Development of a Full-Body CAD Dataset for a Computational Model of the 5th Percentile Female

By

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**ABSTRACT**

Motor vehicle injuries and fatalities remain a leading public health concern worldwide. In 2013, the World Health Organization reported more than 1.2 million deaths as a result of motor vehicle crashes (MVCs). MVCs are also estimated to have a yearly economic cost of nearly $145 billion. To mitigate the societal impact of these crashes, researchers are using a variety of tools, including finite element (FE) models, to develop and evaluate vehicle safety devices. Such models are often developed to represent an average (50th percentile by height and weight) male occupant. However, a 2011 study from the University of Michigan Transportation Research Institute marked the first year that there were more licensed female drivers than male drivers in the United States. In order to predict the response of a greater portion of the vehicle occupants beyond the 50th percentile male, there is interest in developing such models for other cohorts among the driving public. This work focuses on the female driver in the 5th percentile of height and weight. In the past, 5th percentile female models have primarily been scaled from existing average male models, since scan data for such a specific target anthropometry is limited. However, as part of the Global Human Body Models Consortium (GHBMC) project, comprehensive image and anthropometrical data of a 5th percentile female was acquired. The long range objective of this project is to develop a detailed 5th percentile female full body finite element model in two postures: standing (pedestrian) and seated (vehicle occupant). This thesis describes the first phase of this effort, in which this multi-modality dataset is used to develop a whole-body set of CAD data for the 5th percentile female.
Chapter I: Introduction and Background

ROAD TRAFFIC INJURIES AND HUMAN BODY MODELING

According to the 2013 World Health Organization’s (WHO) Report on Road Safety, road traffic injuries are the 8th leading cause of death globally [1]. This translates to more than 1.2 million deaths worldwide each year. With the increasing number of global motorists, the WHO also predicts road traffic injuries to become the 5th leading cause of worldwide death by 2030. Motor vehicle crashes (MVCs) also have significant economic costs, with a reported $518 billion being spent each year due to injuries sustained during traffic accidents. In the United States alone, MVCs account for over 30 thousand deaths, 1.5 million injuries, and $230 billion in economic costs each year [2].

To mitigate the toll of these injuries, a variety of techniques have been used to both develop and evaluate safety features for motor vehicles. One of the emerging tools for this purpose is the finite element method. Using this method, full body computational models can be used to conduct motor vehicle crash simulations with the purpose of observing how the model interacts with the vehicle and how it performs with specific safety devices, such as seatbelts and airbags. By developing high biofidelity models, researchers gain the ability to investigate localized tissue responses to impacts, thereby improving the understanding of how the body behaves during dynamic impact events and potentially providing new insights into injury mechanisms.

Traditionally, human body models are developed to represent an average male (50th percentile in terms of height and weight). While these models can provide a valuable assessment of the mid-sized adult male, real world motor vehicle crashes
involve occupants of various size, age, and gender. This provides motivation to expand
the suite of high biofidelity human body models available to researchers.

**MOTIVATION FOR SMALL FEMALE MODELS**

Historically, the National Highway Traffic Safety Administration’s (NHTSA)
Federal Motor Vehicle Safety Standard (FMVSS) for the protection of occupants in the
United States required the used of 50th percentile male dummies for its frontal crash tests.
However, reports of air bag fatalities and other serious injuries in otherwise low severity
impacts raised concerns about the effectiveness of safety devices for protecting occupants
of other sizes and seat positions. One of the common factors in each of these air bag
related deaths was proximity to the steering wheel. Also, stature, age, and gender seemed
to play a role. NHTSA has cited that 78% of drivers who sustained fatal injuries due to
air bag deployment in minor to moderate vehicle crashes were female. Of the females
who died in these crashes, 82% were less than 163 cm tall [3]. As a result, the use of a
small female ATD is currently required for various regulatory tests such as a frontal
impact, side-pole impact, and a side-barrier impact.

According to a recent study observing gender trends in motorists, there are now
more licensed female drivers than male drivers in the United States [4]. Thus, it is
important to represent the changing demographic. Small females are considered the most
vulnerable driving population; therefore, models of these occupants can be used to
provide a means of conservatively estimating the performance of safety devices.

Previous research has shown that females are considered to be at a larger risk of
sustaining injury during automotive accidents when compared to males [5,6]. Based on a
comparison of drivers in the United States and the United Kingdom using the National
Automotive Sample System (NASS) and the Cooperative Crash Injury Study (CCIS), Mackay et al. found that the median tolerable delta-V was considerably lower for females than for males at any given severity level on the Abbreviated Injury Scale (AIS) [7]. A statistical analysis using data from the Master Accident Record System (MARS) has also found that females have a higher risk of sustaining fatal or disabling injuries as a result of vehicle crash [8]. To reduce confounding effects of crash severity, Evans et al. reported that females are also at a greater risk of injury compared to males in similar severity physical impacts [5]. More specifically, small females are also more likely to sustain chest injuries with a maximum AIS of 2 or greater [9]. Summers et al. also reported that the 5\textsuperscript{th} percentile female experiences 23\% higher chest accelerations and neck injury values (Nij) 2-3 times higher when compared to the 50\textsuperscript{th} percentile male. One of the main reasons for this increased risk is the preferred seated posture of small females closer to the wheel as a result of their stature [10-12]. Increased injuries may also be attributable to lower structural strength within females due to lower bone mineral content [13].

**EXISTING SMALL FEMALE FINITE ELEMENT MODELS**

Computational human body modeling for blunt injury prediction and prevention is a growing technique used in the field of biomechanics. The past 20 years has seen a large increase in the number of these models developed. This growth has been driven by the need to address major public health concerns, including vehicular crash, in novel and cost effective ways.
Until recently, finite element models of the small female have typically been developed by applying scaling techniques to existing 50th percentile male models, since scan data of such a specific target anthropometry is limited. The models are scaled by developing anthropometrical relationships using external anthropometry databases. While these scaled models are able to capture external dimensions of the small female, little data exists to ensure accurate representation of internal structures. Also, care must be taken to account for gender differences between males and females, such as in the thorax and pelvis.

One of the earliest finite element models of the 5th percentile female (F05), the HUMOS2 F05, was produced using European databases of anthropometry to define the external geometry of the body corresponding to specific anthropometric percentiles [14,15]. For this model, relationships were established between internal and external dimensions and then used to develop a statistical method for scaling from an average sized male. Another early model of the small female was developed by Happee et al [16]. The anthropometry of this model was obtained from data in the RAMSIS anthropometry database and was developed by scaling the geometry and joint structures of a human body model of a mid-size male. However, for modeling purposes, the majority of structures within the model were defined as rigid bodies, with only the thorax described as deformable.

In each of the previous two models, gender differences were not taken into account, either geometrically or in terms of material properties. To account for these differences, more recent models of the F05 scaled from an average male model have developed specific regions of the final female model ad hoc. In 2003, Iwamoto et al.
described the development of the Toyota Total Human Model for Safety (THUMS) small female model [17]. This model was developed by scaling the THUMS average sized male using anthropometric data of the small female occupant from the University of Michigan [18]. To account for gender differences, thoracic and pelvic regions of the model were developed separately to accurately capture the female geometry of these structures. Similarly, Kimpara et al. reported on the development of an early version of the American F05 using data on female geometry from the View Point Datalabs database [19]. However, this model did not include internal organs or female specific biomechanical properties and was not fully validated. Kimpara et al. later published work on integrating the THUMS F05 with internal organ models from the Wayne State University Human Thorax Model (WSUHTM) for improved thoracic response [20].

More recently, the THUMS AF05 has been developed to explicitly represent specific internal organs. The geometry for these structures was obtained from high resolution supine computed tomography (CT) scans of a subject representative of the 5th percentile female.

One of the main limitations for each of the models described above is a lack of posture specific organ shape and placement. It has been shown that there are significant differences in these two organ characteristics when a subject moves from a supine to a seated or standing posture [21,22]. These changes in shape and position can have significant effects on the inertial response of the abdomen to blunt loading. Therefore, it is critical to capture these characteristics in order to most accurately predict the abdominal response of the model.
As part of a larger effort within the GHBMC for developing a family of human body models, specific anthropometrical and posture specific medical image data has been acquired for the explicit purpose of developing 5\textsuperscript{th} percentile female finite element models. These models are currently being developed and will represent the small female in both seated and standing postures to model vehicle occupants and pedestrians respectively.

**GLOBAL HUMAN BODY MODELS CONSORTIUM (GHBMC)**

The Global Human Body Models Consortium (GHBMC) is a collection of automotive manufactures, universities, and government agencies whose goal is to create and maintain a set of the world’s most biofidelic human body computational models. The intended use of these models will be for the investigation of the body’s response to blunt trauma using the finite element method. To date, the GHBMC’s 50\textsuperscript{th} percentile male (M50) has been developed and validated in a number of blunt impact scenarios. This highly detailed model was derived from extensive external anthropometry and medical image data. The level of CAD detail that is explicitly represented in the model is displayed in Figure 1. The most recent version of the GHBMC M50 finite element model can be seen in Figure 2. Following a similar development strategy, the GHBMC 5\textsuperscript{th} percentile female will be modeled with the same level of detail as exhibited by the M50. The goal of these models is to provide researchers and engineers with a standardized tool for injury biomechanics research. Also, the models will serve as a valuable tool for the development and evaluation of vehicle safety devices.
Figure 1- CAD Representation of the GHBMC 50th Percentile Male Occupant

Figure 2- GHBMC 50th Percentile Male Finite Element Model
CHAPTER SUMMARIES

Chapter II: A Multi-Modality Image Set for the Development of a 5th Percentile Female Finite Element Model

The development of a CAD dataset of the GHBMC 5th percentile female is described. This includes subject recruitment, acquisition of external anthropometry and medical images, anatomical geometry development, and model assembly.

Chapter III: Characterization of Thoracoabdominal Organ Volumes for the 5th Percentile Female

In order to accurately characterize the abdominal response of the small female, it is important to validate the organ volumes used in the model. This chapter reports the abdominal organ volumes for a sample of subjects anthropometrically representative of the 5th percentile female.

Chapter IV: A Technique for Developing CAD Geometry of Long Bones Using Clinical CT Data

Accurate representation of cortical bone thickness is important to correctly characterize biomechanical loading and fracture. However, when the cortical thickness falls below a particular value, limitations inherent to the CT scanner can no longer accurately characterize the thickness. This chapter discusses a previously developed technique for generating accurate CAD of long bones using CT and specifically how to deal with regions below the scanner cut-off. As an expansion of the initial study, this technique was applied to develop cortical thickness CAD of the F05 and M95.
REFERENCES


[16] Happee R., Ridella S., et al., "Mathematical human body models representing a mid size male and a small female for frontal, lateral and rearward impact loading," in


Chapter II: A Multi-Modality Image Set for the Development of a 5th Percentile Female Finite Element Model

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ABSTRACT

To mitigate the societal impact of vehicle crash, researchers are using a variety of tools, including finite element models. As part of the Global Human Body Models Consortium project, comprehensive medical image and anthropometrical data of the 5\textsuperscript{th} percentile female (F05) was acquired. Height, weight, and 15 external anthropomorphic measurements were used to determine subject eligibility. A multi-modality image dataset consisting of CT, MRI, and upright MRI medical images was developed to characterize the subject in the supine, seated, and standing postures. Surface topography and 52 bony landmarks were also acquired for model assembly. The selected subject closely represented the F05 in terms of height and weight, deviating less than 2\% in those measures. For all 15 anthropomorphic measurements, the average subject deviation across all measures was 4.1\%. The multi-modality image set was used to develop and assemble skeletal and organ components of the model. Abdominal organ volumes and cortical bone thickness were compared to literature sources where data was available. The dataset used for the development of this model was acquired with the explicit purpose of developing a full-body finite element model of the F05 for the enhancement of injury prediction.

