impact will dissipate as the bone in compression
undergoes plastic deformation. The energy may start a fracture along the tensile side of
the bone, but not enough energy remains after the plastic deformation to completely drive
the crack through. The child bone uses up most of the kinetic energy in bending, whereas
most of the energy results in direct fracture for the more brittle adult bone. In application
to the pediatric chest, the result is a compression of the rib cage without fracture. Rib
fracture in children is relatively uncommon, but rib cage compression frequently results
in internal injuries as mentioned above (Currey and Butler, 1975; Nahum and Melvin,
1985; Rang, 1983). The differences in mineral content and porosity cause larger plastic
deformations in pediatric bone as compared to adult bone. This plastic nature of the
pediatric bone results in a greater absorption of energy before failure. Despite the
increased energy absorption, pediatric bone will fail at a lower load than adult bone, and
therefore has a lower bending strength (Mabrey and Fitch, 1989).
Takahashi and Frost conducted a study that defined the amount of cortex in a cross section of human ribs. They found that the ratio of cortical bone area to total cross-sectional area was highest in young children, but decreased with age before steadying between 30 and 40 years of age. The ratio also dropped in the elderly. However, younger children have the smallest cortical bone cross-sectional area, bone marrow cross-sectional area, and total cross-sectional area of the rib. Children also had the lowest values for the parabolic index, which is a measure of the relative resistance of the bone to buckling with a longitudinal compressive load (Takahashi and Frost, 1966). Although it seems
counterintuitive for the ratio of cortical bone area to total cross-sectional bone area to be highest in children, the overall lower values for total cross-sectional area compared with the adult indicate a lower moment of inertia. From mechanics, a lower value for the moment of inertia will result in a larger stress in child bone than in adult bone for the same value of applied load. This means that pediatric bone will experience a higher stress at each value of the load and will fail at a lower load than the adult bone. The study by Takahashi and Frost indicates a further need to differentiate between the properties of the pediatric rib and the adult rib.

Little additional data exists in the literature on the values of the mechanical properties for pediatric bone, and what information is available is not applicable to the pediatric rib. Many of the studies that have been conducted focus on the long bones, which tend to experience extreme bowing or “greenstick” fractures. Currey and Butler loaded femoral bone in bending to obtain modulus of elasticity, bending strength, and energy absorption (Currey and Butler, 1975). Several studies in the literature obtain bending properties and ash content for fetal cranial bone (Kriewall, et al., 1981; Margulies and Thibault, 2000; McPherson and Kriewall, 1980). Also, a study from Duke University obtained the compressive and impact stiffness of the human infant head from three infant subjects ages 1 to 11 days old. The heads were tested quasi-statically, as well as dropped from specified heights to determine the deflection at impact (Prang, et al., 2004). Because the skull bone develops at a different rate than that of ribs, and its geometry and relationship to internal organs is different than that of the rib, the results of these studies cannot be applied to the pediatric rib.
1.4 Obtaining Pediatric Biomechanical Data with Limited Subject Availability

The lack of pediatric data available in the literature is mainly due to limited subject availability and moral and ethical concerns with pediatric subject testing. Current methods for determining unavailable pediatric biomechanical data include scaling from adult values. Relying on the principles of dimensional analysis, these scaling methods assume similar geometry between all subjects. Scaling, in effect, results in modeling children as smaller adults. As discussed above, the properties of adult bone cannot be applied to children due to the continuous growth of pediatric bone and its different failure modes.

In addition to scaling mechanical properties from experimental adult data based on size, animal models have frequently been used in the literature to predict loading response when available pediatric specimens are limited. A caprine (goat) model was developed by the Medical College of Wisconsin to investigate pediatric neck strength. Animal ages that corresponded with various human ages (from one-year-old through twelve-year-old and adult) were identified by analyzing CT images for bone maturation levels. Tensile and pure moment testing resulted in scaling values between the young and the fully developed caprine subjects. The authors imply similar relationships can be assumed between pediatric and adult humans (Mayer, 2000; Pintar, 2000).

Another example of an animal model study used to estimate pediatric loading response is a porcine model for abdominal injuries developed by Arbogast, et al. Anthropomorphic abdominal measurements equated a 6-year-old child to an 11-week-old pig. Belt loading to the abdomen of the porcine subjects was used to predict the abdominal belt loading response of a 6-year-old child (Arbogast, 2005).
1.5 Objective and Specific Aims

As presented in previous sections, the lack of available data in the literature indicates that there is still a great need for further research into the mechanical properties of the child rib in order to characterize the pediatric chest and to better understand thoracic injury. It is also noted that animal models may provide a source of available information when pediatric subjects are limited. To this end, the following objective and specific aims have been identified for the present study.

The objective of this study is to determine the material properties of the excised pediatric rib as a function of age and to compare the results to the material properties of the excised porcine rib.

The specific aims of this study are:

- To conduct 3-point bending tests on small samples of porcine ribs and pediatric ribs.
- To analyze the elasticity, bending strength, and material properties as a function of age.
- To compare the results of the bending tests of the pediatric rib with those of the porcine rib.
CHAPTER 2

MECHANICS AND THEORY

2.1 Rib Development and Anatomy

At the beginning of development in the embryo, the ribs form from the vertebral mass. Mesenchymal rib tissue in the form of cartilage develops by day 49. In the ninth week, ossification begins and the cartilage is transformed into bone. The primary ossification centers are located near the vertebrae and ossification occurs in the distal direction (away from the origin of ossification), but the distal ends of the rib remain cartilaginous. The rib is completely ossified at birth, although the mineral content continues to change with age after birth. The pediatric rib has greater flexibility and deformation than the more developed thoracic bone of the adult (Arey, 1965; Larsen, 1998; Tudor, 1981). Figure 2.1 below depicts this development process.
Figure 2.1. Embryological rib development depicting sites of ossification. (Larsen, 1998).

The anatomy of the rib is shown in Figure 2.2 below. Ribs are curved and arc-like. The vertebral end of the rib (on the top right of the image in Figure 2.2) connects to the vertebral body at the head of the rib. The head extends to a narrower portion known as the neck. The tubercle is the site of articulation between the rib bone and the vertebrae. The attachment of the rib to the tendon of the iliocostalis occurs at the angle of the rib, a
line on the rib surface. From this angle, the rib curves and forms the body, or shaft, of the rib. The costal groove is a marked groove on the inner surface of the rib along the length of the body (Roberts and Chen, 1972).

Figure 2.2. Anatomy of the rib. (Gray's Anatomy of the Human Body online)
2.2 Bone Structure and Mechanical Behavior

2.2.1 Bone Composition

Bone has three main functions in the body. The first mechanical function provides a rigid support for the skeletal system, while the second function is to protect the internal organs. Within the thorax, each rib contributes to the overall function of the ribcage that protects the internal organs such as the heart and lungs. The third function is mineral storage of nutrients such as calcium, phosphate, sodium, and magnesium (Schiller, 1994).

The varying types of cells in bone contribute to its ever-changing structure. The bone-lining cells of the periosteum and endosteum assist in controlling the movement of ions between the bone and the body. The periosteum lines the outer surface of the bone, and the endosteum lines the inner surface of the bone. Osteoblasts are derived from these bone-lining cells and lay down the underlying collagenous matrix of bone. Osteocytes are derived from osteoblasts and are the cells within the body of the bone. They connect to other cells via tight junctions for communication. Lastly, the osteoclasts perform bone-destroying function. They are important in bone remodeling and the destruction of diseased bone (Currey, 2002; Schiller, 1994).

The macroscopic organization of bone is characterized by two main types – cortical bone and cancellous bone. Cortical bone is dense, compact bone and forms an outer shell of most long bones. Cancellous bone, also known as trabecular bone, is the coarse, “spongy,” porous bone. The proportion of cortical and cancellous types of bone is dictated by the function of each bone. For example, the femur is a long body of cortical bone that thins at the ends as cancellous bone becomes more prominent. However, the skull is comprised of two layers of cortical bone and limited amount of cancellous bone.
(Currey, 2002; Schiller, 1994). The rib bone used in this study is comprised of a cortical shell with an inner layer of marrow. Due to the dense, compact nature of the cortical bone, this component provides the majority of the strength of the rib when a load is applied.

As with any material, the material properties of bone are dictated by its underlying structure. Bone is a composite material of collagen fibers, calcium and phosphate (mainly in the form of hydroxyapatite crystals) and other mineral deposits, and water. Water accounts for approximately 25% of the total weight of bone (Nordin and Frankel, 1989). The orientation of the collagen fibers, as well as the amount of water and minerals present, greatly affect the bone’s response to loads and deformations in different directions. The extracellular matrix is 10% cells, 60% mineralized, and 30% organic matrix. The mineralized portion of the matrix is a poorly crystalline hydroxyapatite with a net negative charge. The organic matrix is 88% type I collagen, 10% other proteins, and 1-2% lipids. Other proteins within the organic matrix are osteocalcin (a protein produced by osteoblasts, which binds to calcium and attracts monocytes for bone remodeling), osteonectin (a protein made by osteoblasts, present in platelets, and binds calcium hydroxyapatite to collagen), and phosphoproteins. The characterization of bone by a high content of inorganic materials results in the hard and rigid structure (Schiller, 1994). The ossification of bone occurs by endochondral ossification or by intramembranous ossification. In endochondral ossification cartilage is converted to bone, whereas in intramembranous ossification the membrane or fiber tissue of the periosteum becomes bone. As discussed in Section 2.1, the ossification of rib bone occurs by converting cartilage to bone, or by endochondral ossification. This ossification occurs away from
areas known as growth plates. The mechanical significance of the ossification process is that the junction of bone and cartilage at this plate region is mechanically weak compared to the surround areas.

The composition and structure of cartilage is an important factor in the development of the rib bone. Cartilage is comprised of both organic and inorganic phases. The relative amount of water in cartilage is much higher than that of bone – 80% weight as compared to 25% weight in bone. The collagen content of cartilage is 20% type II collagen, whereas the bone collagen content mentioned above is 88% type I collagen (Schiller, 1994). The different types of collagen and varying amount of water composition between bone and cartilage attribute to their vastly different properties. Because the properties of bone are dependent on its composition, the properties of young bone undergoing constant transition from cartilage to bone are influenced by the cartilage structure.

2.2.2 Biomechanical Properties

The mechanical properties of bone are related to its structure and architecture. Bone can be simplified as a two-phase composite material of both mineral and collagen. The result is a strong, brittle material (mineral) embedded in a weaker, non-brittle material (collagen). As is the nature of composite materials, the combined substances exhibit different properties together than each does alone. This behavior is strongly dependent on the interface properties of the composite materials. However, the distribution of the ratio of composites in bone material is nonhomogeneous and the properties vary along the length of bone through each cross-section (Currey, 2002). Also,
the differing architecture of cortical compared to that of cancellous bone introduces another variable when determining the properties of bone. Cortical bone is stiffer and fractures at strains greater than 2%, but cancellous bone is a porous structure that will absorb energy and not fracture until strains greater than 75% (Nordin and Frankel, 1989). As a result, the cortical shell dictates the stiffness of the rib bone. In order for the cancellous bone to absorb energy from the load, a strain would need to be applied. However, the stiffness of the cortical bone limits the amount of strain experienced before failure.

Several variables influence bone behavior in mechanical loading. The direction in which the load is applied in relation to the axis of the bone will alter the overall behavior. There is a different mechanical response whether loaded in the transverse or longitudinal directions due to the orientation of the extracellular matrix, a property known as anisotropy (Currey, 2002; Nordin and Frankel, 1989; Turner and Burr, 1993).

Loads applied to the thorax result in a bending moment on the ribs. In order to mimic this natural loading of the rib, the current study investigated the material properties of the pediatric rib using a bending test. The properties of cortical bone in bending differ from those in tension, compression, or torsion tests. Loading in bending causes the bone to bend about an axis through its cross-section producing both compressive and tensile stresses. The magnitude of these stresses throughout the cross-section is related to the distance from the bending axis (also known as the neutral axis). The neutral axis is the line of zero stress in the cross-section, and corresponds to a transition from compressive to tensile stresses. Because the cross-section of bone is asymmetrical, these stresses are unequal. As long as the specimen is comprised of the same material, the neutral axis will
pass through the centroid of the cross-section (Biewener, 1992; Currey, 2002; Nordin and Frankel, 1989).

The structure of bone causes the material to be weaker in tension than in compression. Therefore, in pure bending, the bone will fail in tension before it fails in compression (Biewener, 1992; Currey, 2002; Cowin, 2001; Nordin and Frankel, 1989). Due to its immature content, bending moments applied to pediatric bone may result in a compression failure from buckling (Currey, 2002; Nordin and Frankel, 1989). Although tensile failure is expected in bending, the calculated strength values of bone in bending are about 40% higher than the calculated strength values of bone in tensile testing. Currey proposes two mechanisms to explain this behavior. Loading in bending allows for significant post-yield deformation. As a result, the bending moment continues to increase, thereby increasing the calculated bending strength, after the yield point has been reached. Additionally, the phenomenon may be explained by the Weibull effect, which takes into account the amount of the specimen volume at high stress. When compared to bending, tensile loading results in a larger percentage of the specimen volume being loaded at high stresses. It is more likely that the weakest section of the specimen is being loaded at the highest stresses in tensile loading than in bending loading (Currey, 1999; Currey, 2002). It is also important to note that the torsional shear strength of bone is weaker than the tensile stress. Any shear stress applied due to torsion may result in a failure in shear (Nordin and Frankel, 1989). The theory and analysis of bending tests will be discussed in detail in Section 2.3.

The water content of bone influences its viscoelastic property. Because bone is a viscoelastic material, it is highly influenced by strain rate. Higher strain rates increase the
brittle behavior and increase the amount of load that can be applied before failure, resulting in higher ultimate and yield stresses. There is dissipation of energy due to the friction of water flowing through the bone structure when a load is applied. Water acts as a “shock absorber” when the bone is loaded, and its resistance to load changes with the rate at which the load is applied. It has been reported that a 15% increase in bone strength will be seen for an increase in strain rate by one order of magnitude (Turner and Burr, 1993). Nordin and Frankel discuss an almost doubling of load to failure and energy to failure in canine tibiae when the loading time is decreased (resulting in an increased loading rate) from 200 seconds to 0.01 seconds (Nordin and Frankel, 1989). Another viscoelastic property is evident as bone continues to deform when loading is held at a constant stress. Similarly when strain is held constant, the induced stress will decrease over time. Finally, if there is significant drying of the bone, the water no longer contributes a viscoelastic property. Dry bone behaves more like a spring without the strain rate effects of the viscoelastic wet bone. An increased value for Young’s modulus and bone strength is seen in dry bone as compared to hydrated bone (Cowin, 2001; Nordin and Frankel, 1989; Turner and Burr, 1993).

Age-related ductile or brittle behaviors can also be influenced by water content. Density changes with aging in the elderly result in reduced cancellous bone and a thinner cortical shell. Water content is also reduced. As a result, bone strength, elasticity, and energy absorption to failure all decrease (Nordin and Frankel, 1989). Young bone is less stiff due to its lower mineralization, as well as higher water content.

Another distinctive characteristic of bone is related to it being a living tissue. Bone has the ability to adapt to the loads it experiences and change the distribution of
material to alter mechanical properties. This phenomenon is explained by Wolff’s Law, which states that bone is deposited in areas exposed to higher loading and resorbed in areas exposed to lower loading. It is believed that this adaptation is controlled by the induced strain from the loading. As a result, the bone environment is mechanically active and constantly changing and adapting to meet loading requirements. However, these active changes are not short-term events. The process of bone remodeling occurs over an extended period of time, and is in response to long-term changes in the loading pattern on bone. Situations in which bone may undergo remodeling include healing after fracture, as well as adapting to a new implant in the bone. Additionally, bone remodeling is a rejuvenating process by which the cells of bone are absorbed and new bone is deposited in its place (Nordin and Frankel, 1989).

Remodeling of bone is a means for altering the moment of inertia, or second moment of area, of the bone cross-section. The moment of inertia is indicative of the cross-sectional distribution of material from the centroid, and is a geometrical property. For two specimens of the same material, a larger moment of inertia corresponds to a higher bending strength. By increasing the distribution of material from the centroid through remodeling, the bone can withstand a higher bending stress. Conversely, if the bone is not subjected to large bending moments, remodeling of bone to decrease the moment of inertia reduces the bending strength (Carter, 1985; Cowin, 2001).

2.3 Beam Theory

Due to the simplicity of the test setup compared to other mechanical testing methods, bending tests are frequently used to determine the material properties of bone.
These experimental tests rely on beam theory from mechanics for methods of analysis. A brief discussion of beam theory for bending is presented in the following sections.

2.3.1 Three-Point versus Four-Point Bending

The difference between three-point and four-point bending is the way in which the load is applied to the specimen. Both methods of mechanical bending tests use point loads to produce a bending moment in beam-like test specimens. The tensile side of the specimen in bending is supported on the two far ends of its span. In three-point bending a single point load is applied to the center of the span on what would be the compressive side of the specimen in bending. The moment in four-point bending is produced by two point loads of equal force applied equidistant from the mid-span of the beam. Both methods produce a maximum bending moment at the center of the beam. For three-point bending this maximum moment is directly under the point load in the middle of the beam. The maximum moment in four-point bending is constant in between the two point loads on the compressive side of the beam (Cowin, 2001; Turner and Burr, 1993). These loading conditions for three-point versus four-point bending are summarized in Figure 2.3 below.
Figure 2.3. Comparison of 3-point versus 4-point loading applied forces and moments.

The advantages of three-point bending over four-point bending are that the test setup is easier to achieve for three-point bending and that three-point bending more closely replicates the type of loading experienced in the body. With four-point bending, it is pertinent to ensure that both of the point loads applied on the compressive side of the beam be exactly equal (Cowin, 2001; Turner and Burr, 1993). Any variation in the curvature of the beam makes this extremely difficult, and the ribs to be tested in this study do have a slight curvature. Although a distributed load is applied to the ribs during inspiration and expiration, the levels of this loading are not enough to cause failure. Rib fractures and injuries to the thorax are frequently due to a distributed load applied over an area of the body. In two-dimension space, a distributed load through a cross-section can
be analyzed as a concentrated point load applied at the middle of the distributed load length. Three-point bending more closely mimics this loading condition.

The advantages of four-point bending for bone over three-point bending are related to the constant moment between the two middle point loads (Cowin, 2001). Because bone is nonhomogeneous, the weakest section may not be the cross-section to which the single load in three-point bending is applied. The constant maximum moment in four-point bending will cause failure at the weakest cross-section over the length between the two loads on the compressive side of the beam. Also, the transverse shear stress is equal to zero over the length of constant maximum moment in four-point bending. A three-point bending test setup causes a constant shear between the end-points of the span to the applied point load (Cowin, 2001; Turner and Burr, 1993). Also, ASTM standards indicate that three-point bending analysis is only valid for strains less than 5% (ASTM D 790-03). Four-point bending creates less deformation, and therefore less strain, due to induced shear stresses.

After analysis of the two options for bending testing of bone, the three-point bending test setup was chosen for this study. The complications of ensuring a constant bending moment in four-point bending made the simplicity of the three-point bending test setup more desirable. Also, because bone failure in shear is related to torsional loads, as opposed to bending moment loads, the induced transverse shear in three-point bending may not be as large of a concern as ensuring pure bending and no torsion during test setup. As stated earlier, bone will fail in tension before compression when a bending moment is applied. Because failure of bone usually occurs in strains of around 2%, the invalidity of three-point bending with strains of greater than 5% was not of concern. It is
important to note, however, that the 2% strain failure is specified for adults, but it is unknown whether or not the same strain failure is applicable for pediatric bone.

Furthermore, three-point bending tests on ribs have been previously published in the literature (Cormier, 2005; Stein and Granik, 1976; Yoganandan and Pintar, 1998).

2.3.2 Assumptions in Three-Point Bending

Three-point bending test setups are designed for analyzing the behavior of straight beams when a bending moment is applied. In order for the analysis to be valid several assumptions are made:

- General equations for analysis of three-point bending behavior are valid only before yielding occurs, and the material remains linearly elastic up until the point of failure (Currey, 2002; Turner and Burr, 1993).

- The beam is a straight beam; ie, the ratio of the radius of curvature to the depth of the beam is greater than five (R/d > 5) (Boresi, 1993). This straight beam assumption has also been reported as R/d > 8 (Cowin, 2001).

- Error from the flexure formula is negligible if the test specimen is sufficiently long and slender; ie, the ratio of the span of the beam length to the maximum cross-sectional dimension of the beam must be larger than five (L/d > 5) (Boresi, 1993). This minimizes the effects of the induced transverse shear stresses from three-point bending, as there are no shear stresses in pure bending (Carter, 1985; Stevens, 1987). This negligible shear assumption has also been reported as L/d > 16 (ASTM D 790-03, 2003; Cowin, 2001; Turner and Burr, 1993).

- Moduli remain constant throughout the test specimen (Cowin, 2001).
• The neutral and centroidal axes coincide (Biewener, 1992; Cowin, 2001). The first moment of area about the neutral axis is equal to zero.
• Material is homogeneous and isotropic (Stevens, 1987).

2.3.3 Asymmetrical Cross-Sections and Minimum Potential Energy Theorem

The cross-section of bone is not symmetrical, and therefore involves a more complex analysis than derived equations for standard cross-sections. The equation of the neutral axis is found by equating the stress flexure formula to be zero. This gives the neutral axis as a line through the centroid at some angle from the horizontal axis (Ugural and Fenster, 1995). The minimum potential energy theorem may provide means for predicting the neutral axis experimentally.

The general equation for bending stress is given by $\sigma = \frac{My}{I}$, where $\sigma$ is the bending stress, $M$ is the bending moment, $y$ is the distance from the centroid to the outermost fibers in the vertical direction, and $I$ is the moment of inertia. Holding all variables constant in this equation and varying only the moment of inertia $I$, it can be shown that the maximum applied stress for a given load (or applied bending moment) will occur at the orientation with the lowest value of the moment of inertia, $I_{\text{min}}$. Because the moment of inertia is indicative of the distribution of area from the centroid, its value changes depending on the orientation of the axis about which it is calculated. Also of note is the correlation between the axes of the principal stresses of the cross-section with the axes of the principal values of moment of inertia (Ugural and Fenster, 1995). The values of the moment of inertia at the orientation of these principal axes will give the maximum
and minimum moment of inertia calculations over the entire cross-section. The axis of the maximum moment of inertia corresponds to the axis of the minimum stress, whereas the axis of the minimum moment of inertia corresponds to the axis of the maximum stress (Biewener, 1992; Ugural and Fenster, 1995). Failure would be expected through the orientation experiencing the largest stress value; i.e., failure is expected through the orientation with the minimum moment of inertia. This would result in a correlation of the principal axes of stress and the principal axes of the moment of inertia.

The minimum potential energy theorem indicates that a system with an external load will be in equilibrium when the energy is at a minimum value (Fung 1965). The strain energy is directly related to value of the moment of inertia (Timoshenko 1958). A larger value of moment of inertia will result in a higher strain energy. Therefore, the orientation with the minimum value of the moment of inertia would correspond to the minimum energy value. Additionally, as discussed in the previous paragraph, loading through the axis with the maximum stress for a given load is expected to cause failure in three-point bending. This represents the orientation through which the least amount of energy is stored before failure will occur. In order for the minimum potential energy theorem to hold, the orientation of the axis of the minimum moment of inertia is expected to be the same as the bending axis. From this argument, it is again expected that failure will occur through the orientation with the minimum moment of inertia.

The above-discussed arguments indicate that theoretical assumptions may predict failure through the cross-sectional orientation corresponding to a bending moment about the axis of the minimum moment of inertia. However, the axes of principal stress are dictated by the direction of the applied load. In order for the assumption that failure
occurs through the orientation about the axis of the minimum moment of inertia, the bone must be oriented through this axis during testing. Whether or not the 3-point bending test setup allows for the rib to rotate to this position during testing needed to be determined. Verifying the validity of these assumptions was the focus of the preliminary testing discussed in the following chapter.
CHAPTER 3

PRELIMINARY INVESTIGATION

3.1 Preliminary Questions

Although three-point bending test methods and analyses are well established for many materials, the use of a three-point bending setup to analyze the material properties of rib bone raised additional questions. The theory behind beam bending tests assumes that the beam cross-section is constant throughout the test sample. However, when using a non-machined biological specimen, such as a rib, the constant cross-section assumption does not hold. There will be slight variations in this cross-section throughout the length of the beam. Because the cross-section along the length of the rib bone is not completely uniform, rotation may occur during testing as the orientation of the neutral axis changes over the testing length. Additionally, the rib surface is rounded and it is being loaded on a rounded support. This leads to the possibility that the orientation during testing may be dictated more by the external contour of the bone sample than by bending about the axis of the minimum moment of inertia. There are no absolute certainties for the testing of biological specimens, but it is not known whether the proposed test setup for bend testing of the rib samples would allow for values within an acceptable range of error for biological specimens.
3.2 Preliminary Study Test Goals

There are two main goals for the preliminary testing in order to determine if beam theory and the minimum potential energy theory hold for the proposed test setup, or if the bending orientation is dictated by the external contours of the bone specimen.

1. The first goal is to trace the bone cross-section rotation during testing. The purpose of this goal is to determine if the bone cross-section being loaded continually rotates throughout the loading or if an equilibrium point is established. Rotation during loading may also be indicative of torsion over the length of the rib.

2. The second goal is to determine the orientation of the bending axis. The orientation of this bending axis will correlate to the orientation of the neutral axis. The moment of inertia about the bending axis can then be determined. By also calculating the value of the minimum moment of inertia, as well as the angle between the bending axis and the axis of the minimum moment of inertia, comparisons can be made between the actual and theoretical values in order to determine if the deviations are within acceptable ranges.

3.3 Preliminary Findings

The first data set used to investigate the questions discussed above utilized porcine rib samples from levels 9 through 14. Each sample was cleaned of surrounding tissue, including the periosteum. A felt-tip pen was used to draw a line around the cross-
section at which the load would be applied. Small sections of a radio-opaque catheter were cut and affixed to the sides of each sample with Gorilla Glue®. These catheter sections are visible in a CT (computed tomography) scan (see Figure 3.1 below), and marked the cross-section of interest in the CT image, as well as provided reference points for orienting the CT image on the screen. All samples were CT scanned prior to loading to ensure that cross-sections obtained from the images would be the true shape before any possible changes to the bone cross-section as a result of testing.

![CT image](image_url)

**Figure 3.1.** CT image of test specimen, depicting radio-opaque catheter markers.
The three-point bend test fixture shown below was used for all preliminary investigations.

![Three-point bend test fixture](image)

Figure 3.2. Three-point bend test fixture.

The test fixture connected to the cross-head of an MTS Material Testing System® (MTS Systems Corporation; Eden Prairie, MN). A load cell measured the load being applied to the cross-section of interest. The bottom of the test fixture is aligned such that the cross-head and top point of the fixture loads the bone at the middle of the span. This setup loaded the top of the bone (external surface of the rib) in compression and the bottom (internal surface of the rib) in tension. This loading configuration simulates a single external force applied to the rib causing bending.

The bone was held in place prior to testing with a pre-load of 5 Newtons. This was necessary so that the bone did not fall out of the test fixture before testing commenced. The bone was loaded at a quasi-static deflection rate of 1mm/minute. For
the first preliminary trial, a video camera was positioned at each side of the MTS device in order to record the bone movement during the test. Scales were placed within view of the camera to allow for calibration of the distance on the image. Figure 3.3 below indicates an image from the video recordings that was used to determine the orientation of the bone cross-section at various moments in time. The surface of the test fixture was used as the flat surface, and vertical measurements were taken from this surface to the center of each section of radio-opaque catheter.

Figure 3.3. Image from video of rib level 14R from a 14-week-old pig.

After calibrating the image pixel size to the scales depicted in the camera view, the image was oriented so that the flat surface of the test fixture was along the horizontal. Vertical measurements from this flat surface to the center of each marker were
determined. From these vertical displacements, simple geometric calculations yielded an angle $\theta$ from the horizontal. Figure 3.4 below indicates the necessary measurements.

$$\sin \theta = \frac{D}{h}$$

![Diagram showing measurements](image)

Figure 3.4. Measurements taken from video images for calculation of orientation angles.

In order to analyze goal (1) above to determine the amount of rotation of the cross-section during testing, these angle measures were taken at several points throughout the video. The amount of rotation was defined as the difference between the maximum angle value and the minimum angle value.

The values for these angles from the horizontal were used to rotate a CT image of each cross-section of interest using the program ImageJ (version 1.36b) and the macro
"MomentMacro" written by Dr. Chris Ruff of Johns Hopkins University. (See Chapter 4 – Study Design for a more detailed description of this program.)

Comparisons between the minimum moment of inertia value \( I_{\text{min}} \) and the moment of inertia about the horizontal axis \( I_{xx} \) were made using Equation 3.1. This value represented the percent difference between the principal minimum axis \( I_{\text{min}} \) and the actual axis \( I_{xx} \) values for moment of inertia.

\[
\text{%Deviation} = \frac{I_{xx} - I_{\text{min}}}{I_{\text{min}}} \times 100\% \tag{3.1}
\]

The first trial to secure the catheters was unsuccessful, as the glue expanded around the bone and provided a loading surface not consistent with the surface of the bone. This caused the test device to load the glue and not the bone, as indicated in Figure 3.5 below.

![Image](image_url)

Figure 3.5. Image from video recording of rib level 12R from a 4-week-old pig.
The force versus deflection data from the rib level 12R from a 4-week-old pig is presented in Figure 3.6 below. The vertical axis is the load in Newtons, and the horizontal axis is the displacement in millimeters. The loads measured are more indicative of the force required to remove the glue from the bone than of the mechanical properties of the bone itself. The steep drop-off in force prior to 0.8 mm of deflection in the plot corresponds to the point in time when the glue separated from the rib and loading was applied directly the bone, as opposed to loading on the surface of the excess glue coating the bone.

Figure 3.6. Force versus deflection plot for rib level 12R from a 4-week-old pig.
In addition to the error introduced by the glue, the accuracy of the measurements from the video images was not certain. The problems with the test setup resulted in large deviations of the moment of inertia calculated about the actual bending axis, when compared to the moment of inertia calculated about the principal axis of the minimum moment of inertia. These values are show in Table 3.1 below.

<table>
<thead>
<tr>
<th>Name</th>
<th>Imin (mm^4)</th>
<th>I about bending axis from video (mm^4)</th>
<th>% Deviation from Imin</th>
</tr>
</thead>
<tbody>
<tr>
<td>4Weeks_9R</td>
<td>7.51</td>
<td>6.86</td>
<td>-8.69%</td>
</tr>
<tr>
<td>4Weeks_9L</td>
<td>7.10</td>
<td>23.74</td>
<td>234.43%</td>
</tr>
<tr>
<td>4Weeks_13R</td>
<td>6.77</td>
<td>8.08</td>
<td>19.35%</td>
</tr>
<tr>
<td>4Weeks_13L</td>
<td>8.57</td>
<td>8.95</td>
<td>4.46%</td>
</tr>
<tr>
<td>4Weeks_12R</td>
<td>8.05</td>
<td>12.47</td>
<td>54.93%</td>
</tr>
<tr>
<td>4Weeks_12L</td>
<td>8.51</td>
<td>9.86</td>
<td>15.83%</td>
</tr>
<tr>
<td>4Weeks_11R</td>
<td>9.30</td>
<td>9.59</td>
<td>3.10%</td>
</tr>
<tr>
<td>4Weeks_11L</td>
<td>6.28</td>
<td>7.37</td>
<td>17.39%</td>
</tr>
<tr>
<td>4Weeks_10R</td>
<td>5.32</td>
<td>6.74</td>
<td>26.78%</td>
</tr>
<tr>
<td>4Weeks_10L</td>
<td>8.14</td>
<td>7.91</td>
<td>-2.77%</td>
</tr>
<tr>
<td>14Weeks_9R</td>
<td>245.53</td>
<td>263.17</td>
<td>7.19%</td>
</tr>
<tr>
<td>14Weeks_10R</td>
<td>181.59</td>
<td>155.18</td>
<td>-14.54%</td>
</tr>
<tr>
<td>14Weeks_11R</td>
<td>276.29</td>
<td>268.80</td>
<td>-2.71%</td>
</tr>
<tr>
<td>14Weeks_12R</td>
<td>183.15</td>
<td>206.63</td>
<td>12.82%</td>
</tr>
<tr>
<td>14Weeks_13R</td>
<td>150.70</td>
<td>169.25</td>
<td>12.30%</td>
</tr>
<tr>
<td>14Weeks_14R</td>
<td>129.96</td>
<td>121.75</td>
<td>-6.32%</td>
</tr>
</tbody>
</table>

Table 3.1. Moment of inertia data from first test of preliminary investigation.

Although there were problems with the test setup, most of the test samples showed a % deviation for the minimum moment of inertia of less than 20%. The findings indicated that the majority of samples did bend about the principal axis of the minimum
moment of inertia, even if there were problems with the loading on glue and not on the bone itself.

Measurements of the amount of rotation during testing showed that one bone rotated approximately 25° during testing. This was rib “4Weeks_9L,” which also had the largest deviation of the bending axis from the principal axis. The next largest amount of rotation was approximately 8°. However, it was not known if this rotation occurred before or after yielding as there was no way to link the video data with the force versus deflection curve. The small amount of rotation did, however, indicate that the rib bone was loaded through a fairly constant orientation.

It is important to note a discrepancy presented in Table 3.1 above. In some of the samples, the calculated minimum moment of inertia is larger than the moment of inertia about the bending axis determined from the video during testing. The explanation for this is the program used for the calculations at the time. One version of ImageJ was used to calculate the bending axis moment of inertia, while a different Macintosh-compatible version of NIH Image was used to calculate the minimum moment of inertia. Variation between the two programs increased the error with this method. Also, the angle difference between the orientation of the bending axis and the orientation of the principal axis of the minimum moment of inertia were not known. A newer version of the “MomentMacro” was identified for use with ImageJ and used in subsequent trials. This version allowed for both calculations of the minimum moment of inertia, as well as the moment of inertia about the bending axis, from the same program. Additionally, the difference in orientation between the two axes could be determined.
In order to address the issues with the above methods, additional testing was conducted using less glue and better technique for affixing the catheters, as well as using a FaroArm® (FARO Technologies, Inc; Lake Mary, FL) to take the orientation measurements. The FaroArm® directly calculated the same angle shown in Figure 3.4, minimizing the error in these measurements as compared to the video image measurements. FaroArm® measurements were taken every 0.5 mm of deflection. This allowed for linking the yield point of the bone with the orientation measurements. Additionally, a new macro for ImageJ was identified that calculated the angle the bending axis was off from the principal axis of the minimum moment of inertia. The analyses of the total amount of rotation, as well as the comparisons between the actual ($I_\alpha$) and the theoretical ($I_{min}$) values for moment of inertia are the same as discussed above for the first test.