*Keywords* 5\textsuperscript{th} percentile female, Anthropometry, Injury, Modeling, Segmentation
2.1 Introduction

Computational human body modeling for blunt injury prediction and prevention is a growing field within biomechanics. The last 20 years has seen a large increase in the number of human body models being developed [1]. This growth has been driven by the need to address major public health problems, including vehicular crash. Motor vehicle injuries and fatalities remain a leading public health concern worldwide. In 2013, the World Health Organization reported more than 1.2 million deaths as a result of motor vehicle crashes [2]. Computational tools, such as the finite element method, offer a cost-effective way to evaluate and design safety systems within a dynamic impact environment. They also have the ability to provide a greater understanding of injury mechanisms and can be used as a basis to calculate injury criteria. Such models, whether they are used for kinematic studies or to model local level trauma, require accurate representation of the body habitus central to the research. To accomplish this, model developers have relied on a number of data sources to accurately represent the human anatomy. For example, external anthropometry and medical imaging databases have been used in the past to assemble subject specific geometries into a model coordinate system [3,4].

Traditionally, human body models are developed to represent an average male (50\textsuperscript{th} percentile in terms of height and weight). While these models can provide a valuable assessment of the mid-sized adult male, real world motor vehicle crashes involve occupants of various size, age, and gender. This provides motivation to expand the capabilities of human body models. In a recent study looking at road traffic gender trends, 2010 marked the first year that there were more licensed female drivers than male
drivers in the United States [5]. Thus, it is important to develop models that represent the changing demographics of drivers. This study focuses on the female driver in the 5th percentile of height and weight.

Previous research has shown that females are considered to be at a larger risk of sustaining injury during automotive accidents when compared to males [6,7]. Based on a comparison of drivers in the United States and the United Kingdom using the National Automotive Sample System (NASS) and the Cooperative Crash Injury Study (CCIS), Mackay et al. found that the median tolerable delta-V was considerably lower for females than for males at any given severity level on the Abbreviated Injury Scale (AIS) [8]. A statistical analysis using data from the Master Accident Record System (MARS) has also found that females have a higher percentage of sustaining fatal or disabling injuries as a result of vehicle crash [9]. To reduce confounding effects of crash severity, Evans et al. reported that females are also at a greater risk of injury compared to males in similar physical impacts [6]. More specifically, small females are also more likely to sustain chest injuries with a maximum AIS of 2 or greater [10]. One of the main reasons for this increased risk is the preferred seated posture of small females closer to the wheel as a result of their stature [11-13]. Increased injuries may also be attributable to lower structural strength within females due to lower bone mineral content [14].

Until recently, finite element models of the F05 have typically been developed by applying scaling techniques to existing 50th percentile male models, since scan data of such a specific target anthropometry is limited. To scale these models, anthropomorphic relationships are established using external anthropometry databases. For example, the HUMOS2 F05 was produced using European databases of anthropometry to define the
external geometry of the body that corresponds to specific percentiles [15,16]. Relationships between internal and external dimensions were then used to develop a statistical method for scaling. Another early model of the small female was developed by Happee et al [17]. This model was developed using anthropometry from the RAMSIS anthropometry database. In 2003, Iwamoto et al. described the development of the Toyota Total Human Model for Safety (THUMS) small female model [18]. This model was developed by scaling the THUMS average sized male using anthropometric data of the small female occupant from the University of Michigan [19]. However, in order to account for gender differences, thoracic and pelvic regions of the model were developed \textit{ad hoc} to represent the small female. Similarly, Kimpara et al. reported on the development of an early version of the American F05 using data on female geometry from the View Point Datalabs database [20]. However, this model did not include internal organs or female specific biomechanical properties and was not fully validated. Kimpara et al. later published work on integrating the THUMS F05 with internal organ models from the Wayne State University Human Thorax Model (WSUHTM) for improved thoracic response [21]. More recently, the THUMS AF05 has been improved to include more accurate models of specific internal organs. The geometry for these structures was obtained from high resolution supine computed tomography (CT) scans of a subject representative of the 5\textsuperscript{th} percentile female.

The objectives of this study are two-fold. The first is to present a method for obtaining comprehensive image and anthropometrical data of the F05. This includes medical images from both CT and magnetic resonance imaging (MRI) scans to obtain subject specific images in the supine, seated, and standing postures. Recent studies have found
significant differences in abdominal organ positioning and shape between supine and seated postures [22,23]. Therefore, in order to most accurately characterize the internal organs of a model, medical images need to be obtained in a variety of postures. The second objective is to present the techniques for 3D geometry development and model assembly. This model will be the foundation for the development of the Global Human Body Models Consortium’s (GHBMC) F05 finite element model. The consortium’s mission is to create and maintain the world’s most biofidelic computational human body models. The data presented on the development of the F05 is intended to provide an anatomic reference to engineers and researchers to aid in the advancement of automotive safety.

2.2 METHODS

The medical imaging protocol was approved by the Wake Forest University School of Medicine’s Institutional Review Board (IRB, #5705). As an initial solicitation for a single individual to represent the 5th percentile female, target height and weight of 150.9 cm and 49 kg were used. Once candidates were identified, 15 anthropomorphic measurements were acquired and compared to existing anthropometry values presented by Gordon et al [24]. For inclusion in the study, the subject was to be within 5% deviation across all measurements. Applicants also had to be in generally good health and have all organs present. Additional exclusion criteria related to the imaging component of the study, such as claustrophobia and any implanted metals, were also included to ensure subject safety.

2.2.1 Medical Imaging Protocol
The selected subject was carefully screened to ensure safety prior to scanning, and all images were reviewed by a faculty radiologist. In order to fully characterize the subject for model development, a multi-modality image dataset was collected [3,25]. This dataset was comprised of CT, MRI, and upright MRI (uMRI) to obtain images in the supine, seated, and standing postures. CT scans allowed for accurate reconstruction of skeletal structures. Seated and standing uMRI scans were used to assemble geometries segmented from the higher resolution supine MRI scans. This approach increased the biofidelity of both the shape and placement of structures for improved response in the subsequent models. The field of view and slice thickness for each body region scanned can be seen in Table 1. Examples of the multi-modality image dataset can be seen in Figure 3.

CT scans were acquired using a GE LightSpeed, 16-slice scanner. All images were acquired with the scanner in helical mode, with the subject placed in both supine and quasi-seated postures. To place the subject in the quasi-seated position, custom foam and acrylic inserts were attached to the scanning table. For all scans, the matrix size was 512 mm x 512 mm.

Supine MRI images were acquired using a 1.5 Tesla Twin Speed scanner (GE, Milwaukee, WI). A 3D fast spoiled gradient recalled pulse sequence (FSPGR) was used and the ratio of echo time (TE) to repetition time (TR) was selected to have fat and water signals out of phase. This produced images with an enhanced outline between the viscera, fat, and muscle for improved segmentation. In order to reduce motion artifact during thoracic and abdominal scan acquisition, the volunteer was trained with breath-holding techniques. Scan acquisition parameters were, TR = 5.26 ms, TE = 1.8 ms, flip
angle = 10⁰, and bandwidth = 62.5 MHz. An 8-channel-phased-array body coil was used to collect the majority of the data. For the head and neck however, an 8-channel neurovascular coil was used. Upon completion of image acquisition, all images were reformatted to a matrix size of 512 x 512.

The upright MRI scans were obtained using a 0.6 Tesla Fonar Upright MRI (Fonar, Inc., Melville, NY). Again, 3D gradient pulse sequences were used to place fat and water out of phase. A quadrature head coil and a set of spine and body coils were used for image acquisition. Acquisition parameters for the uMRI were TR = 14.7 ms, TE = 5.6 ms, flip angle = 30⁰, and an acquisition matrix of 200 x 200. Slice thicknesses for acquisition varied from 1.5 mm to 2 mm. Images were acquired in both seated and standing positions. In the seated scans, the seat back angle was set to 23⁰. The images taken in the standing posture were of the shoulder, thorax, abdomen, and load-bearing knee.

<table>
<thead>
<tr>
<th>Region</th>
<th>Modality</th>
<th>Field of View (mm)</th>
<th>Slice Thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head and Neck</td>
<td>CT</td>
<td>250</td>
<td>0.63</td>
</tr>
<tr>
<td></td>
<td>MRI</td>
<td>300</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td>uMRI</td>
<td>320</td>
<td>1.5</td>
</tr>
<tr>
<td>Thorax</td>
<td>CT</td>
<td>500</td>
<td>0.63</td>
</tr>
<tr>
<td></td>
<td>MRI</td>
<td>480</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td>uMRI</td>
<td>400</td>
<td>2</td>
</tr>
<tr>
<td>Upper Extremity</td>
<td>CT</td>
<td>500</td>
<td>0.63</td>
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<tr>
<td></td>
<td>MRI</td>
<td>260</td>
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<td></td>
<td>uMRI</td>
<td>400</td>
<td>2</td>
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<tr>
<td>Abdomen and Pelvis</td>
<td>CT</td>
<td>250</td>
<td>0.63</td>
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<tr>
<td></td>
<td>MRI</td>
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<td>2</td>
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<tr>
<td></td>
<td>uMRI</td>
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<tr>
<td>Lower Extremity</td>
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</tr>
<tr>
<td></td>
<td>MRI</td>
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<tr>
<td></td>
<td>uMRI</td>
<td>250</td>
<td>1.5</td>
</tr>
</tbody>
</table>
Figure 3. Examples of the Multi-Modality Image Set, a) Axial slice of supine CT, b) Axial slice of supine MRI, c) Axial slice of seated uMRI

2.2.2 External Anthropometry

A 7-axis 3D digitizer (Faro, Platinum Model arm, 2.4 m, Lake Mary, FL) was used for the collection of the subject’s surface topography and external landmarks [4]. The landmarks were used to characterize the posture of the subject throughout the body: head (n=9), spine (n=6), thorax (n=9), pelvis (n=5), upper extremity (n=12) and lower extremity (n=14). To facilitate future modeling efforts, this information was collected from the subject in both seated and standing postures. Data were collected in a single session and in the following order: seated landmark data, seated surface data, standing landmark data, and standing surface data.