The results shown below in Table 3.2 show that only one test sample had a deviation from the minimum moment of inertia by more than 15%.
<table>
<thead>
<tr>
<th>Sample ID</th>
<th>Deflection (mm)</th>
<th>Angle from Principal Axis</th>
<th>$I_{min}$ (mm$^4$)</th>
<th>$I_x$ (mm$^4$)</th>
<th>% Deviation from $I_{min}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>6Week_9L</td>
<td>0.0 mm</td>
<td>-7.34</td>
<td>55.31</td>
<td>56.28</td>
<td>1.75%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-7.77</td>
<td>54.16</td>
<td>55.30</td>
<td>2.10%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-4.63</td>
<td>55.54</td>
<td>55.93</td>
<td>0.71%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-4.61</td>
<td>54.94</td>
<td>55.33</td>
<td>0.72%</td>
</tr>
<tr>
<td>6Week_10L</td>
<td>0.0 mm</td>
<td>1.68</td>
<td>40.90</td>
<td>40.98</td>
<td>0.20%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>3.45</td>
<td>42.59</td>
<td>42.94</td>
<td>0.80%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>4.35</td>
<td>42.26</td>
<td>42.79</td>
<td>1.27%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>3.27</td>
<td>42.23</td>
<td>42.53</td>
<td>0.72%</td>
</tr>
<tr>
<td>6Week_11L</td>
<td>0.0 mm</td>
<td>-12.75</td>
<td>42.19</td>
<td>47.17</td>
<td>11.82%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-14.05</td>
<td>41.43</td>
<td>47.36</td>
<td>14.33%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-13.77</td>
<td>41.67</td>
<td>47.42</td>
<td>13.80%</td>
</tr>
<tr>
<td>6Week_12L</td>
<td>0.0 mm</td>
<td>-0.39</td>
<td>33.25</td>
<td>33.25</td>
<td>0.01%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-1.56</td>
<td>34.40</td>
<td>34.45</td>
<td>0.14%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-1.00</td>
<td>33.35</td>
<td>33.37</td>
<td>0.06%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>1.11</td>
<td>33.20</td>
<td>33.23</td>
<td>0.07%</td>
</tr>
<tr>
<td>6Week_13L</td>
<td>0.0 mm</td>
<td>1.74</td>
<td>23.53</td>
<td>23.56</td>
<td>0.21%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>1.89</td>
<td>23.81</td>
<td>23.87</td>
<td>0.24%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>0.18</td>
<td>22.39</td>
<td>22.39</td>
<td>0.00%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>0.42</td>
<td>22.23</td>
<td>22.24</td>
<td>0.01%</td>
</tr>
<tr>
<td>6Week_14L</td>
<td>0.0 mm</td>
<td>1.36</td>
<td>19.25</td>
<td>19.31</td>
<td>0.32%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>3.99</td>
<td>19.28</td>
<td>19.79</td>
<td>2.65%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>1.45</td>
<td>19.32</td>
<td>19.39</td>
<td>0.36%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-3.02</td>
<td>20.06</td>
<td>20.35</td>
<td>1.49%</td>
</tr>
</tbody>
</table>

Continued

Table 3.2. Moment of inertia data from second test of preliminary investigation.
Table 3.2 continued

<table>
<thead>
<tr>
<th>Sample ID</th>
<th>Deflection</th>
<th>Angle from Principal Axis (mm^4)</th>
<th>$I_{\text{min}}$ (mm^4)</th>
<th>$I_{x}$ (mm^4)</th>
<th>% Deviation from $I_{\text{min}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>8Week_9R</td>
<td>0.0 mm</td>
<td>-1.92</td>
<td>118.98</td>
<td>119.22</td>
<td>0.19%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-2.11</td>
<td>120.53</td>
<td>120.81</td>
<td>0.23%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-5.44</td>
<td>118.86</td>
<td>120.66</td>
<td>1.52%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-4.59</td>
<td>116.12</td>
<td>117.48</td>
<td>1.16%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-7.55</td>
<td>117.78</td>
<td>121.36</td>
<td>3.03%</td>
</tr>
<tr>
<td>8Week_10R</td>
<td>0.0 mm</td>
<td>-10.72</td>
<td>85.32</td>
<td>91.36</td>
<td>7.08%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-9.65</td>
<td>83.30</td>
<td>88.25</td>
<td>5.94%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-8.95</td>
<td>85.37</td>
<td>89.56</td>
<td>4.91%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-10.97</td>
<td>85.11</td>
<td>91.41</td>
<td>7.40%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-12.52</td>
<td>85.16</td>
<td>93.32</td>
<td>9.59%</td>
</tr>
<tr>
<td>8Week_11R</td>
<td>0.0 mm</td>
<td>-17.59</td>
<td>69.22</td>
<td>79.56</td>
<td>14.93%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-22.82</td>
<td>69.85</td>
<td>86.56</td>
<td>23.92%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-20.46</td>
<td>65.88</td>
<td>78.34</td>
<td>18.91%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-18.30</td>
<td>70.17</td>
<td>80.99</td>
<td>15.43%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-17.52</td>
<td>70.61</td>
<td>80.80</td>
<td>14.43%</td>
</tr>
<tr>
<td>8Week_12R</td>
<td>0.0 mm</td>
<td>-15.72</td>
<td>69.75</td>
<td>75.76</td>
<td>8.62%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-18.57</td>
<td>68.81</td>
<td>76.93</td>
<td>11.80%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-16.23</td>
<td>69.26</td>
<td>75.83</td>
<td>9.50%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-17.49</td>
<td>67.88</td>
<td>75.08</td>
<td>10.60%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-19.47</td>
<td>69.46</td>
<td>78.66</td>
<td>13.24%</td>
</tr>
<tr>
<td>8Week_13R</td>
<td>0.0 mm</td>
<td>-2.90</td>
<td>59.06</td>
<td>59.34</td>
<td>0.48%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>2.62</td>
<td>57.34</td>
<td>57.57</td>
<td>0.41%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>3.09</td>
<td>58.07</td>
<td>58.39</td>
<td>0.55%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>10.15</td>
<td>57.05</td>
<td>60.53</td>
<td>6.09%</td>
</tr>
<tr>
<td>8Week_14R</td>
<td>0.0 mm</td>
<td>-1.83</td>
<td>54.18</td>
<td>54.36</td>
<td>0.33%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-3.89</td>
<td>53.60</td>
<td>54.40</td>
<td>1.49%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-3.46</td>
<td>54.32</td>
<td>54.96</td>
<td>1.18%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-1.09</td>
<td>53.36</td>
<td>53.43</td>
<td>0.12%</td>
</tr>
</tbody>
</table>
The changes in orientation during testing showed that one rib rotated up to 13°, but all others rotated less than 7°. Again, this shows that the bone was loaded at a consistent orientation, with negligible rotation during testing.

Because all ribs tested in the second test were from rib levels inferior to those being used in the full-scale testing, the samples used in the third test were from rib level 8, in order to determine if the same results were seen. The same setup as the second test was used. The moment of inertia results shown below in Table 3.3 show that all ribs had less than 12% deviation of the moment of inertia about the bending axis from the moment of inertia about the minimum principal axis. Only two of these bones were slightly higher than 10%.
<table>
<thead>
<tr>
<th>Sample ID</th>
<th>Deflection</th>
<th>Angle from Principal Axis</th>
<th>$I_{\text{min}}$ (mm$^4$)</th>
<th>$I_x$ (mm$^4$)</th>
<th>% Deviation from $I_{\text{min}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>8L_3Weeks</td>
<td>0.0 mm</td>
<td>-4.79</td>
<td>7.99</td>
<td>8.04</td>
<td>0.67%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>1.36</td>
<td>7.82</td>
<td>7.83</td>
<td>0.06%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>0.84</td>
<td>7.49</td>
<td>7.49</td>
<td>0.01%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>5.38</td>
<td>7.63</td>
<td>7.70</td>
<td>0.87%</td>
</tr>
<tr>
<td>8L_4Weeks</td>
<td>0.0 mm</td>
<td>-12.01</td>
<td>12.62</td>
<td>13.26</td>
<td>5.12%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-17.47</td>
<td>12.95</td>
<td>14.29</td>
<td>10.36%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-16.09</td>
<td>12.71</td>
<td>13.81</td>
<td>8.70%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-15.96</td>
<td>13.26</td>
<td>14.35</td>
<td>8.20%</td>
</tr>
<tr>
<td>8L_6Weeks</td>
<td>0.0 mm</td>
<td>12.96</td>
<td>59.03</td>
<td>62.72</td>
<td>6.24%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>11.60</td>
<td>59.45</td>
<td>62.31</td>
<td>4.81%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>9.45</td>
<td>59.04</td>
<td>60.96</td>
<td>3.25%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>8.17</td>
<td>59.05</td>
<td>60.45</td>
<td>2.37%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>6.16</td>
<td>58.11</td>
<td>58.94</td>
<td>1.40%</td>
</tr>
<tr>
<td>8L_7Weeks</td>
<td>0.0 mm</td>
<td>9.43</td>
<td>85.84</td>
<td>87.41</td>
<td>1.83%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>4.48</td>
<td>89.02</td>
<td>89.38</td>
<td>0.40%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>8.27</td>
<td>86.89</td>
<td>88.03</td>
<td>1.31%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>7.51</td>
<td>88.13</td>
<td>89.09</td>
<td>1.09%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>6.23</td>
<td>86.59</td>
<td>87.25</td>
<td>0.76%</td>
</tr>
</tbody>
</table>

Table 3.3. Moment of inertia data from third test of preliminary investigation.
Table 3.3 Continued

<table>
<thead>
<tr>
<th>Sample ID</th>
<th>Deflection</th>
<th>Angle from Principal Axis</th>
<th>I_min (mm^4)</th>
<th>I_x (mm^4)</th>
<th>% Deviation from I_min</th>
</tr>
</thead>
<tbody>
<tr>
<td>8L_9Weeks</td>
<td>0.0 mm</td>
<td>1.72</td>
<td>13.22</td>
<td>13.25</td>
<td>0.22%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-4.72</td>
<td>13.37</td>
<td>13.59</td>
<td>1.64%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-1.03</td>
<td>12.69</td>
<td>12.90</td>
<td>0.08%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-2.48</td>
<td>12.69</td>
<td>12.75</td>
<td>0.49%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-7.74</td>
<td>12.91</td>
<td>13.46</td>
<td>4.27%</td>
</tr>
<tr>
<td>8L_10Weeks</td>
<td>0.0 mm</td>
<td>-17.95</td>
<td>425.38</td>
<td>465.55</td>
<td>9.44%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-20.10</td>
<td>425.53</td>
<td>475.07</td>
<td>11.64%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-18.83</td>
<td>430.02</td>
<td>474.64</td>
<td>10.38%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-19.94</td>
<td>432.24</td>
<td>481.92</td>
<td>11.49%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-16.59</td>
<td>427.20</td>
<td>461.81</td>
<td>8.10%</td>
</tr>
<tr>
<td>8L_19Weeks</td>
<td>0.0 mm</td>
<td>11.47</td>
<td>864.25</td>
<td>883.20</td>
<td>2.19%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>16.13</td>
<td>895.32</td>
<td>932.23</td>
<td>4.12%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>13.80</td>
<td>896.47</td>
<td>923.85</td>
<td>3.05%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>17.46</td>
<td>892.87</td>
<td>937.38</td>
<td>4.98%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>14.96</td>
<td>885.62</td>
<td>917.48</td>
<td>3.60%</td>
</tr>
<tr>
<td></td>
<td>2.5 mm</td>
<td>16.23</td>
<td>895.24</td>
<td>933.62</td>
<td>4.29%</td>
</tr>
<tr>
<td></td>
<td>3.0 mm</td>
<td>18.82</td>
<td>895.32</td>
<td>944.50</td>
<td>5.49%</td>
</tr>
<tr>
<td>8L_20Weeks</td>
<td>0.0 mm</td>
<td>-11.79</td>
<td>1266.54</td>
<td>1289.22</td>
<td>1.79%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>-11.32</td>
<td>1271.47</td>
<td>1292.13</td>
<td>1.62%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>-12.55</td>
<td>1258.41</td>
<td>1293.67</td>
<td>1.99%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>-10.87</td>
<td>1258.94</td>
<td>1278.21</td>
<td>1.53%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>-12.65</td>
<td>1264.53</td>
<td>1289.29</td>
<td>1.96%</td>
</tr>
<tr>
<td></td>
<td>2.5 mm</td>
<td>-12.18</td>
<td>1267.71</td>
<td>1291.33</td>
<td>1.86%</td>
</tr>
<tr>
<td></td>
<td>3.0 mm</td>
<td>-9.11</td>
<td>1266.90</td>
<td>1280.32</td>
<td>1.06%</td>
</tr>
</tbody>
</table>
The changes in orientation during testing showed that two ribs rotated up to 9°, but all others rotated less than 7°. Again, this shows that the bone was loaded at a consistent orientation, with negligible rotation during testing.

The fourth test addressed the concern of the accuracy of taking FARO data points from the location at which the catheter was glued to the bone. It was suggested that a penned-on cross-hair would provide more confident and accurate results. (See section 3.5 for numerical analysis of the error in the test setup.) Because there would be no radio-opaque marker affixed to the bone during CT imaging, the orientation of the cross-section on the image could not be determined. This orientation needed to be determined before the bone was transported to CT, and the cross-section needed to be maintained at the same orientation until scanning was complete. The orientation at which the rib went through the CT scanner was determined by a pre-CT FaroArm® measurement instead of measuring angles with the ImageJ program. This was accomplished by holding the ribs at a constant orientation from the point FaroArm® measurements were taken until the bone went through CT scanning. Figure 3.7 below shows the foam holders constructed to ensure that the bone was at the same angle from the horizontal both before and during CT scanning. The bone was rigidly fixed in the support.
The moment of inertia results presented in Table 3.4 show that one bone had a deviation of more than 30% from the principal axis, while a second bone had a deviation of more than 50% from the principal axis. However, the rib with the larger than 50% deviation became unloaded after an initial 25N of force was applied. The mechanism for this release is unknown. Results again confirmed that each rib sustained a minimum amount of rotation during loading. (See Appendix B.) Concerns from this test were that the rib sections with the largest deviation from the minimum moment of inertia axis were not selected from the straightest portion of the beam. It is believed that the error would be less if the bones were more carefully cut to ensure a straight portion of beam, as the test setup is meant for straight beams.
<table>
<thead>
<tr>
<th>Sample ID</th>
<th>Deflection</th>
<th>Angle from Principal Axis (mm^4)</th>
<th>I_min (mm^4)</th>
<th>I_x (mm^4)</th>
<th>% Deviation from I_min</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 weeks</td>
<td>0.0 mm</td>
<td>10.42</td>
<td>11.19</td>
<td>11.31</td>
<td>1.08%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>8.82</td>
<td>11.05</td>
<td>11.13</td>
<td>0.80%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>23.19</td>
<td>11.11</td>
<td>11.66</td>
<td>4.99%</td>
</tr>
<tr>
<td>9 weeks</td>
<td>0.0 mm</td>
<td>22.28</td>
<td>10.34</td>
<td>13.62</td>
<td>31.73%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>24.42</td>
<td>10.31</td>
<td>14.25</td>
<td>38.21%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>21.93</td>
<td>10.36</td>
<td>13.57</td>
<td>30.93%</td>
</tr>
<tr>
<td>12 weeks</td>
<td>0.0 mm</td>
<td>11.09</td>
<td>970.10</td>
<td>994.86</td>
<td>2.55%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>12.69</td>
<td>969.85</td>
<td>1002.13</td>
<td>3.33%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>14.99</td>
<td>969.45</td>
<td>1014.31</td>
<td>4.63%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>12.12</td>
<td>970.90</td>
<td>1000.44</td>
<td>3.04%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>11.91</td>
<td>971.54</td>
<td>1000.12</td>
<td>2.94%</td>
</tr>
<tr>
<td>13 weeks</td>
<td>0.0 mm</td>
<td>29.37</td>
<td>342.04</td>
<td>479.60</td>
<td>40.22%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>35.75</td>
<td>338.98</td>
<td>534.05</td>
<td>57.55%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>37.50</td>
<td>341.53</td>
<td>552.73</td>
<td>61.84%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>37.19</td>
<td>340.76</td>
<td>549.29</td>
<td>61.19%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>37.87</td>
<td>339.55</td>
<td>553.94</td>
<td>63.14%</td>
</tr>
<tr>
<td></td>
<td>2.5 mm</td>
<td>38.37</td>
<td>340.71</td>
<td>560.60</td>
<td>64.54%</td>
</tr>
<tr>
<td>16 weeks</td>
<td>0.0 mm</td>
<td>4.52</td>
<td>1010.54</td>
<td>1017.93</td>
<td>0.73%</td>
</tr>
<tr>
<td></td>
<td>0.5 mm</td>
<td>6.86</td>
<td>1012.19</td>
<td>1029.21</td>
<td>1.68%</td>
</tr>
<tr>
<td></td>
<td>1.0 mm</td>
<td>11.32</td>
<td>1009.11</td>
<td>1054.75</td>
<td>4.52%</td>
</tr>
<tr>
<td></td>
<td>1.5 mm</td>
<td>4.01</td>
<td>1007.26</td>
<td>1013.11</td>
<td>0.58%</td>
</tr>
<tr>
<td></td>
<td>2.0 mm</td>
<td>7.69</td>
<td>1011.85</td>
<td>1033.25</td>
<td>2.11%</td>
</tr>
</tbody>
</table>

Table 3.4. Moment of inertia data from fourth test of preliminary investigation.
3.4 Statistical Analysis of Preliminary Findings

The deviations of the minimum moment of inertia calculated about the principal axis from the actual bending axis were compared for all samples. Statistical analysis of all percent deviations indicated an average percent deviation of 7.58%, with a standard deviation of ±12.76%, and a 95% confidence interval (±1.96*standard error) of ±5.00%. This indicates that if the minimum moment of inertia is used for all test samples, it is expected that the percent deviation from the moment of inertia about the actual bending axis of 95% of the data points will be within the range of 2.58% and 12.58%. In other words, using the minimum moment of inertia for each sample would result in an underestimation of the moment of inertia about the actual bending axis of 2.58% to at most 12.58% (95% CI).

3.5 Error Analysis of Proposed Test Methods

Four investigators calculated the initial bone orientation using ImageJ for the same CT image for the same bone. The test method for Tests 2 and 3 calculated these orientations using the radio-opaque catheters that were glued to the bone. The concern was that the large size of the catheter in relation to the diameter of some of the smaller bones may be inaccurate. For the three repeated trials conducted by each of the four investigators, the standard deviation of the calculated angles was 13% of the average value, corresponding to ±5°. This indicated that an improvement could be made using the FARO arm for the orientation of the bone in the CT image.

Repeatability for the orientation measurement of the FARO arm was conducted by several measurements on each bone fixed in place so that the orientation was known
not to change. The results from this test show an error of at most $\pm 3.3^\circ$. The orientation measurements from the FARO arm before CT were compared to those taken after CT at the same orientation in order to determine if the bone moved in transit to the CT scanner. This error analysis showed a deviation of no more than $\pm 5^\circ$.

The repeatability of the ImageJ program and the MomentMacro installation for the program used to calculate the moment of inertia of each rib cross-section was also quantified. Moment of inertia calculations were performed ten times on the same CT image of the same bone at the same cross-section. This analysis indicated a standard deviation of 0, with 0% deviation among all 10 calculations. This indicates that the program is capable of repeated measurements.

The accuracy of the ImageJ program and MomentMacro was computed to determine the error of the program calculations from theoretical calculations. An ellipse with known major and minor axes was constructed, such that the minimum moment of inertia was known at all times to be the moment of inertia about the major axis. This can be calculated directly using the equation $I = \frac{\pi ab^3}{4}$, with $a$ being the major radius and $b$ being the minor radius. The image was then rotated from $0^\circ$ to $90^\circ$, every $10^\circ$. The MomentMacro was run in order to determine the minimum moment of inertia at each orientation. The calculated minimum moment of inertia was then compared against the theoretical calculation. The program yielded an error of approximately 3%, indicating that the minimum moment of inertia from the ImageJ calculation of the CT image is accurate to the theoretical value.
3.6 Conclusions from Preliminary Investigation

As presented throughout this document, the comparisons of the moment of inertia about the bending axis measured during testing to the moment of inertia about the minimum principal axis yielded relatively low percent deviations. The statistical analysis discussed above predicts that using the minimum moment of inertia for each test give error values for moment of inertia between 2.58% and 12.58% (95% confidence). It is further believed that better sample preparation will lower this error range. However, a few of the samples indicated unacceptable deviations between the actual bending axis moment of inertia and the minimum moment of inertia. As a result, there was still concern that data points may be lost if the actual orientation during testing was not known.

The data presented for each trial of preliminary testing indicates that there is an insignificant amount of rotation in the cross-section of the rib being loaded, after loading begins. As a result, the orientation of the rib after a small amount of deflection (for example 0.5mm), and while loading is still in the elastic region, can be taken as the orientation of the cross-section throughout the duration of testing. Although orientations do change after yielding, the bending axis after yielding is not used in the calculation of the modulus of elasticity, because it is outside the elastic region. Although the analysis above shows that this measurement is not necessary, if the orientation of each rib cross-section is to be determined for each test, it is suggested that only one measurement be taken at the beginning of loading in order to eliminate any possible changes in loading pattern as the test is stopped and restarted several times.
The error analysis presented in the previous paragraph indicates that the test setup and measurement techniques are accurate and repeatable. The only suggested change to the above test setup was to more accurately measure bone sections and trim samples to straighter sections. It was believed that these changes would decrease some of the higher percent deviations seen in the preliminary testing.

In summary, from the conclusions of the preliminary investigation, the following steps were implemented for the full testing of all samples in the study.

1. The actual bending axis was determined for each test sample in order to limit the error from the assumption that the rib will rotate to its principal axis of minimum moment of inertia.

2. Because minimal rotation occurs after testing is initiated, measurements for the bending axis orientation needed to only be taken once during testing.

3. Test samples were selected from the straightest section of each rib, ie, from the section with the least curvature.
CHAPTER 4

STUDY DESIGN

4.1 Subject Population

4.1.1 Porcine Subject Population

Due to the anticipated limited number of pediatric samples that would be available for the study, an animal model of porcine subjects was proposed in order to analyze the changes in material properties as a function of age. Porcine subjects have been used in numerous studies as predictors of pediatric response (Arbogast, 2005). All animals were raised by the same local farmer and processed at a local facility (Richwood Quality Meats; Richwood, Ohio). Because the subjects were not raised specifically for this study, and because processing was conducted at a regulated facility, the study was exempt from review by the Institutional Laboratory Animal Care and Use Committee (ILACUC).

For this study, it was desired to capture a large range of ages in order to trace the development of the rib bone in the porcine subjects. Porcine subjects were spaced one week apart from 1 week through 20 weeks of age. The selection of the desired ages is based on the plot in Figure 4.1 below. The plot indicates the rate of tissue accretion of muscle, fat, and bone. Because these values represent rates and not physical
measurements of the amount of tissue, a decline in the plot indicates that the rate of production has slowed down. The solid lines and the dashed lines indicate two different feeding regimes. However, by 200-250 pounds a pig is considered full-grown in the meat industry. This is the point where the rate of fat production becomes greater than the rate of muscle production. This weight corresponds to approximately six months of age.

![Graph showing the rate of tissue development in pigs as a function of weight and feed.](image_url)

Figure 4.1. Rate of tissue development in the pig as a function of weight and feed.


### 4.1.2 Pediatric Subject Population

The collection of pediatric rib specimens for this study was approved by the Institutional Review Board at Columbus Children’s Hospital (IRB submission # IRB05-00035; Columbus Children’s Hospital Federalwide Assurance # FWA00002860).
Consent for the samples was granted through the clause on the Columbus Children’s Hospital Expiration Record which states: “Tissue may be retained for diagnostic, teaching, and research purposes.” The patient population inclusion age was specified to be full-term infants through subjects of 21 years of age. The exclusion criteria were:

- Patients who experienced significant chest trauma (except CPR) prior to death
- Conditions such as rickets, osteogenesis imperfecta, osteoporosis, and any other genetic, metabolic, or acquired disease affecting bone structure or mineral content
- Patients receiving medication or treatment affecting bone density (ex: chronic steroid use)
- HIV/AIDS, Hepatitis B patients (infectious risk)

Each specimen was assigned a unique identifier and relevant subject information (age, gender, height, weight, and any condition which may have affected bone development) was recorded.

All pediatric specimens were removed by Pathology Assistants performing the autopsy at Columbus Children’s Hospital. The samples were collected between rib levels 5 through 8, as these levels have been used in previous studies (Cormier, 2005; Schultz, 1974; Stein and Granik, 1976; Sturtz, 1980; Takahashi and Frost, 1966; Theis, 1975; Yoganandan and Pintar, 1998). A total of 1-2 samples were removed from each hemithorax of each subject, for a total of 2-4 samples per subject. Figure 4.2 below shows the location for the removal of each specimen. In order to minimize the deviation from
autopsy protocol, samples were removed from the sternum medially to the mid-axillary line, along which the chest wall is cut for access to internal organs during the autopsy.

![Rib Sample Locations](image)

Figure 4.2. Location of rib sample removal from each subject.

All samples were frozen in saline-soaked gauze in the Pathology Department of Columbus Children’s Hospital until an investigator was available for collection.

4.2 Test Methods and Setup

4.2.1 Specimen Preparation

Pediatric rib samples were harvested from each subject by the Pathology Assistants at Columbus Children’s Hospital. As described previously, all samples were already of a length suitable for testing when collected. All excess tissue, including the periosteum, was removed from the bone prior to specimen storage. Each sample was wrapped in saline-soaked gauze to retain moisture, individually bagged and labeled with
a unique sample number, and frozen at -17°C until testing commenced. Freezing of bone has been shown to be an acceptable method of preservation, with limited to no changes in material properties (Pelker, 1984; Turner and Burr, 1993).

The porcine samples were harvested from the local processing facility as full ribcages. In order to take anatomical measurements and note any abnormalities, a CT scan of each ribcage was taken prior to dissection. Each ribcage was then separated into individual ribs (Figure 4.3 below). The spine, sternum, costal cartilage, and rib levels 1 through 4 were discarded. Rib levels 5 through 8 were stripped of excess tissue, wrapped in saline-soaked gauze to retain moisture, individually bagged and labeled with a unique sample number, and frozen at -17°C until testing commenced. As stated above, freezing of bone has been shown to be an acceptable method of preservation, with limited to no changes in material properties (Pelker, 1984; Turner and Burr, 1993).

Figure 4.3. Entire ribcages from porcine subjects were separated into individual ribs.
4.2.2 Test Protocol

Samples were removed from the freezer and defrosted overnight in a refrigerator. As discussed in Chapter 3, preliminary work deemed it necessary to accurately determine the actual bending axis orientation during testing. Because cross-sectional geometric properties were calculated from CT images (see Section 4.3.4), it was also necessary to know the exact orientation of the rib in the image. A line was drawn through the mid-span cross-section of each rib to mark the location at which the load would be applied. Two cross-hairs were drawn approximately 180° apart to denote reference locations for measurements. Each bone was then placed in a foam holder in order to rigidly fix the pre-test orientation of the rib with respect to the horizontal. Figure 3.7 is reproduced as Figure 4.4 below to show this technique.

![Location of cross-hair reference marker](image)

Figure 4.4. Foam holder for ensuring bone orientation was fixed during CT scan.
The foam holders were secured to a flat, horizontal surface and the differences in height between the two cross-hair markers were recorded. The FaroArm® used during preliminary testing was not available during the full-scale testing. Digital calipers were used to measure the straight-line distance between the two cross-hair markers, and a digital height gauge was used to determine the difference in heights between the two markers. A simple trigonometric calculation yielded the value for the orientation of these markers with respect to the horizontal (Figure 3.4 reproduced as Figure 4.5 below for reference). The foam holders were then secured to a rigid board such that the orientation angle with respect to the horizontal remained fixed throughout the CT scan process. Small segments of radio-opaque catheters were inserted into the foam material to mark the location of the cross-section to be loaded in the CT image. In order to ensure that the CT image cross-sections were as close together as possible, the highest possible resolution was used, producing images at 0.6mm intervals.
\[ \sin \theta = \frac{D}{h} \]

**Figure 4.5.** Calculation of angles of orientation with respect to the horizontal.

Rib samples were kept as moist as possible prior to and during the CT scan process. Each sample was again wrapped in saline-soaked gauze and placed in an individual bag to retain moisture. Three-point bend testing was conducted the day following the CT scan.

The three-point bend test fixture was designed with an adjustable span in order to accommodate the range of sizes of samples. All three points from which loads were applied were rounded to eliminate the induced shear that may result from a sharp point load. Additionally, the contacting areas of the supports were made of brass, a softer material, to further limit these effects. (See Figure 4.6.)
Figure 4.6. Schematic of the three-point bend test-fixture showing adjustable span and brass rod supports.

Prior to testing, the span of the three-point bend test fixture was adjusted to an appropriate distance for each test specimen in order to allow for beam overhang on each side of the test fixture. This span was recorded for later calculations. The top portion of the three-point bend fixture was screwed into the cross-head of an MTS Material Testing System® (MTS Systems Corporation; Eden Prairie, MN). The bottom portion with the adjustable span was set on the flat tabletop surface of the machine. Calipers were used to ensure that the top of the fixture would load the rib specimen at the exact mid-span. This setup loaded the top of the bone (external surface of the rib) in compression and the bottom (internal surface of the rib) in tension. This loading configuration simulates a single external force applied to the rib resulting in a bending moment. The test setup is shown in Figure 4.7.
A pre-load of 5 Newtons was applied to the bone to hold it in place. The difference in height between the two cross-hair markers was then determined from the digital height gauge and recorded. Load was applied to the bone at a quasi-static deflection rate of 2.5 millimeters/minute. This deflection rate has been used previously in the literature for three-point bending of small bone samples (Margulies and Thibault, 2000; Yoganandan and Pintar, 1998). The machine was programmed to stop at a deflection of 0.75 millimeters. The difference in height between the two cross-hair markers was again determined from the digital height gauge at this deflection. As discussed in Chapter 3, the results from the preliminary testing indicated that minimal rotation occurred after pre-load. The measurement of orientation at a deflection of 0.75
millimeters was taken in order to monitor any samples that did experience excessive rotation during testing. (Measurements were initially taken after 1.0 millimeters of deflection, but concern arose that some samples may reach yield before 1.0 millimeters of deflection. Therefore, the deflection point at which orientation measurements were taken during testing was changed to only 0.75mm of displacement.) After the orientation measurements at 0.75 millimeters of deflection were recorded, the sample was loaded at a deflection rate of 2.5 millimeters/minute until the recorded force value reached a level value, indicating that the yield point had been reached. As discussed in Chapter 3, force versus deflection data after the yield point was not necessary for calculating the material properties of interest. The force versus deflection data from the MTS was saved for analysis for each sample.

4.3 Data Analysis

The properties calculated for each test sample are the modulus of elasticity (Young’s modulus), the yield strength, and the maximum bending strength. Each was analyzed as a function of age, and comparisons were drawn between any trends seen between the porcine samples and the pediatric samples. Any apparent differences based on rib height were also noted. The calculations of each of these material properties, as well as detailed definitions of each variable are discussed below.
4.3.1 Calculation of Material Properties

The modulus of elasticity (Young’s modulus) is given by

\[ E = \left( \frac{F}{\omega} \right) \frac{L^3}{48I} \]  

(4.1)

where \( E \) is the modulus of elasticity, \( L \) is the span length of the test fixture, \( \left( \frac{F}{\omega} \right) \) is the stiffness, and \( I \) is the moment of inertia. This equation is derived from beam theory for a simply supported beam with a load at mid-span, and can readily be found in any Strength of Materials textbook (Stevens, 1987). The span length \( L \) was recorded during testing, as mentioned previously. The definition of \( \left( \frac{F}{\omega} \right) \) is given in section 4.3.2 below. The calculation of \( I \) from the cross-sectional CT images is presented in section 4.3.4 below.

The yield strength and maximum bending strength are both stress measurements. As a result, the values are calculated from the bending stress equation

\[ \sigma = \frac{My}{I} \]  

(4.2)

where \( \sigma \) is the bending stress, \( M \) is the bending moment, \( y \) is the distance from the centroid to the outermost fibers in the vertical direction, and \( I \) is the moment of inertia. From mechanics, the maximum moment occurs at the mid-span of the beam, and is given by the reaction force at the end of the beam multiplied by the distance from the reaction force to the maximum moment. If ‘\( F \)’ is the load applied to the mid-span of the beam, the reaction force at each end is equal to half of that or \( \frac{F}{2} \). The moment arm is equal to half of the span or \( \frac{L}{2} \). It follows from equation (4.2) that the bending stress is given by
\[ \sigma = \frac{My}{I} = \left( \frac{FL}{2} \right) \frac{y}{2} = \frac{FLy}{4Z} \]  

with all variables as defined previously and the section modulus \( Z \) equal to the moment of inertia \( I \) over the vertical distance \( y \). (See section 4.3.4 for discussion on how these cross-sectional values are determined.)

The difference between the yield strength and the maximum bending strength is the force used to calculate the moment from equation (4.3). For the yield strength, the value of \( F \) is the yield force. (See section 4.3.3 for discussion on how the yield point is determined.) The maximum bending strength uses the peak force from the force versus deflection curve in the stress equation. Maximum bending strength calculated in this way is referred to as the \textit{modulus of rupture} (Ugural and Fenster, 1995).

### 4.3.2 Linear Region

In order to determine the stiffness \( \left( \frac{F}{\omega} \right) \), the slope of the linear elastic region of the force versus deflection curve must be calculated. As the elastic region of bone is actually nonlinear, the slope was determined from the initial nearly linear region. A method was developed to provide a mathematical basis for determining the end of this region formulaically.

When the test was stopped at 0.75mm or 1.0mm deflection in order to take orientation measurements, there was a slight relaxation of force due to the viscoelastic properties of bone discussed previously (Cowin, 2001; Nordin and Frankel, 1989; Turner and Burr, 1993). The data points that can be attributable to this relaxation were removed.
from the plot in order to eliminate any artifact they may introduce to the slope calculation over this elastic region. The artifact points were defined as having a force value less than that at which the test was stopped.

After removal of the artifact points, force and deflection data were converted from negative to positive values for ease of calculations. The values are recorded as negative due to convention for compressive forces and downward displacements applied by the MTS cross-head. A force versus displacement curve was constructed for each sample. The average slope between the first two data points was calculated as the change in force over the change in displacement. The average slope over the first three data points was calculated as the absolute change in force for the three points over the change in displacement for the three points. This process was completed for each consecutive data point. At each point the mean and standard deviation of all preceding average slopes was calculated. **The end of the linear region was defined as the point at which the average slope calculated to that data point deviated above or below the mean of all preceding calculated average slopes by more than one standard deviation.** An example of this definition is shown in Figure 4.8 below for rib level 7R from the 20-week-old pig. The horizontal axis is the deflection in millimeters. The vertical axis values on the left side of the plot are force in Newtons, whereas the vertical axis values on the right side of the plot are average slope in Newtons per millimeter. The dark blue curve is the force versus deflection data recorded from the MTS, the pink curve is the average slope calculated from the first deflection data point to the deflection data point at which it is plotted, and the yellow and teal curves are the mean of all average slopes previously plus and minus the standard deviation, respectively.
Figure 4.8. Example of determination of linear region.

From this plot, the average slopes drop calculated at 0.5 millimeters of deflection falls below one standard deviation from the mean of all previously calculated average slopes. This determines the end of the linear region of the corresponding force versus deflection curve. A linear regression to this point gives the slope of this linearly elastic region to be 251.24 Newtons per millimeter.

In the event that the average slope never deviated above or below the mean of all preceding calculated average slopes by more than one standard deviation, the end of the
linear portion of the curve was determined to be at the start of the power fit. (See Section 4.3.3 below.) This was the case for two porcine samples and five pediatric samples.

4.3.3 Yield Point

Although previous studies have found cortical bone to fail at a strain of approximately 2% (Nordin and Frankel, 1989), it was unknown whether this strain value would also indicate failure of young pediatric bone. Other methods of a certain percentage offset yield (typically 0.2%) may not be valid for pediatric bone, as ultimate failure occurs after extensive plastic deformation. Additionally, characterization of strain in bending is vague. Formulas have been derived for the longitudinal strain in initially straight beams that undergo large enough deflections to induce curvature in the beam as a function of the new radius of curvature (Stevens 1987). However, the rib has an initial slight curvature and is being modeled as a straight beam. The beam will undergo significant strain by the point at which the applied load forces the rib into an actual straight beam. As a result, a method for determining the yield point was developed for this particularly setup. The yield point indicates the transition from the elastic region to plastic deformation. If the force is removed after the rib has been loaded past its yield point, the bone will not return to its initial position. Although not part of this investigation, the difference between the initial position and the position to which it returns is the permanent plastic deformation resulting from that load.

The method used to calculate the yield point is adapted from that developed by Datsko (as cited in Margulies and Thibault, 2000). The initial elastic region is fit with a linear function as described in the preceding paragraphs. The plastic region is fit with a
power law. Because the force versus deflection curves showed no definitive data point at which plastic deformation occurred, this power curve was fit from 10% below the maximum force to the maximum force. Linear fits for the elastic region and power fits for the plastic region were both determined using Microsoft Excel (Microsoft Corporation). The deflection at which these two curves intersected was used to identify the force that was the result of this deflection. *The force corresponding to the deflection at the intersection of the linear elastic curve and the power plastic curve was defined to be the yield force.* This force was used in equation (4.3) to determine the yield stress in bending. Yield stress could not be calculated for one pediatric sample because of extreme fluctuations in the force versus deflection data.

An example of this definition is shown in Figure 4.9 below for rib level 7R from the 20-week-old pig. The horizontal axis is the deflection in millimeters. The vertical axis is force in Newtons. This gives the curve as the force versus deflection data recorded from the MTS.
Figure 4.9. Example of determination of yield point.

From Figure 4.9, the maximum force is a value of 394 Newtons at 4.6 millimeters of deflection. The maximum force minus 10% of the maximum force gives a value of 355 Newtons. A power fit of this portion of the plastic region is shown in the figure. The intersection of the power curve with the linear regression for the linearly elastic region shown in Figure 4.8 occurs at 1.32 millimeters. This corresponding force on the force versus deflection plot for this deflection estimates the yield force to be 233 Newtons.
4.3.4 Geometric Cross-Sectional Analysis

The cross-sectional geometry of each rib was determined from CT images, as discussed previously. Although it is common practice to estimate cross-sectional area, moment of inertia, etc., by assuming the bone is a perfect ellipse and using measurements of “major” and “minor” axes in derived formulas, the use of computer analysis of the CT images was believed to provide a more accurate calculation of these values. Also, because each bone was scanned prior to testing, the images obtained represented the exact cross-section to be loaded in its original state. If the ribs were dissected after testing to take cross-sectional measurements, any changes in cross-section due to loading would not allow for accurate geometric values.