For proper placement in the driving position, a custom, adjustable seat buck and platform were developed [4]. The seat back and pan angles were set to 23° from vertical and 14.5° from horizontal respectively. The adjustable parameters for the buck can be seen in the appendix (Table A1) [4,19]. Each variable parameter in the buck was set to properly fit the stature of the subject. For future modeling purposes, a local coordinate system was defined using markers on the seat buck with the origin defined as the seat’s H-point. This coordinate system was used to define the final model space. To ensure
access to the full topography of the subject while in the seated position, the seat back of the buck was divided into halves, allowing for mid-sagittal spinal landmarks to be collected.

After positioning the subject in the buck, a visual target was oriented at the subject’s eye level to place the head in the Frankfurt plane. In total, 54 bony landmarks were identified through palpation (Table A2). Directly following landmark data collection, while the subject was still seated in the buck, a scan of the subject’s surface topography was acquired. The scan was taken of the right side of the volunteer’s body, with major structures on the left side, such as the thigh and shoulder, scanned to validate mirroring. In order to reduce artifact in the scan, the subject wore white, non-reflective, form-fitting clothing. In order to obtain un-deformed body contours, the seat back panels and pans could be removed from their docked positions individually. Because of the time required for data collection and the fact that the patient had to remain still throughout, the data collection was divided into sessions of 20 minutes. Standing data were acquire using similar techniques further described by Gayzik et al [4].

The external landmarks were symmetrized by fitting a sagittal symmetry plane to the landmark data. Right and left side data points were mirrored across the plane. The transformations of the points were then averaged to establish the final point locations of the extremities. Points found along the mid-sagittal plane of the axial skeleton were projected along the Y axis to the global XZ plane. These points were then defined per the SAE J211 sign convention that was aligned to the mid-sagittal plane, with the origin at the H-point.
2.2.3 Segmentation

After collection of the medical imaging dataset, the images were used to develop 3D representations of the pertinent F05 anatomy for finite element analysis (FEA) model development. As much structural information as possible was segmented from the obtained medical images. Adjacent image stacks from the same modality were merged into a continuous set of images to properly segment geometries that may have not have been fully represented in one image stack (Amira v5.2.2, FEI, Burlington, MA). Mimics software (v16.0, Materialise, Belgium) was used for all segmentations. Segmentation techniques consisted of a mixture of manual and semi-automatic techniques, with the tissue type dictating the approach. Standard segmentation techniques such as region growing, morphological operations, multi-slice interpolation, and Boolean operations were also employed as needed [3].

Bone

The bones of the body were individually segmented. However, fused bones, such as the ilium, ischium, and pubic bone, were not separated. To segment bony structures, a semi-automatic method using thresholding techniques was employed. Bone segmentation began by selecting pixels exceeding 226 Hounsfield Units. In regions with small articular spaces, such as with the interface between the thoracic spine and the ribs, the structures were manually separated. Initial segmentations of bony structures were performed using supine CT data. This dataset was preferred for the initial development of 3D bone data due to its high resolution and contrast. To promote accurate assembly, the 3D polygon data obtained from the supine CT was then imported into the image space
of the seated CT and aligned with the corresponding structures using affine transformations. This allowed the higher resolution scans to be used for segmentation and the seated scans to be used for more accurate placement. One exception to the skeletal placement was the lumbar spine. In this case, the curvature of the grouped lumbar vertebrae from the quasi-seated CT dataset did not capture the seated posture well. Therefore, in this instance, the lumbar spine was aligned to curvature found in the seated uMRI.

**Organs**

The majority of organ segmentations were manually completed using standard segmenting techniques, including flood fills, region growing, and multi-slice interpolation. Supine CT data was used for segmentation of the thoracic organs because the high contrast between the air filled lungs and the surrounding tissues was easily identified via thresholding within the CT scan. In order to outline the heart, a contrast enhanced scan was selected from the Wake Forest University image database and anonymized. The individual was female, with a height of 149 cm, and weight of 49 kg. All remaining organs were segmenting using supine MRI data.

Apart from the white matter, all brain structures that will be used in the model were manually segmented. The white matter was automatically segmented using statistical parametric mapping software (SPM5, Functional Imaging Laboratory, University College London). After the voxels had been selected, they were then transformed back into the subject image space and verified against the subject image set.
In order to account for postural effects on the abdominal organs, after segmentation using the supine MRI data, the 3D surfaces of the abdominal organs were transformed into the upright MRI image space using affine transformations. The surfaces from the supine data were then used as a basis to adjust for the organ shape variation found in the seated posture. This approach was taken to apply the strengths of both scans to the data set. The supine MRI data was preferable for initial segmentation due to its field strength and higher resolution and the seated uMRI was used for its accurate placement and shape. The seated uMRI scans were also used to capture the correct orientation of major abdominal vasculature (such as the inferior vena cava and aorta), the colon, and the small bowel.

2.2.4 Assembly

Following completion of bony segmentation and alignment in the seated scans, the skeletal structures were assembled in the model coordinate system. This was completed using affine transformations from the medical image space to align the structures to the bony landmark data in the model space (SAE J211). In order to ensure that characteristics such as spinal curvature were maintained during assembly, the skeleton was moved in segments consisting of the cervical spine, thoracic spine and ribs, and the lumbar spine. The sacrum was also assembled as a segment with the pelvis. An example of the placement and assembly of the cervical spine can be seen in Figure 4.
The supine MRI scan was used for assembly of brain structures. To assemble the brain, the skull was first segmented from the supine CT scans and aligned within the supine MRI. Due to contrast deficiencies, the skull could not be directly segmented from the MRI data. The skull was then used to develop a transform from the MRI space to the model space. This transform was used to bring brain structures into the model coordinate system.

To complete the assembly of abdominal organs, the axial skeleton segmented from the supine CT scans was aligned within the seated uMRI scan (Figure 5). Model assembly utilized the placement of the skeletal structures within the seated uMRI to develop transforms for soft tissue in the uMRI image space to
the model space. Because the shape and volume of bony structures does not change relative to posture, specific bones or bony segments were used to develop transforms by aligning the bones from the image space to the assembled skeleton in the model coordinate system. For example, a skeletal segment containing portions of the thoracic spine and lumbar spine (T8-L3) and ribs 8-12 was transformed from the upright MRI image space to the assembled model skeleton using a best fit alignment algorithm within Studio software (2014.1.1, Geomagic Inc., Morrisville, NC). The resulting transform was then applied to the solid organs of the upper abdomen (liver, kidneys, gallbladder, spleen, stomach, and pancreas). Best fit alignments using the pelvis and lumbar spine were used for placement of the colon, small bowel, and bladder.

2.3 RESULTS

The selected volunteer (24 years old, female) was a good fit in terms of height and weight (149.9 cm, 48.1 kg) with deviations from the target values of 0.7% and 1.9% respectively. In addition to an initial screening, the subject was also evaluated based on 15 anthropomorphic measurements from Gordon et al [24]. For all 15 measurements, the average subject percent deviation was 4.1% (cutoff for inclusion was 5%). A summary of each measurement can be seen in Table 2. Following recruitment, 54 bony landmarks were recorded in both the seated and standing postures via palpation. A list of the anatomical location of each bony landmark and their abbreviations can be seen in the appendix (Table A2). The X, Y, and Z coordinates of each bony landmark in the SAE J211 coordinate system for both seated and standing postures are reported in Table A3.
Table 2. Anthropometric measurements of the 5th percentile female volunteer

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Subject</th>
<th>Target [24]</th>
<th>Deviation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sitting Height</td>
<td>80.0</td>
<td>79.5</td>
<td>0.6</td>
</tr>
<tr>
<td>Hip Breadth Sitting</td>
<td>35.6</td>
<td>34.2</td>
<td>3.9</td>
</tr>
<tr>
<td>Buttock Knee Length Sitting</td>
<td>52.8</td>
<td>54.2</td>
<td>-2.5</td>
</tr>
<tr>
<td>Knee Height Sitting</td>
<td>46.0</td>
<td>47.4</td>
<td>-3.0</td>
</tr>
<tr>
<td>Bideltoid Breadth, Sitting</td>
<td>40.6</td>
<td>39.7</td>
<td>2.4</td>
</tr>
<tr>
<td>Shoulder-Elbow Length, Standing</td>
<td>33.0</td>
<td>30.8</td>
<td>7.3</td>
</tr>
<tr>
<td>Forearm-Hand Length, Standing</td>
<td>40.4</td>
<td>40.6</td>
<td>-0.6</td>
</tr>
<tr>
<td>Waist Circumference, Standing</td>
<td>78.2</td>
<td>67.6</td>
<td>15.8</td>
</tr>
<tr>
<td>Hip Breadth, Standing</td>
<td>31.2</td>
<td>30.8</td>
<td>1.5</td>
</tr>
<tr>
<td>Foot Length, Standing</td>
<td>22.6</td>
<td>22.4</td>
<td>0.8</td>
</tr>
<tr>
<td>Head Breadth</td>
<td>14.5</td>
<td>13.7</td>
<td>5.9</td>
</tr>
<tr>
<td>Head Length</td>
<td>18.8</td>
<td>17.6</td>
<td>6.6</td>
</tr>
<tr>
<td>Head Circumference</td>
<td>53.6</td>
<td>52.2</td>
<td>2.6</td>
</tr>
<tr>
<td>Chest Circumference</td>
<td>83.8</td>
<td>81.4</td>
<td>3.0</td>
</tr>
<tr>
<td>Neck Circumference</td>
<td>30.5</td>
<td>29.2</td>
<td>4.3</td>
</tr>
</tbody>
</table>

In total, 66 scan series were collected across all modalities for a total of 14,170 images. This included CT scans in supine and quasi-seated postures and MRI scans in the supine, seated, and standing postures. 3D geometries were developed for all skeletal structures and each organ that will be explicitly modeled. The skeleton consists of 182 individual bones. With subsequent computational models in mind, geometrical features that are important for accuracy in biomechanical modeling efforts were captured from the scan data.