The program ImageJ (version 1.36b), a free development of NIH (National Institutes of Health) and the macro “MomentMacroJ v1.2,” written by Dr. Chris Ruff of Johns Hopkins University, were used for the geometric analysis in this study. The MomentMacro utilizes the parallel-axis theorem for calculation of the moment of inertia for irregular cross-sections. This theorem defines the moment of inertia as

\[ I = I_c + Ad^2 \]  \hspace{1cm} (4.4)

where \( I \) is the total moment of inertia of the cross-section, \( I_c \) is the moment of inertia about the centroid axis, \( A \) is the area for an element of area within the cross-section, and \( d \) is the distance from the centroid of the element of area to a given axis on the cross-section. For each CT image, the “element of area” is one pixel in the image. The area of the field of view of the image determines the resolution of the pixel. Each image is 512 pixels by 512 pixels. Due to the small size of the pixels relative to the bone cross-section,
the entire cross-section could easily be divided into these small units to accurately calculate the moment of inertia.

The MomentMacro for ImageJ works with a thresholding technique. Each pixel is assigned a grayscale value for what shade of gray it displays on a black and white scale. These values range from 0 to 255. The thresholds can be adjusted to set the range of grayscale values to be included in the calculation of values for a particular cross-section. Visual inspection and adjustments to the grayscale values were used to select the entire cortical bone region for each rib, without selecting any adjacent cancellous bone or bone marrow. The region of interest was selected for each test sample, and the MomentMacro calculated the minimum moment of inertia, the moment of inertia about the bending axis (I_x), the cross-sectional area, the maximum radii, and other geometrical properties for this selection.

Prior to running the MomentMacro, the difference in orientation from the bone position during CT to the bone position during testing was determined. Each image was then rotated its respective difference in orientation in order for the horizontal axis of the image to correspond to the bending axis during testing. In this way, the value for I_x calculated from ImageJ was the moment of inertia about the actual bending axis. **The value for the moment of inertia about the actual bending axis, I_x, was used as the moment of inertia value in all calculations.** The value for I_min, as well as the angle between the horizontal axis and the principal axis of the minimum moment of inertia, were determined by the MomentMacro.
4.3.5 Statistical Analysis

Each variable of interest was plotted as a function of age, size, etc., for both pediatric samples and porcine samples. Any trends seen between variables were noted. Additionally, Microsoft Excel was used to fit regressions to each of these plots and regression coefficients were obtained. The behavior of each trend (ie, linear, logarithmic, or power functions) was determined by the fit with the best regression coefficient (closest to 1.0). Where appropriate, mean and standard error values were reported to indicate the scatter of the data among subjects.
CHAPTER 5

TEST RESULTS

5.1 Subject Population

5.1.1 Porcine Subject Population

A total of 27 porcine subjects were collected. One subject was harvested at each week from week 1 through week 20 of age, with two subjects from each week between 5 weeks and 11 weeks. One animal at 5 weeks of age and one animal at 6 weeks of age were not used for all analyses, due to lost data on cross-sectional geometry. A total of six rib samples were retained from each subject – rib levels 5, 6, and 7 from both sides of the thorax. As stated earlier, these levels were selected to correspond with previous human rib studies in the literature (Cormier, 2005; Schultz, 1974; Stein and Granik, 1976; Sturtz, 1980; Takahashi and Frost, 1966; Theis, 1975; Yoganandan and Pintar, 1998).

5.1.2 Pediatric Subject Population

A total of 32 samples from 9 different subjects were collected over the duration of the study. Table 5.1 below shows each of these subjects, along with subject age, rib level, hemi-thorax location, subject height, and subject weight (when available).
<table>
<thead>
<tr>
<th>Subject #</th>
<th>Age</th>
<th>Gender</th>
<th>Rib Level</th>
<th>Location</th>
<th>Height</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>11.5 months</td>
<td>Female</td>
<td>6</td>
<td>L</td>
<td>58.7 cm</td>
<td>6.32 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>7</td>
<td>L</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>7</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>4 years</td>
<td>Female</td>
<td>6</td>
<td>L</td>
<td>90.0 cm</td>
<td>11.3 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>7</td>
<td>L</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>7</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>1 day</td>
<td>Female</td>
<td>unknown</td>
<td>unknown</td>
<td>28.0 cm</td>
<td>1.78 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>unknown</td>
<td>unknown</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>15 months</td>
<td>Female</td>
<td>6</td>
<td>L</td>
<td>75.2 cm</td>
<td>unknown</td>
</tr>
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<td></td>
<td></td>
<td></td>
<td>7</td>
<td>L</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>7</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>6 years</td>
<td>Male</td>
<td>5</td>
<td>L</td>
<td>113.0 cm</td>
<td>unknown</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
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<tr>
<td>6</td>
<td>5 months (27 wks gestation)</td>
<td>Unknown</td>
<td>6</td>
<td>L</td>
<td>44.3 cm</td>
<td>2.52 kg</td>
</tr>
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<td></td>
<td></td>
<td>7</td>
<td>L</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>7</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td>7</td>
<td>2 years</td>
<td>Female</td>
<td>6</td>
<td>L</td>
<td>85.0 cm</td>
<td>13.6 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>5</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>1 day (39 wks gestation)</td>
<td>Female</td>
<td>5</td>
<td>L</td>
<td>50.6 cm</td>
<td>2.95 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>L</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>5</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>1 year, 9 months</td>
<td>Female</td>
<td>5</td>
<td>L</td>
<td>71.8 cm</td>
<td>7.79 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>L</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>5</td>
<td>R</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6</td>
<td>R</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 5.1. Pediatric subject information.

5.2 Failure Types

All test samples were tested until the load surpassed the yield point, and not necessarily until ultimate failure. Only one porcine sample, level 5 on the right side of the 19-week-old pig sustained a fracture. The failure occurred as expected on the surface of the rib loaded in tension, causing the cortex to crack. (See Figure 5.1 below).
Figure 5.1. Failure on the tensile surface of rib level 5R from the 19-week-old pig.

The rib samples from one of the pediatric subjects, PedRib#6, showed unexpected fractures. The bone failed in shear longitudinally through the cross-section of the beam. This subject was 5 months old at death, but was born at only 27 weeks gestation. The samples were among the smallest tested in the current test fixture, and had the lowest overhang of bone off of the end supports as a percentage of the total span. The bones displayed signs of significant flexibility during testing. Figure 5.2 below shows the failure that occurred in rib level 6 of the 5-month-old subject.
Figur 5.2. Longitudinal shear failure of a rib from the 5-month-old subject.

Rib level 7 on the right side of the 4-year-old subject, PedRib#2, showed no signs of plastic deformation prior to fracture. The force versus deflection curve for this sample is shown in Figure 5.3. The following picture, Figure 5.4, depicts the failure mode.
Figure 5.3. Force versus deflection plot of rib 7R from the 4-year-old subject.

Figure 5.4. Failure on the tensile surface of rib level 7R from the 4-year-old child.
Rib level 6 on the right side from subject PedRib#1 (11.5 months) failed due to the 5 Newton preload prior to commencing testing. The failure mode indicated buckling.

Figure 5.5. Buckling failure of rib level 6R from the 11.5-month-old child.

As stated previously, samples were loaded only until after yielding had occurred, as determined by viewing the force data during the test. The result was that most samples were not tested to ultimate failure, with the exception of those discussed in detail above. Of note is that most samples displayed deformations due to the loading nose on the compressive side of the rib. This deformation indicates plastic failure due to compression. This plastic compressive deformation is an interesting result, as failure was expected on the tensile surface. Figure 5.6 shows this deformation on a representative sample.
5.3 Rotation during Testing

Measurements taken for the orientation of the loaded cross-section for each rib sample provide an estimation of the amount of rotation within the cross-section over a deflection of 0.75 millimeters. The amount of rotation is defined as the absolute value of the difference between the orientation angle at pre-load (0 millimeters of deflection) and the orientation angle at 0.75 millimeters of deflection.

For the porcine rib samples, the average rotation between the two measurements was 2.6°, with a standard deviation of 3.0°, and a 95% confidence interval (±1.96*standard error) of (2.1°, 3.1°). One porcine rib sample (level 7R from the 16-week-old pig) had a rotation of 26.1°, but all other values were at 10° or less. This rib rotated from an orientation of 13.5° to −12.6°, resulting in only a 2% change in the moment of inertia about the bending axis.
The pediatric rib sample data set had an average rotation of 6.23°, with a standard deviation of 8.09°, and a 95% confidence interval of (3.12°, 9.34°). The maximum rotation was 41.8° for level 5L of the 21-month-old child, resulting in a 21% change in the moment of inertia about the bending axis. The rotation of all other ribs was at 15° or less.

5.4 Moment of Inertia Comparisons

The preliminary investigation of the orientation of the bending axis indicated a need to measure the exact bending axis during testing, as some samples did not rotate to the principal axis of the minimum moment of inertia. Although the values for the actual bending axis of each sample were determined, it was still of interest to analyze the deviations between these two axes. Data is presented separately for porcine samples and pediatric samples. Table 5.2 summarizes the statistical analysis of these findings. Two data sets are presented for the porcine samples and two data sets are presented for the pediatric samples. The first data set for each uses the deviations between the values at the two axes, as calculated from orientation at pre-load. The second data set for each uses the values calculated from orientation at 0.75 millimeters deflection.
<table>
<thead>
<tr>
<th></th>
<th>Porcine - Orientations at Pre load</th>
<th>Porcine - Orientations at 0.75mm Deflection</th>
<th>Pediatric - Orientations at Pre load</th>
<th>Pediatric - Orientations at 0.75mm Deflection</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average % Deviations</td>
<td>9.97%</td>
<td>9.60%</td>
<td>29.42%</td>
<td>33.70%</td>
</tr>
<tr>
<td>Standard Deviation of % Deviations</td>
<td>13.16%</td>
<td>13.17%</td>
<td>30.85%</td>
<td>37.86%</td>
</tr>
<tr>
<td>95% CI</td>
<td>(7.86%, 12.08%)</td>
<td>(7.49%, 11.71%)</td>
<td>(18.38%, 40.46%)</td>
<td>(19.92%, 47.48%)</td>
</tr>
</tbody>
</table>

Table 5.2. Statistical analysis of deviations between values of moment of inertia about actual bending axis compared to the minimum moment of inertia about the principal axis.

For the porcine specimens, statistical analysis of all percent deviations at pre-load orientations indicated an average percent deviation of 9.97%, with a standard deviation of ±13.16%, and a 95% confidence interval (±1.96*standard error) of ±2.11%. The statistical analysis indicates that the percent deviation from the moment of inertia about the actual bending axis of 95% of the data points will be within the range of 7.86% and 12.08%. In other words, the minimum moment of inertia for each sample underestimated the moment of inertia about the actual bending axis by 7.86% to at most 12.08% (95% CI). See Table 5.2 for a summary of all results.

The pediatric samples had larger deviations between the principal axes and the minimum moment of inertia. Statistical analysis of all percent deviations at pre-load orientations indicated an average percent deviation of 29.42%, with a standard deviation of ±30.85%, and a 95% confidence interval (±1.96*standard error) of ±11.04%. The statistical analysis indicates that the percent deviation from the moment of inertia about the actual bending axis of 95% of the data points will be within the range of 18.38% and...
40.46%. In other words, the minimum moment of inertia for each sample underestimated the moment of inertia about the actual bending axis by 18.38% to at most 40.46% (95% CI). See Table 5.2 for a summary of all results.

The change in moment of inertia as a function of age was also investigated. The calculated minimum moment of inertia values for all porcine subjects are plotted in Figure 5.7. Average values at each age pig, along with 95% confidence intervals, are presented in Figure 5.8. The same plots for the moment of inertia about the measured bending axis are below as Figures 5.9 and 5.10 for the porcine samples. The pediatric data follows as Figures 5.11 through 5.14. Red data points indicate rib level 5, orange data points indicate rib level 6, and blue data points indicate rib level 7.
Figure 5.7. The rib minimum moment of inertia versus age for all porcine samples.

From Figure 5.7, a linear regression through the data points results in a relationship of

\[ I_{\text{min}} = 51.33 \text{ (porcine age)} - 205.25 \]

with a regression coefficient of \( R^2 = 0.64 \).
Figure 5.8. The average rib minimum moment of inertia at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.7 for the fit through all data points.
Figure 5.9. The rib moment of inertia about the measured bending axis versus age for all porcine samples.

From Figure 5.9, a linear regression through the data points results in a relationship of

\[ I_{xx} = 56.28 \text{ (porcine age)} - 223.50 \]

with a regression coefficient of \( R^2 = 0.63 \).
Figure 5.10. The average rib moment of inertia about the measured bending axis at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.9 for the fit through all data points.
Figure 5.11. The rib minimum moment of inertia versus age for all pediatric samples.

From Figure 5.11, a linear regression through the data points results in a relationship of

\[ I_{\text{min}} = 0.64 \times \text{(pediatric age)} + 1.57 \]

with a regression coefficient of \( R^2 = 0.65 \).
Figure 5.12. The average rib minimum moment of inertia at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.11 for the fit through all data points.
Figure 5.13. The rib moment of inertia about the measured bending axis versus age for the pediatric samples.

From Figure 5.13, a linear regression through the data points results in a relationship of 
\[ I_{xx} = 0.78 \text{ (pediatric age)} + 2.73 \] with a regression coefficient of \( R^2 = 0.62 \).
Figure 5.14. The average rib moment of inertia about the measured bending axis at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.13 for the fit through all data points.

The above plots show an increase in both the rib minimum moment of inertia ($I_{min}$) and the rib moment of inertia about the horizontal bending axis ($I_{ho}$) as a function of porcine age. This is also true for the pediatric rib samples.
The variation in the data presented here in Section 5.4 for the differences between the minimum moment of inertia about the principal axis and the moment of inertia about the measured bending axis indicated that the use of the moment of inertia about the measured bending axis would provide more accurate results. Also, the results in Section 5.3 on rotation of the rib cross-section during testing show that minimal rotation occurred in a majority of the samples. Because of these findings, the variation in the material properties calculated using the bending axes at the two orientations measured during testing was determined to be insignificant. When relevant, all data in the following sections requiring use of a geometric property about the bending axis was calculated using the orientation of the bending axis at preload for consistency.

5.5 Stiffness

The values for stiffness, the slope of the linear region of the force versus deflection curve, are presented below. The vertical axis is the slope value in N/mm, and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.15 and 5.16 present the porcine data, and Figures 5.17 and 5.18 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.15. Stiffness values versus age for all porcine samples.

From Figure 5.15, a linear regression through the data points results in a relationship of

\[ \text{Stiffness} = 14.50 \times \text{(porcine age)} - 11.16 \], with an \( R^2 \) value of 0.60.
Figure 5.16. The average stiffness values at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.15 for the fit through all data points.
Figure 5.17. Stiffness values versus age for all pediatric samples.

A power fit through the data points of Figure 5.17 (excluding subjects less than 1 day of age) results in a relationship of \( \text{Stiffness} = 0.16(\text{pediatric age})^{1.4} \), with an \( R^2 \) value of 0.63. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.18. The average stiffness values at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.17 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.6 Young’s Modulus

The calculated values for Young’s modulus of the rib bone as a function of age are plotted below for both the porcine samples and the pediatric samples. The Young’s modulus, or elastic modulus, is a property indicating the extent of the elastic behavior of a material. (See Section 4.3.1 for further information.) The vertical axis is the elastic modulus in MPa, and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.19 and 5.20 present the porcine data, and Figures 5.21 and 5.22 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
A linear regression through the data points results in a relationship of

\[ E = -32.06(\text{porcine age}) + 1942.65 \], with an \( R^2 \) value of 0.06. Most data points lie within the range of 1000 to 2000 MPa. However, elastic modulus values for weeks 7 through 10 peak at over 3000 MPa, whereas all data points for weeks 13 and 14 are below 700 MPa.
Figure 5.20. The average Young’s modulus values at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression (diagonal line) on the plot is reproduced from Figure 5.19 for the fit through all data points. The average of all data points over all ages is plotted as a constant at 1610 N/mm.
Figure 5.21. Young’s modulus values versus age for all pediatric samples.

A power fit through the data points (excluding subjects less than 1 day of age) results in a relationship \( \text{Young's Modulus} = 46.74 \times (\text{pediatric age})^{1.21} \), with an \( R^2 \) value of 0.71. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot as a dashed line. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.22. The average stiffness values at each age plotted versus age for the pediatric samples. Error bars indicate +/- 95% confidence intervals.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.21 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.7 Flexural Rigidity (EI)

The quantity EI, known as the flexural rigidity, is plotted as a function of age for both porcine and pediatric specimens below. The flexural rigidity is a sample-dependent property that measures the sample’s resistance to bending. The vertical axis is the flexural rigidity in Nm², and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.23 and 5.24 present the porcine data, and Figures 5.25 and 5.26 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.23. Flexural rigidity values versus age for all porcine samples.

A linear regression through the data points results in a relationship of

\[ EI = 0.06(\text{porcine age}) - 0.19 \]

with an \( R^2 \) value of 0.55.
Figure 5.24. The average flexural rigidity values at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.23 for the fit through all data points.
Figure 5.25. Flexural rigidity values versus age for all pediatric samples.

A power fit through the data points (excluding subjects less than 1 day of age) results in a relationship $EI = 0.00002(pediatric\ age)^{2.37}$, with an $R^2$ value of 0.80. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot as a dashed line. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.26. The average flexural rigidity values at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.25 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.8 Peak Force

The peak force is plotted as a function of age for both porcine and pediatric specimens below. The vertical axis is the peak force in N and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.27 and 5.28 present the porcine data, and Figures 5.29 and 5.30 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.27. Peak force in bending versus age for all porcine samples.

A linear regression through the data points results in a relationship of

\[ F_{\text{peak}} = 24.16 \text{ (porcine age)} - 44.18, \text{ with an } R^2 \text{ value of 0.64.} \]
Figure 5.28. The average peak force in bending at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.27 for the fit through all data points.
Figure 5.29. Peak force in bending versus age for all pediatric samples.

A power fit through the data points (excluding subjects less than 1 day of age) results in a relationship \( F_{\text{peak}} = 0.63 \, (\text{pediatric age})^{1.31} \), with an \( R^2 \) value of 0.68. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot as a dashed line. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.30. The average peak force in bending at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.29 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.9 Rupture Modulus/Bending Strength

The rupture modulus, also known as the bending strength (Section 4.3.1), of each rib is plotted as a function of age for both porcine and pediatric specimens below. The vertical axis is the bending strength in MPa, and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.31 and 5.32 present the porcine data, and Figures 5.33 and 5.34 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.31. Rib bending strength versus age for all porcine samples.

A linear regression through the data points results in a relationship of

$$\sigma_{\text{rupt}} = -0.89(\text{porcine age}) + 58.30$$, with an $R^2$ value of 0.11.
Figure 5.32. The average bending strength at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression (diagonal line) on the plot is reproduced from Figure 5.31 for the fit through all data points. The average of all data points over all ages is plotted as a constant at 49.27 MPa.
Figure 5.33. Bending strength versus age for all pediatric samples.

A power fit through the data points (excluding subjects less than 1 day of age) results in a relationship $\sigma_{\text{rupt}} = 6.08(\text{pediatric age})^{0.72}$, with an $R^2$ value of 0.67. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot as a dashed line. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.34. The average bending strength at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.29 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.10 Yield Force

The yield force, or the force at which the material behavior changes from elastic to plastic, was calculated for each sample by the method explained in detail in Section 4.3.3. The yield force values are plotted as a function of age for both porcine and pediatric specimens below. The vertical axis is the yield force in N, and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.35 and 5.36 present the porcine data, and Figures 5.37 and 5.38 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.35. Yield force in bending versus age for all porcine samples.

A linear regression through the data points results in a relationship of

\[ F_{\text{yield}} = 18.64 \ (\text{porcine age}) - 30.54, \] with an \( R^2 \) value of 0.59.
Figure 5.36. The average yield force in bending at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression on the plot is reproduced from Figure 5.35 for the fit through all data points.
Figure 5.37. Yield force in bending versus age for all pediatric samples.

A power fit through the data points (excluding subjects less than 1 day of age) results in a relationship $F_{yield} = 0.50 \text{ (pediatric age)}^{1.35}$, with an $R^2$ value of 0.69. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot as a dashed line. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.38. The average yield force in bending at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.37 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.11 Yield Stress in Bending

The yield stress was calculated for each sample from the force at yield (Section 5.10), and is plotted as a function of age for both porcine and pediatric specimens below. The vertical axis is the yield stress in MPa, and the horizontal axis is porcine age in weeks or pediatric age in months. Figures 5.39 and 5.40 present the porcine data, and Figures 5.41 and 5.42 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.39. Rib yield stress in bending versus age for all porcine samples.

A linear regression through the data points results in a relationship of

\[ \sigma_{yield} = -0.92(\text{porcine age}) + 49.24 \], with an \( R^2 \) value of 0.14.
Figure 5.40. The average yield stress in bending at each age plotted versus age for the porcine samples.

Error bars indicate +/- 95% confidence intervals. The linear regression (diagonal line) on the plot is reproduced from Figure 5.39 for the fit through all data points. The average of all data points over all ages is plotted as a constant at 39.81 MPa.
Figure 5.41. Yield stress in bending versus age for all pediatric samples.

A power fit through the data points (excluding subjects less than 1 day of age) results in a relationship \( \sigma_{yield} = 4.83 \, (\text{pediatric age})^{0.76} \), with an \( R^2 \) value of 0.66. A linear regression through these data points (excluding subjects less than 1 day of age) is also depicted on the plot as a dashed line. However, as discussed in Section 4.3.5, the fit with the best regression coefficient was defined to best fit the data. The power fit was used for analysis presented later.
Figure 5.42. The average yield stress in bending at each age plotted versus age for the pediatric samples.

Error bars indicate +/- 95% confidence intervals. The power fit and the linear regression on the plot are reproduced from Figure 5.42 for the fit through all data points (excludes subjects less than 1 day of age, which are shown in green on the plot).
5.12 Bending Strength versus Young's Modulus

The calculated bending strength (Section 5.9) was plotted versus the calculated Young's modulus (Section 5.6) for both porcine and pediatric data sets. The results are below as Figures 5.43 and 5.44. The vertical axis is the rupture modulus/bending strength in MPa, and the horizontal axis is Young's Modulus in MPa. Figures 5.39 and 5.40 present the porcine data, and Figures 5.41 and 5.42 present the pediatric data. The values for the two pediatric subjects at less than one day of age are presented in each of the pediatric plots below. However, it was noted during testing that the available length for the sample span was relatively small in comparison to the width of the loading nose. As a result, there was concern that this sample was loaded in compression and not pure bending for our test setup. Accuracy of the data obtained, as it relates to bending, was questioned. The values for the pediatric subjects of less than one day of age are not included in the regressions on each of the pediatric plots below because of this concern.
Figure 5.43. Rib bending strength versus Young’s modulus for all porcine samples.
Figure 5.44. Rib bending strength versus Young’s modulus for all pediatric samples.

As expected, the porcine data set of Figure 5.43 centers around a linear increase in bending strength with increasing Young’s modulus. The pediatric data set of Figure 5.44 also shows this trend. Visual inspection identifies one possible outlying data point is that of rib level 7R from the 4-year-old subject. This point is highlighted in the figure.
CHAPTER 6

DISCUSSION OF RESULTS

6.1 Significance of Results

6.1.1 Rotation during Testing

The data presented in Section 5.3 indicates minimal rotation in the rib cross-section between pre-load and a deflection of 0.75 millimeters. Repeated here for comparison, the 95% confidence interval for the porcine samples was (2.1°, 3.1°). The 95% confidence interval for the pediatric samples was (3.12°, 9.34°). These values predict that most samples will rotate less than 10° of rotation. *The results indicate that the bending axis measured as the horizontal axis after a preload is acceptable for calculations through the yield point.* As no orientation measurements were taken after the yield point, and yielding causes permanent deformation and changes in the stress distribution, it cannot be assumed that the previous statement is valid for calculations after yielding.

For the one porcine sample that rotated a total of 26.1° from preload to the second orientation measurement, the bending axis rotation occurred through the principal axis of the minimum moment of inertia. This lead to minimal effects on the calculated moment of inertia about the bending axis, because the overall distribution of material from the
horizontal bending axis was minimal. The shape of the rib with respect to the horizontal experienced little change.

6.1.2 Moment of Inertia

The values given in Table 5.2 show the 95% confidence intervals for percent deviations between the values of the minimum moment of inertia calculated about the principal axis and the value of the moment of inertia about the measured bending axis. For both the porcine samples and the pediatric samples, the ranges are similar at pre-load as compared to at 0.75 millimeters of deflection ((7.86%, 12.08%) compared to (7.49%, 11.71%) for the porcine, and (18.38%, 40.46%) compared to (19.92%, 47.48%) for the pediatric). The ranges for the 95% confidence intervals of the pediatric samples are larger than anticipated from the results of the preliminary investigation. The statistics indicate that the use of the minimum moment of inertia about the principal axis in calculation of material properties may underestimate the actual moment of inertia about the bending axis by upwards of 40%. Additionally, these results indicate an expectation that the minimum moment of inertia underestimates the actual bending axis moment of inertia by at least 19% for 95% of the data points. These variations are not acceptable for calculating material properties of the rib bone. The results indicate a definite need to measure the actual bending axis orientation when conducting bending tests on rib samples.

From Figures 5.7 and 5.9, the changes in the minimum moment of inertia with increasing porcine age are comparable to those for the moment of inertia values about the horizontal axis. Similar relationships are seen for the slopes of the plots for the pediatric
samples in Figures 5.11 and 5.13. If the linear regression equations for the moment of inertia versus porcine age are set equal to the linear regression equations for the moment of inertia versus pediatric age, criteria for selecting the equivalent age of porcine subjects for pediatric subjects may be developed. Using the equations for the minimum moment of inertia, the relationship is: \((\text{porcine age}) = 0.0125 \times (\text{pediatric age}) + 4.03\).

If the bending axis moment of inertia equations are used, the result is similar:

\((\text{porcine age}) = 0.0139 \times (\text{pediatric age}) + 4.02\). The two equations are plotted in Figure 6.1 below. The blue curve is derived from the minimum moment of inertia values, whereas the pink curve is derived from the moment of inertia values about the measured bending axis.
Figure 6.1. Plot of porcine age as a function of pediatric age, as determined by comparing measured moment of inertia values.

Figure 6.1 is plotted from pediatric age of 0 months through 120 months (10 years). The equations obtained above indicate that the moment of inertia of the pediatric rib within this age range are predicted by porcine subjects aged 4- to 6-weeks only. The tight range of porcine ages indicates that the rate of development for the pig is drastically greater than that of the human.

The plot in Figure 5.8 for the 95% confidence intervals of the porcine minimum moment of inertia values shows that the linear regression trendline does fall within this
interval for most of the pediatric samples. If the 95% confidence intervals are used as an
indication of the range of values, whether or not the trendline passes through this interval
provides information on how well the regression will predict the actual behavior. In the
porcine samples, weeks 1 through 4, week 14, and week 16 are underestimated by the
trendline. Weeks 6 through 8, week 10, week 12, and week 19 are overestimated by the
trendline. The same is true for the plot in Figure 5.10 for the bending axis moment of
inertia, with week 17 also being overestimated by the trendline.

The plot in Figure 5.12 shows that a linear regression trendline for the pediatric
minimum moment of inertia overestimates the values for the 5-month-old and the 4-year-
old subjects. The minimum moment of inertia values for the two-year-old subject is
underestimated by the trendline. The trendline for the bending axis moment of inertia
versus pediatric age overestimates the values for the 5-month-old, 11.5-month-old, and 4-
year-old subjects. The bending axis moment of inertia for the 6-year-old subject is
underestimated.

Visual inspection of the plots indicates a slight trend for rib level dependence.
Many data points show slightly higher moment of inertia values for rib level 6, as
compared to rib level 5 and rib level 7, for both the porcine and pediatric subjects.
However, no statistical analysis was conducted on rib-level dependence.

6.1.3 Stiffness

When set equal and solved for porcine age as a function of pediatric age, the
regressions presented in Figures 5.15 and 5.17 can also be analyzed for how well
pediatric rib behavior can be predicted by porcine rib testing. The resulting relationship is
(porcine age) = 0.0108(pediatric age)^{1.4623} + 0.770. This result is plotted in Figure 6.2 below.

Figure 6.2. Plot of porcine age as a function of pediatric age, as determined by comparing measured stiffness values.

As with Figure 6.1, the plot in Figure 6.2 also predicts a range of porcine ages that will predict behavior of the pediatric rib. Porcine age of less than one week through approximately 12 weeks of age estimate the behavior of pediatric ages 0 through 10
years. Increase in stiffness is anticipated with increase in age, as more mineral is deposited in bone. Because the mineral is a stiffer material than the matrix, increasing the mineral content is anticipated to also increase the stiffness values from the force versus deflection response. Also, increasing size with age will increase the stiffness response.

The regression of stiffness values versus porcine age (Figure 5.16) passes through several of the calculated 95% confidence intervals at each age. However, although the power fit regression of stiffness values versus pediatric age (Figure 5.18) had the largest regression coefficient compared to a linear or logarithmic fit, the trendline fails to capture the 95% confidence intervals of the samples.

Visual inspection of the plots for rib level dependence suggests that stiffness values may vary slightly with rib level. Anatomically lower rib levels (higher value for rib level) frequently display larger stiffness values than the higher rib levels. However, no statistical analysis was conducted on rib-level dependence.

6.1.4 Young’s Modulus

A plot of Young’s modulus as a function of porcine age (Figures 5.19 and 5.20) did not yield results as expected. It was anticipated that older porcine subjects would have an increased mineral content, thereby increasing the stiffness and increasing Young’s modulus of the bone. However, the general linear regression trendline indicates an overall decrease with age. The analysis presented in Figure 5.20 shows high average values for ages 6 weeks through 10 weeks, and low average values for ages 13-16 weeks, 18 weeks, and 20 weeks. The average Young’s modulus value of all samples for all ages was calculated to be 1610 MPa. The average value passes through the 95% confidence
interval of more ages than the linear trendline. Weeks 5 and 6 and weeks 8 through 10 show intervals all higher than the average, whereas weeks 11, 13, 14, 16, and 18 are lower than the average. It appears that the average value better represents the data for Young’s modulus as a function of porcine age, indicating that the elasticity of the porcine rib does not change significantly from one week of age through twenty weeks of age. Values from additional subjects at each age are needed to further investigate this conclusion.

Young’s modulus versus pediatric age (Figures 5.21 and 5.22) is best represented by a power fit through the data points, as determined by the regression coefficient. Although regression analysis found a power function to be the best fit for the data through 6 years of age, it is not expected that Young’s modulus would continue to increase as a power function through adulthood. It is anticipated that a logarithmic trend may better represent development through adulthood. As indicated in Chapter 5, the data from the subjects less than one day of age were removed from the regression due to concerns about whether or not the sample was loaded in compression instead of pure bending. Calculations of Young’s modulus from this force versus deflection plot would lead to invalid results. Bone is stronger in compression than in tension or bending and the force versus deflection response would be altered. Additionally, equation (4.1) for calculating Young’s modulus is only valid for pure bending. Due to the analysis of Young’s modulus versus porcine age as a constant value as discussed above, no comparison of the porcine age predicted from pediatric age is made for the Young’s modulus values.
Young’s modulus may vary with rib level. Although the results vary, several porcine subjects have larger Young’s modulus values for rib levels 5 and 6 as compared to rib level 7. Although limited in data points, pediatric subjects displayed slightly larger values for Young’s modulus for anatomically lower rib level. However, no statistical analysis was conducted on rib-level dependence.

6.1.5 Flexural Rigidity (EI)

The flexural rigidity of a sample is a measure of its resistance to bending. Figures 5.23 and 5.24 indicate a linear increase in flexural rigidity with increasing porcine age. Compared to the 95% confidence interval at each age, the linear regression fits the 95% confidence interval at all ages from 7 weeks through 19 weeks, except for weeks 13, 14, and 16. Although a linear regression best fits all porcine data points, the increasing trend from 1 through 6 weeks of age may represent exponential increase in EI as a function of age. The pediatric data of Figures 5.25 and 5.26 is best fit with a power function (as determined by $R^2$ value), although this trend fails to pass through the 95% confidence interval of most data points. (Again data for the subjects less than one day of age is eliminated from the regression analysis for the concerns discussed previously.)

If the linear regression for the porcine data is set equal to the power fit for the pediatric data, the age relationship that results is:

$(\text{porcine age}) = 0.000309 (\text{pediatric age})^{2.3705} + 3.0$. This equation is plotted as Figure 6.3 below.
Figure 6.3. Plot of porcine age as a function of pediatric age, as determined by comparing measured flexural rigidity values.

The result of Figure 6.3 predicts an increasing function of porcine age as a function of pediatric age by a power of 2.3705. From this plot, a 10-year-old child is represented by a 30-week-old pig. This is in contrast to the relationship developed from the moment of inertia regressions. In Figure 6.1, the moment of inertia of the rib of a 10-year-old child is predicted to be equivalent to that of a less than 6-week-old pig. The relationship in Figure 6.2 predicts the stiffness of a 10-year-old child as a 12.5-week-old pig. Figure 6.3 estimates a newborn human infant at a 3-week-old pig.
For the porcine samples, peak flexural rigidity values occur most often in rib level 7. Also, anatomically lower rib levels of the pediatric subjects tend to display larger values for flexural rigidity. Significance of these trends are not known, due to no statistical analysis conducted on the limited sample size.

6.1.6 Peak Force

The peak force versus porcine age plots in Figures 5.27 and 5.28 show a linear trend towards increasing peak force with increasing porcine age. The regression also captures a majority of the 95% confidence intervals at each age point. Again a possible exponential-type increase is visually noted for weeks 1 through 4, which are underestimated by the trendline. Peak force versus pediatric age in Figures 5.29 and 5.30 indicate a power law fit as the best representation of the data. (Again data for the subjects less than one day of age is eliminated from the regression analysis for the concerns discussed previously.) However, it is again noted that this trend does not pass through the 95% confidence intervals at each age point for the pediatric subjects. Addition of human adult rib data for peak force in 3-point bending is anticipated to result in a logarithmic regression as best fit. The increase in peak force as a function of age in both data sets is as anticipated. As the size of the sample increases, the maximum sustainable bending moment is expected to increase, resulting in a larger peak force necessary to obtain the maximum moment.

As with variables discussed previously, if the linear regression for peak force of the porcine data is set equal to the power fit for peak force of the pediatric data, the age relationship that results is:
(porcine age) = 0.0261 (pediatric age)\(^{1.3125} + 1.83\).

This equation is plotted as Figure 6.4 below.

![Graph showing peak force as a function of porcine age vs pediatric age](image)

**Figure 6.4.** Plot of porcine age as a function of pediatric age, as determined by comparing measured peak force values.

The result of Figure 6.4 predicts an increasing function of porcine age as a function of pediatric age by a power of 1.3125. From this plot, a 10-year-old child is represented by a 16-week-old pig, as compared to a 6-week-old pig by moment of inertia.
predictions, a 12.5-week-old pig by stiffness predictions, and a 30-week-old pig by flexural rigidity predictions. A newborn human infant is estimated by a 2-week-old pig.

Several data points indicate the possibility for peak force dependence on rib level. For both the porcine subjects and the pediatric subjects, slightly higher values of peak force are seen for anatomically lower rib levels. The statistical significance is not known.

6.1.7 Rupture Modulus/Bending Strength

As with Young's modulus, the plot of rupture modulus, or bending strength, as a function of porcine age (Figures 5.31 and 5.32) did not yield results as expected. It was anticipated that older porcine subjects would have an increased mineral content, increased size, and increased peak force, thereby increasing the strength of the bone. However, the general linear regression trendline indicates an overall decrease with age. Visual inspection of the plot indicates that the bending strength values may tend toward a constant value over all ages. The average bending strength value of all samples for all porcine ages was calculated to be 49.27 MPa, but this value did not pass through the 95% confidence interval of several ages. Fairly constant values are seen over most ages, except for low bending strength at 13 weeks, 14 weeks, and 16 weeks. Removal of these data points from the average calculation gives an average value of 52.76 MPa, which is more consistent with the 95% confidence intervals of the remaining data. This leads to the belief that the average value better represents the data for rupture modulus/bending strength as a function of porcine age, indicating that the bending strength of the porcine rib does not change significantly over development. Values from additional subjects at each age are needed to further investigate this conclusion.
Rupture modulus versus pediatric age, as presented in Figures 5.33 and 5.34, is represented by a power fit through the data points. A power fit gave the best regression (as determined by $R^2$ values) for the pediatric data through 6 years of age. However, it is anticipated that the bending strength would not continue to increase as a power function. As with Young's modulus, a logarithmic trend may better predict strength of values from the adult are added to the analysis. Additionally, the values calculated for the 6-year-old subject indicate that a plateau may have been reached in the plot. Properties for several other ages must be determined before this statement can be verified. As indicated in Chapter 5, the data from the subjects less than one day of age were removed from the regression due to concerns discussed previously. Because the above-discussed analysis presents the possible conclusion that bending strength does not change as a function of porcine age, no prediction of equivalent porcine age for each pediatric age can be made.