Explicit representations of 32 organs have also been developed, represented by brain, thoracic, and abdominal organs germane to biomechanical modeling. The following 16
structures are represented within the model of the brain: left and right cerebral hemispheres, corpus callosum, ventricles (3rd, 4th, and lateral), brainstem, fornix, thalamus, cerebellum, falx, tentorium, left and right basal ganglia, and the venous sinuses (transverse and superior). Within the thorax and abdomen, 3D representations of the heart, right and left lungs, liver, spleen, right and left kidneys, gallbladder, stomach, colon, duodenum, pancreas, and the bladder have been developed. Due to its complicated geometry, the small bowel was developed as a control volume. Volumetric data for a selection of modeled organs can be seen in Table 3. Major vasculature is also represented in these regions, including the aorta, vena cava, and hepatic portal vein. The aorta and vena cava were each measured at three discrete points to obtain diametric measurements. The aorta was measured at its superior exit from the heart (26.7 mm), at the inferior portion of the aortic arch (20.1 mm), and parallel to the superior surface of L3 (13.3 mm). The measurements for the vena cava were taken at the superior exit from the heart (17.8 mm), the inferior exit from the liver (20.8 mm), and parallel to the superior surface of L3 (18.5 mm). The primary and secondary branches of the aorta and vena cava were also modeled to include the natural tethers that they provide to organs in the human body. The exterior skin of the model has been developed as a single surface from the external anthropometry laser scanning. The surface area of the skin was 1.46 m², which is within 3% of estimates in the literature for a female of the same height and body weight [26,27]. The assembled model can be seen in Figure 6.
Table 3. Volumetric organ data of the F05 from segmentation

<table>
<thead>
<tr>
<th>Structure</th>
<th>Volume (ml)</th>
<th>Structure</th>
<th>Volume (ml)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left Lung</td>
<td>103.9</td>
<td>Basal Ganglia</td>
<td>15.6</td>
</tr>
<tr>
<td>Right Lung</td>
<td>119.6</td>
<td>Brainstem</td>
<td>31.4</td>
</tr>
<tr>
<td>Heart</td>
<td>56.8</td>
<td>Cerebellum</td>
<td>132.8</td>
</tr>
<tr>
<td>Liver</td>
<td>1024.7</td>
<td>Corpus Callosum</td>
<td>17.4</td>
</tr>
<tr>
<td>Spleen</td>
<td>131.2</td>
<td>Fornix</td>
<td>0.97</td>
</tr>
<tr>
<td>Right Kidney</td>
<td>112.1</td>
<td>Superior Sagittal Sinus</td>
<td>14.9</td>
</tr>
<tr>
<td>Left Kidney</td>
<td>122.3</td>
<td>Thalamus</td>
<td>17.8</td>
</tr>
<tr>
<td>Pancreas</td>
<td>52.1</td>
<td>Lateral Ventricles</td>
<td>10.1</td>
</tr>
<tr>
<td>Bladder</td>
<td>44.2</td>
<td>Third Ventricle</td>
<td>0.95</td>
</tr>
<tr>
<td>Gallbladder</td>
<td>19.1</td>
<td>Fourth Ventricle</td>
<td>2.2</td>
</tr>
</tbody>
</table>

Figure 6. Assembled 3D F05 Occupant. Images show the full skeleton and all organs that will be explicitly represented. Bony landmark data is shown as blue points (Table A3).

2.4 Discussion

This study presents a comprehensive set of image and anthropometrical data of the F05. This included the use of state of the art imaging techniques, surface scanning, and 3D digitization for collection of bony landmark data. A thoroughly pre-screened
volunteer representative of the F05 was selected for the purpose of developing and assembling geometries of a small female in two postures. This approach was advantageous for model development because of the overall size and versatility of the dataset. Ultimately, all bony structures and the majority of soft tissue data that will be modeled were obtained using medical images of this subject. In order to accurately assemble the structures, the full dataset was leveraged, where appropriate, to facilitate model development. Assembly of skeletal structures in the model coordinate system was completed using the external anthropometry dataset and bony structures placed in the seated CT scans. Seated uMRI scans were used for assembly of abdominal organs to ensure correct shape and position.

Height and weight requirements for the subject were taken from nominal values utilized for development of the Hybrid III F05 anthropomorphic test device (ATD). This approach was taken because the response of the subsequent finite element model of the F05 will ultimately be compared to the F05 ATD. This allows for direct comparison to an ATD that is already an integral part of the regulatory process for crash tests. However, subject recruitment was also heavily based on anthropomorphic measurements taken from a survey of U.S. armed service personnel, known as the U.S. Army Anthropometry SURvey (ANSUR) [24]. This dataset was ultimately used because of its large size and the comprehensive nature of the measurements it contains. While it is noted that the population reviewed as part of ANSUR is not necessarily equivalent to the average world anthropometry measurements, at the time of this study, it was the most complete and thorough dataset available. In total, the ANSUR study screened over
25,000 subjects and ultimately reported specific measurements on 2,208 women averaging 26.19 years old.

The assembled skeleton of the female model was also compared to a finite element model of the Hybrid III F05. The Hybrid III ATD model was obtained from Livermore Software Technology Corp (LSTC) and measurements were taken using nodal distances. For the observed anthropometry, the segmented F05 had an average deviation compared to the HIII F05 of 2.7%. The measurements used for the comparison can be seen in Figure 7.

![Figure 7. Comparison of F05 model assembly to F05 Hybrid III finite element model](image)

While data for a specific anthropometric population such as the F05 is limited, comparisons were also made to literature sources where available. In 2004, Geraghty et
al. published the normal distribution of female abdominal organ volumes from the 1\textsuperscript{st} percentile to the 99\textsuperscript{th} percentile using data obtained from clinical CT scans [28]. However, this distribution is representative of organ volume specifically and not anthropomorphic percentiles. For example, a female that is anthropomorphically representative of the F05 does not necessarily have internal organs that fall at the 5\textsuperscript{th} percentile of organ volumes. Despite this, it is currently the best data source to compare organ volumes of the F05. For modeling purposes, comparisons were made of organs germane to crash induced injuries (CII’s). Typically, organs of interest for CII’s are considered the liver, spleen, left kidney, and right kidney. The liver volume was found to closely match the 5\textsuperscript{th} percentile and the right and left kidneys fell in the 10\textsuperscript{th} percentile of organ volumes determined in the population based study. The spleen volume showed the largest deviation falling in the 20\textsuperscript{th} percentile of female spleen volumes. However, it is difficult to assess whether or not these organ volumes are uncommon for a F05. For instance, it may not be unusual for anatomic variability to cause individuals of a select anthropometry to have widely ranging organ volumes.

Initial segmentations of cortical bone thickness in specific long bones (humerus, radius, ulna, femur, fibula, tibia, and femur) were also compared to published data. The dataset was compared to a review of radiographic measurements of cortical bone published by Virtama et al [29]. While the cortical thicknesses reported in this dataset are not specifically related to the F05, they do provide a good initial point of comparison. The results of this comparison can be seen in Table 4. In each case, the literature values were obtained from 25 year old subjects and all values were taken from right limbs at
specified, discrete locations. For each bone, the cortical thickness is within the observed female range.

<table>
<thead>
<tr>
<th>Bone</th>
<th>F05 Subject (mm)</th>
<th>Virtama et al. [29]</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Average (mm)</td>
<td>Minimum (mm)</td>
<td>Maximum (mm)</td>
</tr>
<tr>
<td>Humerus</td>
<td>4.9</td>
<td>4.4</td>
<td>2.95</td>
</tr>
<tr>
<td>Radius</td>
<td>3.5</td>
<td>3.3</td>
<td>2.1</td>
</tr>
<tr>
<td>Ulna</td>
<td>3.7</td>
<td>3.5</td>
<td>2.5</td>
</tr>
<tr>
<td>Femur</td>
<td>8.4</td>
<td>8.5</td>
<td>6.7</td>
</tr>
<tr>
<td>Tibia</td>
<td>5.5</td>
<td>5.75</td>
<td>3.8</td>
</tr>
<tr>
<td>Fibula</td>
<td>4.0</td>
<td>3.3</td>
<td>2.15</td>
</tr>
</tbody>
</table>

In order to more accurately evaluate the development of the F05, additional data pertinent to the small female is required. Future work will involve developing datasets specifically representative of the F05 population. For example, organ volumes specifically measured within the F05 population are necessary to draw a true comparison to a subject selected based on anthropometry. Similarly, cortical thickness measurements from anthropometrically representative females would be useful to increase the accuracy of fracture prediction within subsequently developed models.

The data and techniques outlined in this paper have focused on the assembly and validation of anatomical structures germane to the modeling of CII’s, namely bony structures and organs. However, other structures designed to facilitate passive load transfer and promote accurate kinematics will also be modeled. Based on previous work for the average male dataset, roughly 96 muscles and 26 ligaments, tendons, and other cartilaginous tissues are targeted for inclusion in the final model [3]. With regards to model development, cartilaginous tissue will be created during the meshing phase where
possible, reducing the need for CAD representations. Because the dataset will ultimately be used for the evaluation of tissue response to blunt impact, much of the microstructure of the human body has not been included. This approach was taken since final model validation will be compared to empirical data obtained from experiments conducted at the organ or full-body levels.

2.5 CONCLUSIONS

This study presents a methodology and comprehensive dataset for the development of a 5th percentile female finite element model. The data were collected using a multi-modality medical imaging protocol and a custom, adjustable buck for collection of external landmarks. The dataset is versatile and was obtained to accurately assemble the model in both the occupant and pedestrian postures. Preliminary segmentation work shows that volumetric organ data and cortical bone thickness from the prospectively recruited F05 subject reasonably matches available population based studies. Ultimately, this data will be used as part of a larger effort in developing a detailed finite element model of the 5th percentile female for human injury prediction in vehicular crash.

2.6 ACKNOWLEDGEMENT

Funding for this study was provided by the Global Human Body Models Consortium, LLC (GHBMC) through GHBMC Project Number: WFU-005. Support for CAD generation provided by Zygote Media Group, Inc. (American Fork, Utah).
### 2.7 Appendix

**Table A1. Adjustable buck parameters [4]**

<table>
<thead>
<tr>
<th>Toe Board Pitch (°)</th>
<th>Steering Wheel Pitch (°)</th>
<th>Steering Wheel Height (cm)</th>
<th>Wheel to Ball of Foot (cm)</th>
<th>Seat Position (cm)</th>
<th>Hell Riser Height (in)</th>
</tr>
</thead>
<tbody>
<tr>
<td>58</td>
<td>16</td>
<td>63.5</td>
<td>39</td>
<td>59</td>
<td>0.5</td>
</tr>
</tbody>
</table>