There is no obvious trend for rib level dependence of rupture modulus. There is scatter among various rib levels in the porcine data displayed in Figure 5.31. In some pediatric subjects of Figure 5.33, anatomically lower rib levels had larger values for rupture modulus. However, there is overlap among several data points, and no statistical analysis was conducted to determine significance.

6.1.8 Yield Force

The yield force versus porcine age plots in Figures 5.35 and 5.36 show a linear trend towards increase peak force with increasing porcine age. The regression also captures a majority of the 95% confidence intervals at each age point. Again a possible exponential-type increase is visually noted for weeks 1 through 4, which are
underestimated by the trendline. Yield force versus pediatric age in Figures 5.37 and 5.38 indicate a power law fit as the best representation of the data. (Again data for the subjects less than one day of age is eliminated from the regression analysis for the concerns discussed previously.) However, it is again noted that this trend does not pass through the 95% confidence intervals at each age point for the pediatric subjects. Addition of human adult rib data for yield force in 3-point bending is anticipated to result in a logarithmic regression as best fit. It was noted in Section 6.1.7 that a plateau in bending strength may have been reached by the 6-year-old subject. This is also apparent in the yield force data. However, the lack of pediatric data points beyond this age does not allow for a sound conclusion.

If the linear regression for yield force of the porcine data is set equal to the power fit for peak force of the pediatric data, the age relationship that results is:

\[(\text{porcine age}) = 0.0269 (\text{pediatric age})^{1.3479} + 1.64.\] This equation is plotted as Figure 6.5 below.
Figure 6.5. Plot of porcine age as a function of pediatric age, as determined by comparing measured yield force values.

Compared to the previous plots of Figures 6.1 through 6.4, the yield force comparisons predict a porcine age of 18 weeks for a pediatric age of 10 years. This is in between the estimations from peak force and flexural rigidity values. A newborn human infant is estimated at a porcine age of approximately 2 weeks.

Rib-level dependence of yield force does not display an obvious trend in either the porcine subjects (Figure 5.35) or the pediatric subjects (Figure 5.37). Yield force values for rib levels 6 and 7 tend to be higher than rib level 5 in porcine subjects, but this
trend is not among all samples. The pediatric plot shows larger values for rib level 6 in many subjects.

6.1.9 Yield Stress in Bending

Figures 5.39 and 5.40 for yield stress in bending versus porcine age display similar behavior as the previously discussed plots for Young’s modulus and bending strength as a function of age. Regression coefficients determined the best fit to be a linear regression, although this trend indicates a decrease in yield stress as a function of age when an increase was anticipated. The average value for all yield stress calculations across all ages is 39.81 MPa. Visual inspection of the plot indicates that a constant value appears to better fit the 95% confidence intervals than a linear regression. It is again believed that this data may indicate a constant value for yield stress as a function of age. As noted previously, additional data is required to assert this assumption.

The pediatric data set for yield stress as a function of pediatric age presented in Figures 5.41 and 5.42 again result in a power fit for the pediatric data through 6 years of age. Continuous increase as a power function is not anticipated, and a logarithmic plot may better represent the yield stress values versus age through adult. A possible plateau in the data points is again noted at the 6-year-old subject without validation until additional data points are collected. As indicated in Chapter 5, the data from the subjects less than one day of age were removed from the regression due to concerns discussed previously. Because the above-discussed analysis presents the theory that yield stress does not change as a function of porcine age, no prediction of equivalent porcine age for each pediatric age can be made.
Although the rib level dependence analysis of the plots for porcine subject yield force indicated that rib levels 6 and 7 frequently displayed larger yield force values than rib level 5, values for yield stress in rib level 5 are closer to the other levels tested in this study. No overall trend is apparent by visual inspection. The pediatric data of Figure 5.41 again shows frequently larger yield stress values in rib level 6 than in rib level 5 or 7. However, statistical significance of these visual trends is not known.

6.1.10 Bending Strength versus Young’s Modulus

Figures 5.43 and 5.44 were constructed to analyze the possibility of any outliers within the data sets. Currey presented a linear relationship between bending strength and Young’s modulus (Currey, 2002). In his study, those data points that did not fall along the linear relationship had very high mineral content. Analysis of the plots in the current study shows clustering of all porcine data points around a linear regression with an $R^2$ value of 0.3736. The pediatric plot of Figure 5.44 shows one possible data point which did not lie along this line ($R^2$ value of 0.6999), pointed out as rib level 7R from the 4-year-old subject on the plot. It is possible that this rib had a higher mineral content than other samples, but a definitive conclusion is not possible without additional information.

6.2 Porcine Subjects as an Animal Model for Pediatric Rib Bone

Discussion and analysis of Sections 6.1.1 through 6.1.10 lead to the conclusion that porcine subjects do not represent a good animal model for pediatric rib bone. Comparisons dependent on geometry and size of the sample (moment of inertia, stiffness, flexural rigidity, peak force, and yield force) show varying results for the porcine age
comparable to the pediatric ages of 0 months and 10 years. These estimates are summarized below in Table 6.1.

<table>
<thead>
<tr>
<th>Predicting Variable</th>
<th>Porcine Age</th>
<th>Pediatric Age</th>
</tr>
</thead>
<tbody>
<tr>
<td>Moment of Inertia</td>
<td>4 - 6 weeks</td>
<td>0 - 10 years</td>
</tr>
<tr>
<td>Stiffness</td>
<td>&lt;1 - 12.5 weeks</td>
<td>0 - 10 years</td>
</tr>
<tr>
<td>Flexural Rigidity</td>
<td>3 - 30 weeks</td>
<td>0 - 10 years</td>
</tr>
<tr>
<td>Peak Force</td>
<td>2 - 16 weeks</td>
<td>0 - 10 years</td>
</tr>
<tr>
<td>Yield Force</td>
<td>2 - 18 weeks</td>
<td>0 - 10 years</td>
</tr>
</tbody>
</table>

Table 6.1. Summary of porcine age and pediatric age rib development comparisons.

Analysis of normalized material properties that are not dependent on geometry and size of the sample (Young’s modulus, rupture modulus/bending strength, and yield strength) indicate the possibility that all of these values remain at a near constant value throughout the growth and development of the pig. All pediatric data for these variables showed an increasing trend as a power function for 5 months of age through 6 years of age. Additional data is required in order to understand the accuracy of this power fit prediction, as compared to a possible logarithmic regression, but the increase in values does not match the predicted constant values for the pig.

There are several possible explanations for the inability to use a porcine animal model to determine the behavior of the pediatric rib. Pig development progresses at a
much faster rate than human development. A pig is full grown around 6 months of age, whereas a human is not fully grown until skeletal maturity is reached in the late teen years. It may be that the development seen in the first few years of human life is captured completely in the first week or two of porcine life. Additionally, pigs are upright shortly after birth, whereas humans do not walk until around one year of age. As a quadruped, the samples harvested from the porcine subjects were affected by gravity and body loads during development in a different manner than the pediatric ribs.

Finally, the porcine subjects used in this study were not specifically raised for this research. They were supplied by a local farmer, and, therefore, were not subject to a strict diet or strict guidelines for raising the animals. It is possible that rib samples from porcine subjects raised in a controlled environment will display different trends for each variable as a function of age.

6.3 Comparison to Previous Work in the Literature

The comparison of the results of the current study to those previously published in the literature is necessary to fully understand the significance of the present findings. Several of the studies discussed in the Introduction are relevant to the variables calculated in this work. Each is analyzed below in comparison to this study.

A porcine model for pediatric behavior was developed for abdominal injuries by Arbogast, et al. Anthropomorphic and geometric measurements equated a 6-year-old child to an 11-week-old pig (Arbogast, 2005). Regressions presented above for porcine age as a function of pediatric age predict a 6-year-old child to be a 5-week pig, a 6-week pig, an 11-week pig, a 9-week pig, or a 10-week pig for moment of inertia, stiffness,
flexural rigidity, peak force, and yield force comparisons, respectively. However, the
discussion of section 6.2 details why porcine subjects are not a good animal model for the
pediatric rib. The porcine age corresponding to a 6-year-old child predicted by flexural
rigidity (EI) in this study matches that of Arbogast, et al.

As reported in Chapter 1, the only study identified for testing of pediatric ribs was
that by Theis (Theis, 1975), as discussed by Sturtz (Sturtz, 1980). Sturtz reports that the
study by Theis on human ribs found an average dynamic loading fracture load of 234
Newtons for all samples from children under the age of 14 years, as tested in dynamic
bending (Sturtz, 1980; Theis, 1975). In the current study, all pediatric subjects were
under 14 years of age. The average peak force of all pediatric samples was 41.85
Newtons, with a maximum peak force of 120.7 Newtons for one of the ribs from the six-
year-old subject. The regressions of peak force as a function of pediatric age (see Figures
5.29 and 5.30) indicated that there was an increase in peak force with increasing pediatric
age. It can be anticipated that addition of samples from older subjects through 14 years of
age would result in larger peak forces, thereby increasing the expected average peak force
of all samples. However, direct comparison of the data in the current study to the average
value presented in the study by Theis is not feasible. Theis’ samples were tested
dynamically, whereas the current study used quasi-static loading. As discussed in Chapter
2, the viscoelastic properties of bone show a dependence on strain rate with larger values
anticipated for higher strain rates (Cowin, 2001; Nordin and Frankel, 1989; Turner and
Burr, 1993). The only relationship that can be noted is that the average peak force for the
current study is less than that reported by Theis. This is as expected due to the lower
strain rate in the current study, as well as the younger ages included in the average.
The results of quasi-static bending tests on human adult ribs in the literature are also of interest to the current findings. Yoganandan and Pintar loaded adult ribs in 3-point bending at the quasi-static rate of 2.5 millimeters/minute, as in this current study. The study reports average Young’s modulus values of 2318 (± 364.6) MPa for rib level 7, and Young’s modulus values of 1886 (± 287.0) MPa for rib level 8. The pediatric rib Young’s modulus results of the current study are in the range of 120 MPa to a maximum of 8184 MPa, although the second largest Young’s modulus value was 5256 MPa. It was anticipated that the current study would yield Young’s modulus values lower than the adult values in the literature, which is not the case with the study by Yoganandan and Pintar. Although Yoganandan and Pintar report different values for the Young’s modulus of the adult rib based on rib level 7 or rib level 8, overlap in the confidence intervals indicated no significant difference between levels 7 and 8 for Young’s modulus. Although no statistical comparison based on rib level was conducted in the current study due to the limited number of samples at each age, the most noticeable larger values for Young’s modulus were calculated for rib level 7. The larger Young’s modulus values are also evident in rib level 7 of the study by Yoganandan and Pintar. Additionally, the Yoganandan and Pintar study reported average moment of inertia values of 0.14 cm$^4$ and 0.10 cm$^4$ for rib levels 7 and 8, respectively. All pediatric moment of inertia values were lower than these values, as anticipated due to their smaller size.

Stein and Granik report an average rupture modulus (bending strength) of 92.87 MPa for 3-point-bending of adult ribs. The peak rupture modulus in the current study was 102.8 MPa for a 4-year-old subject. Porcine data had a peak of 85.54 MPa, with an average of 49.27 MPa. As Figures 5.33 and 5.34 indicate an increase of rupture modulus
with pediatric age, it would be expected that the data in Stein and Granik would be larger than that of the current study. Also of note is a linear decrease in breaking load, rupture modulus, and cross-sectional area as a function of age for the Stein and Granik study. However, the decrease reported was from human adult age of 27 years through 83 years (Stein and Granik, 1976). Additional data points between the ages of 6 years and 27 years are needed to fully understand the changes in these properties as a function of age.

Schultz, et al, reported geometrical and force-deformation properties of ribs. Of note with respect to the current study is their report of nonlinear regions of force versus deflection curves (Schultz, 1974). Similar nonlinear regions were found for the ribs tested in the current study.

Cormier, et al, analyzed adult human rib properties in bending under dynamic loads. As mentioned in the discussion of the study by Theis, direct comparison between the values obtained under dynamic loading to the values of the current study under quasi-static loading cannot be made. However, it is anticipated that values in dynamic testing will be higher than those in quasi-static testing. The dynamic loading produced an average peak moment of 2.9 Nm for the anterior region of the adult rib (Cormier, 2005). The maximum peak moment for quasi-static loading of the porcine data set was 8.63 Nm, whereas the maximum peak moment for quasi-static loading of the pediatric data set was 2.12 Nm. Quasi-static loading of the pediatric data resulted in lower peak moments, although the porcine data set displayed slightly higher peak moments. It is unknown whether or not this variation is related to the data set being from an animal model or not.

In addition to investigations in the literature that determined material properties of adult ribs, pediatric material properties in bending have been reported for fetal cranial
bone. One study reports a range of values for Young's modulus from 1145 MPa to 4413 MPa for several specimens of fetal cranial bone at different gestational ages (Kriewall, 1981). Although properties of fetal cranial bone are expected to differ from those of the pediatric rib, the range of values is comparable to those obtained in this study for Young's modulus of the pediatric samples (see Figures 5.21 and 5.22).

Margulies and Thibault report on mechanical bending properties for human infant cranial bone as compared to porcine infant bone. Of note is the finding that Young's modulus of the human infant data set is approximately 3-4 times larger than the porcine data set. Average values for rupture modulus are in the range of 2-3 times larger for the human infant data set as compared to the porcine infant data set (Margulies, 2000). This supports the findings in this current study of larger calculated values of Young's modulus in the pediatric data set as compared to a majority of the data points of the porcine data set.

Although several studies in the literature investigate the material properties of long bones in children, the values obtained from a load-bearing bone such as the femur are not comparable to the pediatric rib. However, one study of interest is that by Currey that tested pediatric femurs in 3-point bending. Modulus of elasticity in bending, as well as bending strength, is reported over a range of ages. Plotting both of these variables as a function of age shows an overall increase in the material property values as age increases. Figure 6.6 plots the normalized Young's modulus (value divided by maximum) values for the femur data of Currey's study as a function of age, as well as the normalized Young's modulus (value divided by maximum) values for the pediatric rib data of the
current study as a function of age. Figure 6.7 is the same plot for the variable of bending strength.

Figure 6.6. Comparison of current pediatric rib study Young’s modulus age-dependence with femur Young’s modulus age-dependence (Currey, 1975).
Figure 6.7. Comparison of current pediatric rib study bending strength age-dependence with femur bending strength age-dependence (Currey, 1975).

The data from Currey is presented above as Figures 6.6 and 6.7, and shows a relationship that compares to the ages tested in the current study. Linear regressions on the normalized data from both studies indicate similar slopes, indicating that the effects of age on changes in Young’s modulus values and bending strength values are consistent. Average femur Young’s modulus increases 68% from 2 years of age through 6 years of age (Currey, 1975), whereas average rib Young’s modulus from the current study increases 97% from 2 years of age to 6 years of age. Average femur bending strength
increases 31% from 2 years of age through 6 years of age (Currey, 1975), whereas average rib bending strength increases 29% from 2 years of age through 6 years of age.

6.4 Discussion of Theoretical Assumptions

One of the criteria listed in Section 2.3.2 for three-point bending of beams was that the material remains linearly elastic up until the point of failure. However, there is plastic behavior that is encountered in bending before complete failure. This can be summarized in Figure 6.8 below (Currey, 2002). As the beam is initially loaded, stress and strain are both linear through the cross-section and symmetric on both sides of the neutral axis. Stress is given by the dotted line, whereas strain is given by the solid line. With the addition of increasing bending moment, stress and strain increase proportionally from the neutral axis to a maximum at both the compressive and tensile outer surfaces (Figure 6.8, A). As stress and strain continue to increase, the values remain proportional until the yield stress has been reached (Figure 6.8, B). The outermost surface at the highest value of stress eventually reaches the yield stress at the bone, but the rest of the cross-section is not at the yield point (Figure 6.8, C). An increasing bending moment continues to increase strain because the bone has not reached ultimate failure. Eventually the stress will be increased through the cross-section until the entirety has reached yield. As a result, when the outer tensile surface reaches the yield stress of the material, a greater bending moment can still be applied. The result is a larger strength measurement than would be measured in a tensile test (Currey, 2002; Timoshenko, 1958). In other words, there is plastic deformation in the bone before failure is reached.
Figure 6.8. Stress and strain in a beam cross-section loaded in bending (Currey, 2002).
The second assumption in 3-point-bending was the beam is a straight beam, and the ratio of the radius of curvature, $R$, to the depth of the beam, $d$, is greater than 5. (Boresi, 1993). The radius of curvature was calculated from the following equation for the radius of a circle:

$$R = \frac{4b^2 + c^2}{8b}$$

$$R = \frac{4(\Delta y)^2 + l^2}{8(\Delta y)}$$

where ‘c’ is the length of a known chord and ‘b’ is the perpendicular length from the chord to the edge of the circle. The chord length was taken to be the length of the span in the test setup. The perpendicular height ‘b’ was calculated as the measured height from the MTS table surface to one of the penned-on cross-hair markers minus half of the thickness of the rib minus the height from the MTS table surface to the end support of the test fixture. This would give the radius of curvature to the inner surface of the rib. (See Figure 6.9 below for clarification.) Because the inner surface of the rib would have a smaller radius of curvature than the radius of curvature of the neutral axis, this value is a worst-case scenario for calculating the lower bound limit for the ratio $R/d$. The depth of the beam ‘d’ was found as two times the maximum radius of the cross-section in the vertical direction, as determined from the cross-sectional analysis of the CT image of each sample. In order for this analysis to be valid, the assumption that the penned-on cross-hair marker is at the vertical location of the centroid of the cross-section must be
made. Also it is assumed that the radius of curvature of the rib is constant throughout its length, and the radius can be approximated by a circle.

Figure 6.9. Variables for calculation of radius of curvature.

From these calculations, all pediatric samples show an R/d ratio greater than 5, thereby indicating that the straight beam analysis results in negligible error as compared to a curved beam analysis (Boresi, 1993). (It is important to note that R/d could not be determined for 3 of the smallest pediatric samples, as height measurements during testing could not be taken.) The average value for the R/d ratio for the pediatric samples was 157.
14.3 ± 4.6 (95% confidence interval). For the porcine rib, 6 of the 150 porcine samples have an R/d ratio of less than 5. From the R/d > 5 assumption, these six samples may have a significant error in calculating material properties as a straight beam, as opposed to a curved beam. Due to the error in the height measurements during testing, the calculation for radius of curvature as described above resulted in negative values for eight of the porcine samples. The absolute values of the R/d ratio for these samples were all greater than 5. Because so few samples are affected, analysis as a curved beam is not expected to alter the results of the study. The average value for the R/d ratio for the porcine samples was 17.5 ± 7.3 (95% confidence interval).

The third assumption was that the beam is sufficiently long and slender. Approximately one-third of the porcine test samples did not meet the criterion for the ratio of the span length to the specimen thickness of L/d > 5 (Boresi, 1993). If the ratio is increased to the criteria that L/d > 16 ((ASTM D 790-03, 2003; Cowin, 2001; Turner and Burr, 1993), no porcine samples meet this criteria. The average value for the L/d ratio for the porcine samples was 6.3 ± 0.27 (95% confidence interval). All pediatric samples meet the L/d > 5 criterion, but only one sample has an L/d ratio of greater than 16. The average value for the L/d ratio for the pediatric samples was 11.3 ± 0.98 (95% confidence interval). As the curvature of the rib varies along its length, sample length was dictated by the section of rib that appeared straightest. The result is the possibility of non-negligible shear stresses throughout the beam that were not accounted for in the calculation of the material properties of the pediatric rib. If the negligible shear is only valid for an L/d ratio greater than 16, this may explain the outcome of the ribs from the 5-month-old subject that failed in shear longitudinally through the rib (Figure 5.2).
The assumption that moduli remain constant throughout the test specimen was not investigated in this study. The heterogeneous characteristics of bone material may have not met this criterion, as material content varies along the length of the bone. Bone elastic modulus values are not the same in tension and compression, resulting in asymmetric distribution throughout the cross-section loaded in bending.

Another assumption listed in Section 2.3.2 stated that the neutral and centroidal axes must coincide. This is not always the case for asymmetrical cross-sections (Ugural and Fenster, 1995).

The final assumption for 3-point-bending of beams was stated to be that the material is homogeneous and isotropic. The properties of bone have been well documented as nonhomogeneous and anisotropic due to its composite structure (Currey, 2002). Some cross-sections along the bone length are weaker than others. However, in three-point bending the maximum moment is applied at the midspan of the beam. If the weakest cross-section is not directly under this load, it is possible that the strength of the bone is overestimated.

The testing of bone in three-point bending requires several assumptions if analysis of a simple beam is to be used. Bending tests have been widely used in the literature as a means of estimating the material properties of bone (ANSI/ASE S459, 2003; Biewener, 1992; Cormier, 2005; Cowin, 2001; Currey, 2002; Currey, 1975; Kemper, 2005; Kriewall, 1981; Margulies, 2000; Nordin and Frankel, 1989; Stein and Granik, 1976; Turner and Burr, 1993; Yamada, 1970; Yoganandan and Pintar, 1998). Complex analyses may give more accurate results, but use of standard test procedures allow for comparisons to previous studies, as well as advancing the knowledge of bone mechanics.
6.5 Limitations of the Study

The above discussion (Section 6.4) on the shortcomings of the theoretical assumptions that must be made for three-point bending indicates the first set of issues that arise as limitations of the current study. Another limitation is also related to bone structure and the biomechanical testing of bone in general. As discussed in Chapter 2, the viscoelastic properties of bone result in vast differences in the material properties of dry bone and wet bone (Cowin, 2001; Nordin and Frankel, 1989; Turner and Burr, 1993). Care was taken to ensure that the bone remained moist by wrapping the rib samples in saline-soaked gauze when not in preparation or being testing. However, the foam holder technique for fixing the orientation of the bone cross-section in the CT image required that the bones be exposed to the air for an extended period of time. Bones were sprayed with saline prior to transport to the CT scanner, and again wrapped in saline-soaked gauze and returned to individual bags as soon as possible after imaging. However, excessive drying of the bone was noted upon return from the CT scan. The result may be increased values of Young’s modulus and bone strength if the bone became too dry (Cowin, 2001; Nordin and Frankel, 1989; Turner and Burr, 1993).

Yield point was determined for each sample with a mathematical analysis developed specifically for this study. The force versus deflection response of rib bone does not show a visually clear transition from linearly elastic to nonlinearly elastic to plastic regions. The correlation between the results from the mathematical analysis to yield point determined experimentally is not known.

Another limitation of the current study is the small sample size. Only one subject was available at a majority of the age points in the study. Variation among biological
subjects is a constant concern in biological testing. By increasing the sample size, subject-dependent effects can become negligible. For example, although a wide range of porcine subjects were included in the testing, samples were analyzed as a function of age. Any factors unique to individual animals could not be accounted for. The trends shown in the plots of Chapter 5 frequently indicate that the material properties of the porcine rib may not vary with age. However, comparison of averages at each age show increases over certain age ranges, as well as decreases in others. Whether or not these trends are indicative of actual age-related behavior or subject-dependent properties cannot be determined without additional data. The small range of pediatric samples available also limits the statistical conclusions that can be drawn from the study results.

A further limitation of this investigation is the source of porcine subjects being a local farmer. The animals were raised for meat production, and the feeding and living environments were not controlled. Many inconsistent variations over the age ranges were noted. Whether or not the same variations would be evident in animal subjects raised in a controlled environment specifically for a study of this nature is not known.

The testing of isolated excised ribs in bending introduces the limitation that the loading may not replicate real-world behavior. The ribs are fixed at both the vertebrae and the sternum, although movement is allowed. Interaction with intercostal muscles and underlying organs of the thorax may alter the response of the pediatric rib cage to loading. Also, the current test setup loaded each rib at a quasi-static strain rate. However, loading to the thorax, especially of the type of loads that result in injury or trauma, is usually of a dynamic nature. As discussed previously, the behavior of bone to static loading or in dynamic loading give different values for the material properties.
6.6 Conclusions

Although the small sample size of the current study limits the statistical significance of the findings of the current study, several interesting results have been noted throughout this document. These conclusions are listed and summarized below.

6.6.1 Porcine Testing Conclusions

- When loaded in three-point bending, the porcine rib experiences negligible rotation of the loaded cross-section. Orientation of the principal axis of the minimum moment of inertia does not always correspond with the bending axis.

- The geometry- and size-dependent variables of the porcine rib (moment of inertia, stiffness, flexural rigidity, peak force, and yield force) all increased linearly as a function of age.

- Normalized variables of the porcine rib (Young’s modulus, rupture modulus, and yield stress) showed decreasing linear regressions as a function of age. It was noted for each of these variables that a possible constant value of each as a function of age is a possibility. Increased sample size is needed for justification of this statement.

6.6.2 Pediatric Testing Conclusions

- When loaded in three-point bending, the pediatric rib experience negligible rotation of the loaded cross-section. Orientation of the principal axis of the minimum moment of inertia does not always correspond with the bending axis.
• The moment of inertia of the pediatric rib increased linearly as a function of age from infant through 6 years.

• Rib stiffness, Young’s modulus, flexural rigidity, peak force, and yield force of the pediatric rib all increased as a power function with exponent greater than 1 as a function of age from infant through 6 years.

• Strength measurements as the rupture modulus (bending strength) and yield stress of the pediatric rib increased as a power function with an exponent of less than 1 as a function of age from infant through 6 years.

6.6.3 Significance of Results

• Correlation with data presented in the literature is noted.
  
  o Although statistical analysis was not conducted, similar rib-level dependencies were seen as compared to previous adult rib quasi-static bending tests (Yoganandan and Pintar, 1998).

  o Larger values seen for calculation of Young’s modulus for human pediatric bone as compared to porcine bone. This result is similar to that of Margulies and Thibault (Margulies and Thibault, 2000) for fetal cranial bone.

  o Normalized data comparisons for both Young’s modulus and bending strength show similar age dependencies for the pediatric femur bending properties of Currey (Currey, 1975) and for the pediatric rib bending properties in the current study.
The results of this study indicate that porcine subjects do not present an acceptable animal model for the behavior of the pediatric rib. Development rates between humans and the pig are vastly different. Additionally, the load-bearing characteristics of the two vary.
CHAPTER 7

FUTURE RECOMMENDATIONS

7.1 Changes to Current Test Setup

In order to address some of the concerns that arose during this study, changes to the current test setup are proposed. First, a new test fixture should be designed to accommodate the smaller rib samples. This would eliminate the concern that small samples (similar in size to the two subjects of less than one day of age) are loaded in compression instead of bending. As much development occurs in the first few months after birth, it is important that the test fixture allows for accurate determination of the material properties of the rib from these subjects.

The drying of the rib samples was also a limitation for confidence in the results of this data set. As discussed in Chapter 6, dry bone may display larger values for Young’s modulus and bone strength. Testing samples from the same subject at various stages of drying would allow for determining the extent of the increase in these values from drying. If the increase is significant, future testing needs to address the issue of ensuring the bone remains moist. One suggestion is transporting fewer samples to the CT scanner at a time. This would allow for shorter preparation time, shorter time in transport and at
the scanner, and shorter time for returning samples to storage. As a result, the bones would have less exposure to air, thereby limiting the amount of drying prior to testing.

Assumptions for three-point bending of bone should also be addressed. It was noted in Chapter 6 that most of the samples did not meet the $L/d > 16$ assumption. Priority was placed on obtaining a straighter sample of bone, rather than meeting the length specification. Additional testing may indicate what extent these variations have on the material properties obtained. Machined coupon samples from the rib bone may also be a possibility to ensure that the test setup mimics a simply supported beam. Strain gauges glued to the samples have also been used to evaluate the bending properties. However, the small size of the pediatric rib may prevent machining of smaller samples, or the placement of strain gauges without altering the recorded response to loading.

Also, samples should be tested to ultimate failure as opposed to visual confirmation of yielding. Data through ultimate failure would allow for calculation of additional variables of interest, such as toughness, energy absorbed to failure, and analysis of plastic behavior. The anticipated greater plastic deformation of pediatric ribs as compared to adult ribs present an argument for better characterization of the material with these added properties.

As discussed in Section 6.5, the correlation between the mathematically determined yield point in this study with the actual yield point is not known. It is suggested that this mathematical method be analyzed against experimental data. This can be achieved by loading samples partially and then unloading. The load at which the bone does not return to its original position is indicative of plastic behavior, and indicates that
the yield point has been reached. This value can be compared against the mathematically calculated value.

A final suggestion for continuing the current study is to increase the sample size. As discussed previously, the need for a large sample size would decrease the effects of subject-dependent variances that are of concern with biological testing. Many subjects are needed at each age in order to fully understand the changes in the material properties of the pediatric rib as development continues. An increased sample size will allow for statistical conclusions of the changes in properties with age.

It is not suggested that porcine rib testing be continued for this study. The results presented here indicate that porcine subjects are not acceptable models of pediatric rib behavior. The investment and efforts for continuing porcine testing will not aid in understanding the change in material properties as a function of age in humans.

7.2 Design of a Handheld Testing Device

A limitation noted in Section 6.5 was that the testing of the excised rib may not result in the same material properties as loading to the rib \textit{in situ}. Relationships with internal muscles and organs may alter the response. As a result, the design of a handheld testing device that would allow for the testing of rib samples \textit{in situ} has been proposed. The objective of the design is an easy-to-use device that can be transported to autopsy in order to record the force versus deflection response of the rib \textit{in situ}. Two initial designs were developed by a Mechanical Engineering Senior Design class at The Ohio State University.
The first design proposed the use of a stepper motor for applying the deflection rate. Force data would be obtained by means of a load cell. Displacement can be measured by a linear variable displacement transducer (LVDT) or by the advancement of a power screw. See Figure 7.1 below for a schematic of this design.

Figure 7.1. Proposed design #1 for a handheld *in situ* rib testing device.

The second design proposed the use of a linear actuator for the deflection rate. A mini load cell was used to collect force data. The connection of a linear potentiometer to the linear actuator was suggested as a means for determining deflection. A 12-volt battery
would supply power to drive the device, and a mini data acquisition system could be installed in the handle. The design suggested the use of a pre-existing handheld drill housing. See Figure 7.2 for a schematic of this design.

![Proposed design #2 for a handheld in situ rib testing device.](image)

Figure 7.2. Proposed design #2 for a handheld in situ rib testing device.

The feasibility of the two designs remains to be determined. However, the final design must meet several design goals. The device must be portable, so that testing can be conducted during an autopsy. Additionally, there needs to be a means for data storage in order to allow for data analysis after completion of testing and removal from the autopsy suite. The two end loading supports of the device must be capable of fitting behind the rib to mimic the loading of the current study. The application of the load at midspan from the outer surface of the rib must not cause significant deflection or bending.
of the supports on the inner surface of the rib. Additionally, these end supports must be adjustable to allow for several sizes of samples. All assumptions discussed in Chapter 2 for the testing of bone in three-point bending must be considered in the design of the handheld device as well.

7.3 Dynamic Testing

The quasi-static loading used to determine the material properties of the pediatric rib in this study does not account for the response to dynamic loading as seen in real-world impacts. It is suggested that future testing account for this type of test setup. This may be achieved by design of a test fixture that connects the loading nose at midspan to a column of a specified height. The release of this top portion of the test fixture would result in an impact at the midspan of the beam. The test setup used in the current study may also be altered by increasing the deflection rate applied by the MTS. Large deflection rates be similar to the impact loading that causes injuries in real-world situations. The values obtained for material properties from dynamic testing would be expected to be higher than those obtained from quasi-static testing.

7.4 Long-Term Direction

Future direction of this project would lead to better characterization of the pediatric thorax. As discussed in the Introduction of Chapter 1, understanding the behavior of the pediatric thorax will assist in preventing child injuries, particularly with more biofidelic crash test dummies. The large relative size of the pediatric thorax results
in loading to the thorax affecting the whole-body response of the child. Determining the material properties of the pediatric rib is only a beginning step in this characterization.

Future work is needed to understand how the material properties of the excised rib relate to the overall mechanical response of the pediatric thorax. Additional testing may include investigating the result of impact to the full thorax of subjects. However, the availability of pediatric subjects is limited, and several ethical issues arise with pediatric testing.

One use for the material properties of the excised rib is development of finite element models of the pediatric thorax. By characterizing the geometry and material properties of all components of the pediatric thorax, a model can simulate how the entire thorax will respond to loading. Finite element modeling is of great interest because it allows for analysis of mechanical responses without the use of human subjects or biological tissue.

In summary, the arguments presented in this document indicate a strong need to better characterize the behavior of the pediatric thorax. The current study is an initial investigation into an area that has not been explored in the literature. Future work is still needed to fully understand the mechanical response of a developing child.
LIST OF REFERENCES


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APPENDIX A

IRB APPROVAL
October 5, 2005

Gary Smith, MD
Center for Injury Research & Policy

Re: IRB05-00035
DETERMINATION OF MATERIAL PROPERTIES OF THE PEDIATRIC RIB

Dear Dr. Smith:

The contingencies for the above study have been reviewed and approved by the Institutional Review Board on 10/04/05 - CONTINGENCIES SATISFIED.

Date of Initial Approval: 06/21/05
Date of Expiration: 06/21/06

This approval is only for one year. A Continuing Review Report must be approved before this study can proceed beyond the date of expiration. Please be aware that all modifications to the research protocol, consent form, or any other aspect of this study must receive prospective IRB approval. Regulations require that provisions are made for ascent of subjects age nine and older.

The Federalwide Assurance number assigned to the IRB at Children's Hospital, Inc. is FWA0000286.

If we can provide additional assistance, please do not hesitate to call this office at ext. 22708.

Sincerely,

Alexander Sokolowski, MD, Chair
Institutional Review Board

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APPENDIX B

DATA FROM PRELIMINARY TESTING
Figure B.1. Force vs. displacement for 4-week-old pig (preliminary test #1).
Figure B.2. Force vs. displacement for 14-week-old pig (preliminary test #1).
Figure B.3. Force vs. displacement for 6-week-old pig (preliminary test #2).
Figure B.4. Force vs. displacement for 8-week-old pig (preliminary test #2).
Figure B.5. Force vs. displacement for rib level 8 for several pigs (preliminary test#3).
Figure B.6. Force vs. displacement for rib level 8 for several pigs (preliminary test #4).
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DATA FROM CURRENT STUDY
Figure C.1. Force vs. deflection curves from 1-week-old pigs.
Figure C.2. Force vs. deflection curves from 2-week-old pigs.
Figure C.3. Force vs. deflection curves from 3-week-old pigs.
Figure C.4. Force vs. deflection curves from 4-week-old pigs.
Figure C.5. Force vs. deflection curves from 5-week-old pigs.
Figure C.6. Force vs. deflection curves from 6-week-old pigs.
Figure C.7. Force vs. deflection curves from 7-week-old pigs.
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Figure C.9. Force vs. deflection curves from 9-week-old pigs.
Figure C.10. Force vs. deflection curves from 10-week-old pigs.
Figure C.11. Force vs. deflection curves from 11-week-old pigs.
Figure C.12. Force vs. deflection curves from 12-week-old pigs.
Figure C.13. Force vs. deflection curves from 13-week-old pigs.
Figure C.14. Force vs. deflection curves from 14-week-old pigs.
Figure C.15. Force vs. deflection curves from 15-week-old pigs.
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Figure C.20. Force vs. deflection curves from 20-week-old pigs.
Figure C.21. Force vs. deflection curves from subject PedRib#1.
Figure C.22. Force vs. deflection curves from subject PedRib#2.
Figure C.23. Force vs. deflection curves from subject PedRib#3.
Figure C.24. Force vs. deflection curves from subject PedRib#4.
Figure C.25. Force vs. deflection curves from subject PedRib#5.
Figure C.26. Force vs. deflection curves from subject PedRib#6.
Figure C.27. Force vs. deflection curves from subject PedRib#7.
Figure C.28. Force vs. deflection curves from subject PedRib#8.
Figure C.29. Force vs. deflection curves from subject PedRib#9.