**Table A2. Bony landmarks acquired for model assembly**

<table>
<thead>
<tr>
<th>Landmark</th>
<th>Abbreviation</th>
<th>Bone(s) used for landmark determination</th>
<th>Landmark</th>
<th>Abbreviation</th>
<th>Bone(s) used for landmark determination</th>
</tr>
</thead>
<tbody>
<tr>
<td>Top of Head</td>
<td>TH</td>
<td>Skull</td>
<td>7th Cervical Vertebrae (C7)</td>
<td>C7</td>
<td>C7</td>
</tr>
<tr>
<td>Back of Head</td>
<td>BH</td>
<td>Skull</td>
<td>4th Thoracic Vertebrae (T4)</td>
<td>T4</td>
<td>T4</td>
</tr>
<tr>
<td>Tragion*</td>
<td>T</td>
<td>Skull</td>
<td>8th Thoracic Vertebrae (T8)</td>
<td>T8</td>
<td>T8</td>
</tr>
<tr>
<td>Glabella</td>
<td>G</td>
<td>Skull</td>
<td>12th Thoracic Vertebrae (T12)</td>
<td>T12</td>
<td>T12</td>
</tr>
<tr>
<td>Infraorbitale*</td>
<td>I</td>
<td>Skull</td>
<td>3rd Lumbar Vertebrae (L3)</td>
<td>L3</td>
<td>L3</td>
</tr>
<tr>
<td>Corner of Eye*</td>
<td>CE</td>
<td>Skull</td>
<td>5th Lumbar Vertebrae (L5)</td>
<td>L5</td>
<td>L5</td>
</tr>
<tr>
<td>Lat. Clavicle*</td>
<td>LatC</td>
<td>Clavicle, Scapula</td>
<td>Radial Styloid*</td>
<td>RS</td>
<td>Radius, Trapeziun</td>
</tr>
<tr>
<td>Suprasternale (Manubrium)</td>
<td>SSM</td>
<td>Sternum</td>
<td>5th Metacarpal*</td>
<td>FM</td>
<td>5th Metacarpal, 5th Prox. Phalange</td>
</tr>
<tr>
<td>Substernale (Xyphoid Process)</td>
<td>SSX</td>
<td>Sternum</td>
<td>Femoral Condyle*</td>
<td>LFC</td>
<td>Femur, Fibula</td>
</tr>
<tr>
<td>Lat. Humeral Condyle*</td>
<td>LHC</td>
<td>Humerus, Radius</td>
<td>Femoral Condyle*</td>
<td>MFC</td>
<td>Femur, Tibia</td>
</tr>
<tr>
<td>Med. Humeral Condyle*</td>
<td>MHC</td>
<td>Humerus, Ulna</td>
<td>Supra-patella*</td>
<td>SP</td>
<td>Patella</td>
</tr>
<tr>
<td>Ulnar Styloid*</td>
<td>US</td>
<td>Ulna, Triquetral</td>
<td>Lat. Malleolus*</td>
<td>LM</td>
<td>Tibia, Calcaneus, Foot Complex*</td>
</tr>
<tr>
<td>Anterior Superior Iliac Spine*</td>
<td>ASIS</td>
<td>Pelvis</td>
<td>Med. Malleolus*</td>
<td>MM</td>
<td>Tibia, Talus, Foot Complex</td>
</tr>
<tr>
<td>Landmark</td>
<td>Abbreviation</td>
<td>Domain</td>
<td>Description</td>
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<td>----------------------------------</td>
<td>--------------</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Posterior Superior Iliac Spine*</td>
<td>PSIS</td>
<td>Pelvis</td>
<td>Ball of Foot*</td>
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<td></td>
</tr>
<tr>
<td>Pubic Symphysis</td>
<td>PS</td>
<td>Pelvis</td>
<td>5th Metatarsal*</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>5th Metatarsal, 5th Prox. Phalange, Foot Complex</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Calcaneus*</td>
<td>C</td>
<td></td>
<td>Calcaneus, Foot Complex</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* Indicates the landmark measurement was taken on both the right and left side of the body
+ Foot Complex was treated as a single unit for placement in final model space
<table>
<thead>
<tr>
<th>Landmark Name</th>
<th>Seated X</th>
<th>Seated Y</th>
<th>Seated Z</th>
<th>Standing X</th>
<th>Standing Y</th>
<th>Standing Z</th>
</tr>
</thead>
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Figure A1. Abdominal CAD represented in the GHBMC F05
Figure A2. Brain structures modeled in the GHBMC F05
2.8 REFERENCES


Chapter III: Characterization of Thoracoabdominal Organ Volumes for the 5th Percentile Female

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ABSTRACT

Objective: Thoracoabdominal injuries commonly occur as a result of motor vehicle crashes. In order to mitigate the severity of these accidents, researchers are using a variety of tools, including computational human body models. While research has been conducted to provide morphological and volumetric data for the thoracoabdominal cavity of the average male, there is currently an interest in developing models for a wider range of occupants. One particular cohort of interest is the 5th percentile female by height and weight because of their use as a conservative estimate for the performance of vehicle safety devices. Therefore this study aims to quantify prominent anatomical features of the thoracoabdominal cavity a 5th percentile female in terms of height and weight (F05). To develop the dataset representative of the thoracoabdominal contents of an F05, volumetric organ measurements were taken using clinical CT and MRI datasets.

Methods: Anonymized clinical CT and MRI images were used to segment organs relevant to crash induced injuries: namely the liver, spleen, left kidney, right kidney, and heart. Segmentations were conducted using semi-automatic techniques. Following segmentations, all organs were rendered into 3D geometries for increased accuracy in volume calculation. Additionally, diametric measurements of the vena cava and aorta were obtained from the medical images in 3 discrete locations using linear measurement tools.

Results: A total of 11 adult scans were selected with an average subject height and weight of 51.4 ± 4.0 kg and 153.6 ± 6.3 cm respectively. Volumetric measurements were
obtained via segmentation for the following organs: the liver (1194.1 ± 213.2 ml), spleen (157.7 ± 87.1 ml), left kidney (118.7 ± 24.6 ml), right kidney (113.0 ± 21.4 ml), and heart (397.9 ± 20.5 ml). The average diameters were 19.1 ± 3 mm and 17.4 ± 6.0 mm for the vena cava and aorta respectively.

**Conclusion:** There is currently no dataset that characterizes organ volumes specifically for females in the 5th percentile of height and weight. However, this data is important in validating the geometries used in computational models of the small female, including models both derived from scaling techniques and those developed using subject specific medical imaging. The goal of this study was to use a sample of females anthropometrically representative of the 5th percentile female to evaluate the average volume for organs commonly injured in motor vehicle crashes. Ultimately, these measurements will be useful for the validation of computational models of the small female.

**Keywords** – Small female, 5th percentile female, modeling, abdominal organ volume, segmentation
3.1 INTRODUCTION

Motor vehicle crashes commonly result in blunt thoracoabdominal trauma. Specifically, injuries to the thorax and abdomen are found to rank second and third respectively behind head injuries in terms of the commonly injured body regions and economic costs [1,2]. It is estimated that 19,000 such injuries occur within the United States each year [1]. Abdominal trauma also accounts for 16.5% of AIS 4+ injuries and 20.5% of AIS 5+ injuries [3,4]. Despite the prevalence and significance of these injuries, no current anthropomorphic test devices (ATDs) are equipped to represent or provide organ specific response to impact loading. This lack of structure specific representations stems from the large bio-variability with regards to thoracoabdominal organs [5]. Also, there are currently no established standards for inferring injury to these soft tissues.

Computational human body modeling is a growing field within biomechanics that has the capability of assessing localized responses of internal organs. One such tool being used by researchers to assess the risk of injury to thoracoabdominal organs is the finite element method. The past 20 years has seen a large growth in the number of full body finite element models used to evaluate blunt impact loading [6]. Traditionally, these models were explicitly developed to represent an average male (50th percentile in terms of height and weight). While these models can give excellent assessment of the mid-sized adult male, real world motor vehicle crashes involve occupants of various size, age, and gender. In a recent study observing road traffic gender trends, there are now more licensed female drivers than male drivers in the United States [7]. With these shifting demographics in motorists, it is important that models be developed to represent
the expanding cohort of drivers. This study focuses on the female driver in the 5th percentile of height and weight.

Accurate modeling of the small female is important because it is generally believed that they are at a greater risk of sustaining automotive related injury compared to the average sized male. This is a result of their preferred seated posture closer to the wheel and lower structural strength [8-10]. Based on an analysis using the National Automotive Sampling System (NASS) in the United States and the Cooperative Crash Injury Study (CCIS) in the United Kingdom, Mackay et al. found that at any severity level on the Abbreviate Injury Scale (AIS), the median tolerable delta-V was considerably lower for females than for males [11]. Work completed by Tavris et al. also suggests that female occupants are more susceptible to injury during collisions [12]. With regards to specific injuries, work from Welsh et al. found that small females are more likely to obtain chest injuries with a maximum AIS of 2 and above and Dischinger et al. found that they are at a greater risk to sustain fractures to the lower extremities [13,14]. Because of their increased risk of injury, the response of the small female generally serves as a conservative estimate for safety design relative to the general public.

In the past, 5th percentile female models have primarily been developed by scaling existing 50th percentile male models, since scan data for such a specific target anthropometry is limited. The techniques used to scale these models rely on data from external anthropometry databases [10,15,16]. While these models accurately capture the external anthropometry, internal organs volumes must be compared to literature in order to ensure that they were appropriately scaled. A similar comparison is also appropriate when the model is developed using scans of a specific F05 subject because of the
biodiversity of human subjects. However, there is currently no dataset that specifically addresses the volumes of thoracoabdominal organs for the 5\textsuperscript{th} percentile female. Due to the role that changes in organ volume can play with regards to the inertial response of the abdomen and thorax, it is important in the development of these models that the internal organs be accurately represented.

Medical images are commonly used to assess organ volumes \textit{in vivo} [17-20]. Currently, one of the best datasets available for organ volume comparison for development of these size specific models is work published by Geraghty et al [21]. In this study, the authors reported the normal distribution of female organ volumes for the 1\textsuperscript{st} percentile organ volume to the 99\textsuperscript{th} percentile. However, these values are not necessarily representative of a female subject of a specific height and weight. For example, a 5\textsuperscript{th} percentile female in terms of height and weight may not necessarily have organs that fall in the 5\textsuperscript{th} percentile of a normal distribution of organ volumes. In order to bridge this gap, the objective of this study was to characterize the volumes of abdominal organs using a sample of individuals specifically representative of the 5\textsuperscript{th} percentile female. Data acquisition focused on organs that are relevant to crash induced injuries (CII’s), namely the liver, spleen, right kidney, left kidney, and the heart. In addition, data was collected on large thoracoabdominal vasculature. This dataset will provide a valuable tool for the development and anatomical accuracy of future finite element models of the 5\textsuperscript{th} percentile female.
3.2 METHODS

The medical images used in this study were obtained from the radiological database at Wake Forest University Baptist Health. To be included in the study, patients were required to fit the height and weight requirements of the F05 (150 ± 12 cm and 49 ± 8 kg), be free of abdominal or systemic injuries, and have all organs present. All images were anonymized and approved for use under Wake Forest University’s Institutional Review Board (IRB# 00006511). In total, 1 magnetic resonance imaging (MRI) scan and 10 contrast-enhanced computed tomography (CT) scans of the chest, abdomen, and pelvis were used. The CT scans were acquired for routine clinical evaluation with a patient population comprised of inpatients and outpatients. The MRI scan of the abdomen was completed as part of a larger dataset obtained explicitly for model development [22,23]. In this case, the subject was selected specifically as a representative F05 for the development of the Global Human Body Models Consortium’s (GHBMC) fully body finite element model of the small female. The radiologist’s report for each examination was reviewed and careful study of the patient’s medical records and diagnosis were used to determine normalcy. Patients whose organs may have been affected by local or systemic disease or injuries were not used. Also, cases where the CT examination or radiologist’s report indicated trauma or surgery in the area of the proximity of the abdomen were excluded. Additional exclusion criteria included the presence of calcification or areas of high attenuation indicative of tumors. Within the selected scans, no abnormal features, such as blurring or movement artifact were seen. Selected patients were between the ages of 18 and 50 to reduce the effects of age related volume changes.
The CT scans were acquired using a GE LightSpeed, 16-slice scanner. Scanning parameters were dependent on the clinical indication with which the patient presented. Images were acquired in the helical mode with slice thicknesses ranging from 0.65 mm to 2.5 mm and all studies were contrast enhanced. The MRI data was collected on a 1.5 Tesla Twin Speed scanner (GE, Milwaukee, WI) [22]. A 3D Fast Spoiled Gradient Recalled pulse sequence was used with the echo time (TE) and repetition time (TR) ratio selected such that fat and water signals were out of phase, enhancing the ability to distinguish organ boundaries.

The DICOM headers for each subject were also reviewed prior to evaluation to ensure the correct scans had been pulled. All segmentations were performed using Mimics software (v. 16.0, Materialise, Leuven, Belgium) and were visualized using a Wacom Cintiq 22HD digitizing monitor for enhanced accuracy. Window levels were set using default values for soft-tissue within Mimics (Min: 874, Max: 1374) to promote consistent segmentations. The organs were manually segmented using standard segmenting tools, such as outlining, flood-filling, region growing, and morphological and Boolean operations.

Segmentation was limited to thoracoabdominal organs germane to CIIs, namely the liver, spleen, right kidney, left kidney, and heart. In each case, multi-slice interpolations using sections of 2-5 slices were used to segment the organs. As per Geraghty et al., caution was exercised while segmenting the liver and the kidney to avoid the inclusion of any extraneous volume [21].

With regards to the liver, the inferior vena cava was excluded from the segmentation. This was done by manually segmenting the vena cava, and then using
Boolean operations to remove it from the mask of the liver. The hepatic veins however, are considered intraparenchymal and were therefore included in the total volume of the liver. In the case of the portal venous system, regions in which the vasculature was intrinsic to the liver were included, while regions protruding from the liver were removed from the segmentation. Also, regions were removed where the longitudinal and portal fissures of the liver were large. In the kidneys, the collecting system and vasculature were not included in the mask. Ultimately, this resulted in a mask of only the medulla and cortex for final evaluation. Additionally, heart segmentations were limited to areas of myocardial tissue and were truncated at the great vessels. An example of final masks can be seen in Figure 8. The images in this figure were acquired from a scan with a slice thickness of 0.625 mm.
Figure 8- Masks of Liver, Spleen, Left Kidney, Right Kidney, and Heart. Images are ordered from superior to inferior.

After segmentation, each organ was rendered in 3D and used for volume calculations using optimal resolution settings within Mimics. After rendering, the contour of the 3D surface of each organ was used to compare to the original mask. The
volume calculations for the regions of interest were completed using algorithms within Mimics. This employs a merging cubes algorithm and smoothing parameters to accurately evaluate the areas between slices. Examples of the rendered organs can be seen in Table 5.

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</table>

In addition to volume measurements, diametric measurements of the vena cava and aorta were also included. The measurements were taken using linear measurement tools within Mimics and were acquired in the axial view to evaluate maximum diameter. In both vessels, the diameters were evaluated at three discrete locations. For the vena cava, the first measurement was taken at the superior portion of the vena cava, at the point of its exit from the heart. The second measurement was taken at the vena cava’s inferior exit from the liver. The last measurement was taken at the level of the superior surface of L3. With regards to the aorta, the first measurement was taken at the exit of the aorta from the heart and the second was taken from the inferior portion of the aortic arch, at the superior portion of the descending aorta. Similar to the vena cava, the last measurement for the aorta was taken at the level of the superior surface of L3. This location was chosen for the vena cava and the aorta because, across all subjects, it was far
enough from vessel bifurcation that the splitting did not have a confounding effect on the diameter when comparing subjects. An example of the location for diameter measurements of the aorta can be seen in Figure 9.

![Image of aorta measurements](image)

**Figure 9- Sagittal View of Aorta Measurements.** From left to right, the images depict the superior aorta, inferior portion of the aortic arch, and the aorta at the superior level of L3.

### 3.3 Results

In total, the medical images of 11 subjects representative of a 5th percentile female were reviewed. The average height and weight of the selected subjects was 153 ± 6.3 cm and 51.3 ± 4.1 kg respectively. These measurements had target deviations of 2.4% and 4.8% respectively. Height, weight, and slice thickness values for each subject can be seen in Table 6.
Table 6- Height, Weight, and Slice Thickness for Each Subject

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age (yrs)</th>
<th>Weight (kg)</th>
<th>Height (cm)</th>
<th>Slice thickness (mm)</th>
</tr>
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<tbody>
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<td>1</td>
<td>25</td>
<td>49.9</td>
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<td>0.625</td>
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<td>2</td>
<td>29</td>
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<td>149.9</td>
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<tr>
<td>4</td>
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<td>SD</td>
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</table>

For each subject, the spleen, liver, right kidney, left kidney, and heart were segmented and their 3D volume was calculated. The liver for the 5th percentile females had an average volume of $1194.1 \pm 213.2$ ml. The spleen was most variable with regards to total volume, with an average volume of $133.6 \pm 36.6$ ml. The average volumes of the left and right kidneys were $118.7 \pm 24.6$ ml and $113.0 \pm 21.4$ ml respectively. The heart was the least variable with an average volume of $397 \pm 20.5$ ml. Subject specific organ volumes can be seen in Table 7.

Measurements were also taken to evaluate the diameter of the vena cava and the aorta. In some subjects, the vasculature was difficult to visualize due to artifact or incomplete representation of the superior aspect of the vessels in the scans. In these cases, the diameter measurements at that level were excluded. This resulted in the exclusion of a relatively modest number of data points, 4 vena cava (out of 33) and 2 aorta (out of 33). The average diameter of the vena cava was found to be $19.1 \pm 3.7$ mm.
The average diameter of the aorta was $17.4 \pm 6.0$ mm. Values of the diameter of the vessels at each location can also be seen in Figure 10. Within this figure, VC Superior and Aorta Superior correspond to the measurement of the vena cava and aorta taken at their superior exits from the heart respectively. VC inferior refers to the measurements taken at the inferior exit of the vena cava from the liver. The aorta inferior measurements refer to values obtained from the inferior portion of the aortic arch. Lastly, VC L3 and Aorta L3 refer to values taken at the level of the superior surface of the 3rd lumbar vertebrae.

![Figure 10](image)

**Figure 10-** (Left) Average organ volumes, (Right) Average diameters for the vena cava and aorta

### 3.4 DISCUSSION

Medical images have proven to be a valuable diagnostic tool for evaluating organ volumes *in vivo* [17,24-26]. Advances in software algorithms have aided in both the speed and accuracy of obtaining organ specific volumes [27]. Also, as the resolution of
medical images has increased, it has made it possible to more accurately distinguish the border between organ parenchyma and fatty tissue. Using computer aided techniques in commercial software such as flood filling, multi-slice interpolation, morphology operations, and region growing also increases the throughput of organ modeling. These advancements make the development of datasets for specific anthropomorphic populations more feasible.

While individual organ size is dependent on a number of factors, it is common for height and weight to play a role in determining volume [21]. To account for this effect, Geraghty et al. reported the normal distribution of female organ volumes by correcting for height and weight. This allowed a number of patients from a wide range of body sizes to be evaluated together. Using this dataset, organ volumes can be compared to sizes ranging from the 1st percentile to the 99th percentile. However, it is unlikely that these ranges can be considered indicative of anthropomorphic percentiles. Therefore, it is important to characterize the organ size at a specific anthropometry of interest.

The data obtained in this study indicates that the sample of 5th percentile females in terms of height and weight has slightly larger organs than the reported 5th percentile organ volume in Geraghty et al. (Figure 11). The organ with the largest variance from the distribution was the spleen. For the F05, the average spleen value of 157.7 ± 87.1 ml was more than twice the reported value of 71.1 ml. Also, no splenic volumes were below the published 5th percentile volume. The minimum splenic value observed of 89 ml was nearly in nearly the 10th percentile of spleen volumes. Compared to the normal distribution of organ volumes, the average splenic volume for the F05 fell into the 35th percentile of female organ volumes. The average F05 liver was 1194.1 ml compared to
the normal distribution 5th percentile of 974.9 ml and only one subject had a liver volume that fell into the published range of the 5th percentile liver. Similar trends were seen with the right and left kidneys. For the liver and the left kidney, the average F05 volume was in the 10th percentile of normal female organs. The right kidney fell into the 15th percentile of normal organ volumes.

Figure 11- Comparison of F05 organ volumes to Geraghty et al. Organ volumes were normalized to the published target 5th percentile organ volume.

In addition to the abdominal organs, volumetric measurements of the heart were also obtained. The heart is important from an injury biomechanics standpoint because of its role in crash induced injuries such as aortic tear. While injuries to the lungs are also commonly sustained from blunt impact loading, no data was collected on these organs due to their high volumetric variability. Also, for modeling purposes, the space occupied by the lungs is effectively delineated by the mediastinum, diaphragm, and ribs, which allows for accurate representation of these organs from a modeling perspective.

Because of their role as organ tethers, accurate modeling of large thoracoabdominal vasculature is also important from an injury biomechanics standpoint.
It has been shown that height and weight do have a correlation with vessel diameter\[28,29\]. Prince et al. reported that the mean diameter of the vena cava was 20 ± 3 mm. Similar results were found for the F05 vena cava with a mean diameter 19.1 ± 4 mm. However, the diametric measurements from Prince et al. were only taken at one location in the vena cava and do not reflect the variance in diameter at discrete locations. It should also be noted that fluid intake can have an effect on the vessel diameter. In 2008, Kosiak et al. found that vena cava diameter was 20.1 ± 2.3 mm. However, depending on the level of fluid intake, they showed that this measurement can fluctuate approximately 8% \[30\].

Kosiak et al. also reported the average diameter of the aorta post fluid intake to be 16.01 ± 1.12 mm. These measurements were taken 5 – 10 mm superior to the celiac trunk. Similarly, Hager et al. reported the aortic diameter at the isthmus to be 24.7 ± 4 mm\[31\]. In comparison, the average diameter of the F05 aorta at the inferior portion of the aortic arch was 16.9 ± 1.5 mm. The diameter of the ascending aorta was found by Mao et al. to be 31.1 ± 3.9 mm\[32\]. Comparatively, the mean F05 ascending aorta diameter was 24.2 ± 3.2 mm.

There are numerous applications for evaluation organ volumes of specific anthropometrics. For example, the data presented in this study can be used as a benchmark for the development of future finite element models with explicit abdominal representations. This includes both subject specific models and also models derived by from scaling. In both cases, it is important that the proper inertial response of the abdomen be represented. The data can also be used in the future to model and determine
potential disease states affecting the small female. Beyond modeling, clinicians could use datasets of this kind as a means to establish normalcy for a specific population.

In the future, datasets such as this can be extended to include a number of other important anthropometries. For example, the 95th percentile male is often used in ATDs for motor vehicle impact tests and as a target for finite element model scaling. These population specific representations will be critical to most accurately simulating the inertial response of the abdomen in dynamic impact environments.

3.5 CONCLUSION

In order for computational models to accurately predict injury, they must be developed to include geometries that are anatomically based on both gender and body size. In total, 44 organs have been segmented and 60 vessel diameter measurements of representative 5th percentile female subjects have been collected. The results were compared to published data of the normal distribution of female organ volumes. This comparison found that the 5th percentile female in terms of anthropometry does not necessarily have organs volumes that are within the 5th percentile. This dataset will provide a useful comparison for both current and future full body models of the small female.
3.6 REFERENCES


### 3.7 Appendix

Table 7- Volume and Diameter Data for Each Subject

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>Liver (ml)</th>
<th>Spleen (ml)</th>
<th>L Kidney (ml)</th>
<th>R Kidney (ml)</th>
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<th>VC Inferior (mm)</th>
<th>VC L3 (mm)</th>
<th>Aorta Superior (mm)</th>
<th>Aorta Inferior (mm)</th>
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Chapter IV: A Technique for Developing CAD Geometry of Long Bones Using Clinical CT Data

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ABSTRACT

This study presents a method to generate computer aided design (CAD) representations of long bones from clinically based computed tomography (CT) scans of a 5th percentile female (24 yr, 48.1 kg, 149.9 cm), a 50th percentile male (26 yr, 79.4 kg, 174.9 cm), and a 95th percentile male (26 yr, 102.5 kg, 189.5 cm). A secondary aim is to characterize the cortical thickness along the main axis of these bones.

Literature studies have shown that cortical thickness estimations from CT are not accurate when the thickness of the cortical bone falls below 1.5 times the full width at half max (FWHM) of the scanner point spread function (PSF). For the scanner used in the current study, the cut-off was determined to be 2.75 mm. Starting with standard clinical CT images, six long bones along the right side of the body from each subject (the humerus, H, ulna, U, radius, R, femur, F, tibia, T, and fibula, Fi) were automatically segmented using predefined thresholds for compact bone and converted to CAD models. A semiautomatic script was written to calculate the characteristic cortical thickness, Tc, at one mm increments along the bone axis. For all bones, the measurements were made proximal to distal and the maximum Tc was greater than the scanner cut-off. In the epiphyses where the value of Tc fell below the cut-off, the endosteal bone layer from the segmentation was replaced with literature values projected inward from the periosteal surface. G1 continuous transition regions were implemented to reduce discontinuity between (projected) epiphyseal and (segmented) diaphyseal bone and averaged 10.9% of the bone’s length. On average, 74.7 ± 7.4% of the bone models geometry along the main shaft was greater than the scanner cut-off and was therefore derived from the CT scan.
data. The maximum $T_c$ and its relative distance along the shaft were also determined for each bone. The reported methods provide a means to leverage existing clinical CT data. Also, cortical thickness measurements are traditionally taken at discrete points along the shaft of the bone. The use of this method will allow for the characterization of the average thickness along the length of the bone. It may also aid in the development of orthopaedic implant design and subject specific finite element models (FEMs) to predict fracture and ultimately prevent injury.

**Keywords:** Biomechanics, full width half max, subject specific bone model, G1 continuity, CAD, finite element model
4.1 INTRODUCTION

Numerical models of bone have become an important part of orthopedic biomechanics in recent years [1]. Such models have been used in the development and evaluation of orthopedic implants [2,3], surgical guidance [4], and subject specific finite element models (FEMs) for biomechanics research [5]. In particular, FEMs combined with greater computing power are an integral part to understanding injuries to the human body [6-8]. To achieve useful results from FEMs, it is necessary to have precise geometries, along with accurate material definitions and boundary conditions. Cortical thickness is a geometrical aspect that is particularly important because of how it affects bone stiffness and, if fracture is predictable in a given model, the timing, location, and pattern of fracture [8-11]. An overestimation of cortical thickness will cause an overestimation of stiffness and result in an unrealistic response of the model to loading [11,12].

Many research studies have used CT to determine the cortical thickness of bones [5,13-20], primarily for the purpose of developing computer aided design (CAD) representations and FEMs of these structures [5,17,19,21]. However, it has been shown in both computational [22] and experimental studies [13-15] that the limit thickness where the compact bone be accurately identified falls between 1.5 and two times the full width at half maximum (FWHM) of the point spread function (PSF) of the scanner. The PSF is essentially the degree to which an imaging system blurs a point input. Three factors affect shape of the PSF: blurring due to geometric effects (size and shape of focal spot, x ray scattering, etc.), partial volume effects, and the reconstruction kernel [22]. The epiphyseal regions of long bones are particularly vulnerable to cortical thickness
overestimation through segmentation due to the thin layer of cortical bone found in this region [19].

The aim of this study is to present and apply a method to generate accurate CAD representations of selected long bones. For this study, CAD will be developed from 3 individuals representing a range of target anthropometries, a 5\textsuperscript{th} percentile female (F05), a 50\textsuperscript{th} percentile male (M50), and a 95\textsuperscript{th} percentile male (M95). These 3 individuals were selected based on height, weight, and 15 anthropomorphic measurements. The sizes were selected to correspond to the three main driving adult driving populations evaluated by the National Highway Traffic Safety Administration (NHTSA) using anthropomorphic test devices (ATDs) and full body FEMs. In each subject, the following six long bones were obtained from clinically based CT scans of each participant: the humerus, ulna, radius, femur, tibia, and fibula. A secondary aim of this study is to characterize the characteristic cortical thickness, $T_c$, along the mid-shaft of these bones where CT data is reliable. Comparisons to the literature are made in order to validate the findings in specific regions. Related efforts have already been reported to model cortical thickness in bones with more complicated geometries like the ribs and pelvis [5,23,24].

Long bones were selected for this study because of their single axis, for their important role in mobility, and because they are relevant to researchers with interests ranging from blunt injury to orthopedic implant design. Not all long bones were analyzed for this study due to scanner limitations. For example, long bones such as the metatarsals and metacarpals of the feet and the phalanges could not accurately be reconstructed due to the thinness of their cortical layers. In these instances, surveys of published literature studies tended to report cortical thickness values at a specific location.
on a bone. However, in the case of most bones, particularly in long bones, the thickness of the cortical bone varies substantially throughout the structure, allowing for increased cortical thickness in areas where larger biomechanical loads are placed. Because of this, the current study focused on a method to generate thickness estimates along the entire main axis of each selected long bone.

There are numerous examples of applications of bone models. One example of how these models are already in use is the development of subject specific orthopedic implants for knee and hip replacement surgeries [2]. These models are also used in the evaluation of implant lifespan and the effects of implant loading on surrounding bone tissue [3]. Subject specific bone models are often used to determine injury tolerance [5,25] and even have the potential for modeling disease states such as osteoporosis [26].

Long bones of the upper and lower extremities are a common site of injury in motor vehicle crashes [27,28]. It has been shown that lower extremities are the most common body region to have an AIS 2+ injury [29] and that over half of all lower extremity injuries involved with frontal crashes are to the knee-thigh-hip complex [30]. Also, a query of the National Automotive Sampling System (NASS) database for frontal belted crashes from model year 1998 to the present showed that long bone fractures account for nearly 40% of the most common injuries (defined by Abbreviated Injury Score (AIS) codes). When observing injuries on the AIS scale, long bones are found in three of the top five most common injuries (femur, tibia, and radius) and six of the top 13 most commonly reported injuries (femur, tibia, radius, ulna, humerus, and fibula). As computers become more powerful and the predictive capabilities of FEMs increase, accurate representations of cortical thickness will be essential to increasing the accuracy
of computational models. The data collected using the presented technique for measuring cortical bone thickness will be valuable for assigning cortical thicknesses to bones used in blunt impact simulations. By applying the resulting cortical thickness map, shell offsets can be made to the elements of the representative bone within FEMs to characterize cortical thickness. By examining the $T_c$ along the length of the shaft, models can be developed to include regions of the bone that are biomechanically relevant to bone loading.
4.2 METHODS

CT images of a 5th percentile female (24 yr, 48.1 kg, 149.9 cm), a 50th percentile male (26 yr, 79.4 kg, 174.9 cm), and a 95th percentile male (26 yr, 102.5 kg, 189.5 cm) subject were used for this study [31]. The individuals were scanned as part of a larger effort in developing global standard finite element models of the human body. The scanning protocol was approved by the Wake Forest University IRB (#57065). Numerous external anthropometric target values were matched between the subjects and their target representations from the literature [31-33]. No abnormalities were found upon review of the radiology. CT scans were acquired using a 16-slice GE LightSpeed Pro scanner (GE Healthcare, Waukesha, WI). The field of view (FOV) for the upper extremity scans was 50 cm and the matrix size was 512 x 512 pixels for a pixel edge length of 0.98 mm. The FOV for the lower extremity scans was 40 cm, with the same matrix size, resulting in a pixel edge length of 0.78 mm. The slice thickness of all scans was 0.63 mm. A large FOV was necessary in most cases because of the size of the subjects and the desire to minimize total number of scans taken.

Because of its ability to produce a visible contrast between cancellous and cortical bone, CT is the primary modality for determining cortical thicknesses. Cortical bone was segmented from CT images using a commercial image processing software package (Mimics v. 15.0, Materialise, Leuven, Belgium). Periosteal and endosteal surfaces were identified using automatic thresholding to identify adult compact bone (Figure 12-A). Manual segmenting was used to separate bones from proximal structures in joints and in regions too thin for thresholding to capture. The polygonal surfaces obtained from the initial segmentations were then conditioned and used to create Non-Uniform Rational B-
Spline (NURBS) surfaces (Figure 12-B) using Studio software (v12, Geomagic, Rock Hill, SC). Each bone was modeled with an outer layer, representing the periosteal surface of the bone, and an inner layer, representing the endosteal surface. The interior and exterior NURBS surfaces were then used in the analysis of cortical thickness. This process was used to create CAD representations of the long bones considered in this study: namely the femur, tibia, fibula, humerus, radius and ulna.

During image collection, a portion of the proximal ulna of the 50\textsuperscript{th} percentile subject was truncated. In order to compensate for this, data from a post-mortem human subject (PMHS) meeting the same height and weight targets (male, 47 y/o, 178 cm, and 79 kg) was used for this bone only. The PMHS was scanned on the same CT scanner as noted above. A three dimensional registration of the radius of both the PMHS and the volunteer was conducted using Studio to obtain scaling data. The uniform scale factor...
from the subject to the PMHS was determined to be 92.9%. This uniform scaling was applied to the periosteal and endosteal surface segmentations of the PMHS ulna. The CAD model of the ulna used in this study was constructed from these scaled PMHS segmentations.

The cortical thickness in a given plane is considered the average distance from the periosteal to endosteal surface of the bone. To determine an average for thickness that varied along the length of the bone, a visual basic script was written in a CAD software package (Rhinoceros v. 4, McNeel, Seattle, WA). It was designed to determine the $T_c$ at a finite number of planes perpendicular to the long axis (Z axis) of the bone (Figure 13-A). To conduct this analysis, a basis plane position was established at the proximal end of the bone, perpendicular to the long axis. Transverse contour planes were taken at one mm increments along the length of the bone, revealing two closed, concentric curves from the intersection of the cut plane, one representing the circumference of the endosteal layer and the other the circumference of the periosteal layer. (Figure 13-B). The location of the contour line pairs was identified by the distance from the basis plane along the Z-axis.
Figure 13- Process for determining characteristic cortical thickness values. (Exterior bone surface is transparent) (A) Mid-shaft with cut plane perpendicular to Z axis (B) Isometric view of interior and exterior contours of the isolated cross section of bone (C) Top view of isolated cross section of bone showing exemplar outer and inner radius

Due to changes in the relative distance between these contours as evidenced in Figure 13, a simple difference between the outer radius, $R_o$, and the inner radius, $R_i$, for measuring the thickness at a discrete location was not used. Instead, the $T_c$ was calculated at each cross section along the bone axis using the inner and outer circumference per Equation 1, (Figure 13-C). Using this method, $T_c$ values were determined along the main axis of the bone in one mm increments.
\[ \mu_{ct} = \frac{A}{\frac{1}{2}(p_{outer} + p_{inner})} \]

where \( A = \) cross sectional area between the outer and inner contours (mm\(^2\))

\( p_{outer} = \) outer contour perimeter length (mm)

\( p_{inner} = \) inner contour perimeter length (mm)

Dougherty et al. showed that the lowest cortical thickness that can be accurately reconstructed is between 1.5 to 2 times the FWHM of the PSF of the CT scanner used [22]. In the study, a relationship between FWHM and FOV was provided for scans with a FOV of up to 42 cm. Good agreement was found between the FWHM of the scanner used in the present study and the data provided by Dougherty. Based on this data, for an FOV of 50 cm, and assuming an accuracy limit of 1.5 times the FWHM of the scanner, the limit thickness for cortical bone reconstruction was considered 2.75 mm.

The \( T_c \) along the Z axis was calculated for each bone, showing a peak at a location generally near the mid-shaft and a gradual decrease toward the proximal and distal epiphyses in roughly a bell shaped curve. Regions of thickness overestimation were determined using the thickness cut-off value. The location at which \( T_c \) dropped below the scanner cut-off signaled the location on the CAD model where the endosteal surfaces were no longer considered accurate. These surfaces were deleted. Cortical thickness from literature sources were used in the epiphyseal regions by projecting inward from the periosteal surface. A literature survey of cortical thickness values was conducted and is shown in Table 8. In regions where no data was provided, the cortical
thickness was assumed to be 1 mm. Thickness values used in the epiphyseal projections for the 50th percentile male long bones have been previously published as part of a manuscript detailing the CAD data development of a full body finite element model [31]. The endosteal surface between the segmented and projected surfaces was developed to ensure G1 continuous CAD using standard CAD software.
<table>
<thead>
<tr>
<th>Bone</th>
<th>Location / Notes</th>
<th>Value, mm</th>
<th>Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>Humerus</td>
<td>Proximal diaphysis</td>
<td>4.8 ± 0.96</td>
<td>[34]</td>
</tr>
<tr>
<td></td>
<td>Distal diaphysis and epiphysis</td>
<td>0.5 – 3</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Distal mid-shaft</td>
<td>♂: 4, ♀: 3</td>
<td>[35]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft (~ 1/3 from proximal end)</td>
<td>♂: 5.1 ± 0.48</td>
<td>[36]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft (~ 1/3 from distal end)</td>
<td>♂: 6.6 ± 0.75</td>
<td></td>
</tr>
<tr>
<td>Ulna</td>
<td>Proximal mid-shaft</td>
<td>♂: 4-3, ♀: 3-2.7</td>
<td>[35]</td>
</tr>
<tr>
<td></td>
<td>Proximal mid-shaft</td>
<td>♂: 4.5 ± 0.24</td>
<td>[36]</td>
</tr>
<tr>
<td>Radius</td>
<td>45% from distal end (mid-shaft)</td>
<td>2.61 ± 0.49</td>
<td>[37]</td>
</tr>
<tr>
<td></td>
<td>Neck of radius</td>
<td>♂: 2.6 ± 0.13</td>
<td>[36]</td>
</tr>
<tr>
<td></td>
<td>Proximal mid-shaft</td>
<td>♂: 3.9 ± 0.20</td>
<td>[36]</td>
</tr>
<tr>
<td></td>
<td>Lateral border</td>
<td>3.8 ± 0.95</td>
<td>[38]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft</td>
<td>3.1 ± 0.23</td>
<td>[36]</td>
</tr>
<tr>
<td>Femur</td>
<td>Average thickness</td>
<td>♂: 2.9± 2.3</td>
<td>[25]</td>
</tr>
<tr>
<td></td>
<td>Proximal femur, neck</td>
<td>♀: 0.98</td>
<td>[39]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft, right femur</td>
<td>♂: 6.81±0.07</td>
<td>[40]</td>
</tr>
<tr>
<td></td>
<td>Femoral neck, (NHANES)</td>
<td>1.71 – 1.78</td>
<td>[41]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft, right femur</td>
<td>♂:10.3 ± 0.27</td>
<td>[36]</td>
</tr>
<tr>
<td></td>
<td>Lateral cortex</td>
<td>♂: 6.23 ± 1.17</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Calcar femorale (inferior neck)</td>
<td>♂: 7.05 ± 1.94</td>
<td>[42]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft (thickest portion)</td>
<td>♂: 11.11 ± 1.37</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Femoral neck, inferior</td>
<td>♂: 3.5</td>
<td>[43]</td>
</tr>
<tr>
<td></td>
<td>superior</td>
<td>♂: 0.7</td>
<td></td>
</tr>
<tr>
<td></td>
<td>posterior</td>
<td>♂: 0.8</td>
<td></td>
</tr>
<tr>
<td></td>
<td>anterior</td>
<td>♂: 1.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Femoral neck, inferior</td>
<td>♀: 3.6</td>
<td>[44]</td>
</tr>
<tr>
<td></td>
<td>superior</td>
<td>♀: 1.1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>posterior</td>
<td>♀: 1.3</td>
<td></td>
</tr>
<tr>
<td></td>
<td>anterior</td>
<td>♀: 1.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Distal femur, distal to epiphyseal growth plate</td>
<td>♂: &lt; 1</td>
<td>[45]</td>
</tr>
<tr>
<td>Tibia</td>
<td>Mid-shaft</td>
<td>♂: 6.8 ± 0.30</td>
<td>[36]</td>
</tr>
<tr>
<td></td>
<td>Mid-shaft</td>
<td>♀: 6.1 ± 0.6</td>
<td>[46]</td>
</tr>
<tr>
<td>Fibula</td>
<td>Mid-shaft</td>
<td>♂: 3.7 ± 0.16</td>
<td>[36]</td>
</tr>
</tbody>
</table>
In each case, a threshold was placed demarcating the regions of the bone where
cortical thickness measurements initially fell below 2.75 mm (the scanner cut-off).
However, in certain cases, there are regions below the scanner cut-off with structures of
particular importance in biomechanics with thicknesses greater than 2.75 mm. For
example, on the proximal portion of the femur, the threshold where the cortical thickness
falls below 2.75 mm for the 95\textsuperscript{th} percentile male is 39 mm from the femoral head.
However, on the bottom of the femoral neck and the lesser trochanter, areas in the region
below the threshold of the scanner cut-off, the cortical thickness is above the threshold.
These areas are biomechanically important due to their contributions to sustaining load
and injury mechanisms. Therefore, it is important to ensure that these regions are
captured in order to accurately model the bone. This was accomplished using deviation
maps in Studio to assess the continuous thickness of the bones. Then, using morphing
techniques in Studio, specific regions of the bones could be offset while maintaining the
integrity of regions necessary to accurately model the biomechanical response.

Algorithms within Studio, tangential splines, and conditioning were used to join
the segmented and prescribed regions while maintaining G\textsubscript{1} (tangential) continuity. G\textsubscript{1}
continuity is maintained when the direction of the tangent vectors between the surface
curves are parallel. This prevented sharp edges in the bone surface, which is consistent
with the true morphology of the bone and is an important consideration for the accuracy
of the bone thickness throughout the structure and for future use in computational
modeling. The result is the projected epiphyseal region and diaphysis surface from the
scan data connected by a transition region. This is shown schematically in Figure 14.
Data from the scanner (Z Scan, green) was used for each bone to the point along the T\textsubscript{c}
curve that fell below the transition value (gray plane). The epiphyseal region of the bone was prescribed through projecting the periosteal surface inward a set distance according to literature values (Z Projection, pink). The transition region (Z Transition, blue) was defined as the region of the long bone geometry bounded by the scan and projection regions.

![Diagram of bone transition regions](image)

**Figure 14-** Evaluation of bone transition regions about the Z axis. Z Projection, Z Transition, and Z Scan are representative of the regions of bone captured by the processes used in this study. The inner and outer radius at a representative location is shown. Radial locations used to evaluate the transition distance are also shown.

Two qualities of the Z transition region were assessed: the G1 continuity condition over the region and the Z axis length of the region. G1 continuity was verified using algorithms within Studio for each long bone region to ensure the quality of the transition regions. To quantify the Z transition length, the radial thicknesses of the bones were evaluated in three planes: $0^\circ$, $120^\circ$, and $240^\circ$ extending radially from the central axis (Figure 14). The planes were developed by placing a point at the centroid of each contour of the endosteal surface. The planes were then extended radially and intersection
points were found along the contours of both the periosteal and endosteal surfaces. The outer and inner radii were then found for each plane. These values were then used to establish an average thickness in the proximal and distal portions of the bone. The transition length was calculated as the distance along the Z axis from the scanner cut-off to the average of the $R_o-R_i$ local minimum along the contours in the T-Z plane. The length of transition of each region was then divided by the total Z length of each bone to determine a percent transition length.
4.3 RESULTS

The mathematical NURBS surfaces with associative patchwork were developed to improve the surface topology from the initial segmentation. The final CAD models were composed of G1-continuous, concentric NURBS surfaces representing the periosteal and endosteal cortical layers. In order to check the final 3D CAD representations, both the periosteal and endosteal surfaces of each bone were imported back into the image space and compared to the original scan data. For each bone, the regions in which the $T_c$ along the shaft was found to be greater than the scanner cut-off value can be seen in Figure 15.
Figure 15- Characteristic cortical thickness values of long bones in regions above the scanner cut-off of 2.75 mm. Data for the male and female cortical thickness ranges were taken from Virtama, et al.
The length of each bone in the study and other data regarding the maximum cortical thickness and transition lengths are presented in Table 9. On average, $74.7 \pm 7.4\%$ of the original CAD data on the long axis was retained based on the $T_c$ analysis. The maximum cortical thickness with its corresponding location along the shaft of each bone can be seen in Table 9. The resulting cortical thickness compares favorably to the literature in both diaphyseal and epiphyseal regions (Table 8). The location of the transition region was determined at the location where the difference between $R_o$ and $R_i$ reached a local minimum with respect to the scanner cut-off. The average transition lengths were 5.4% and 5.5% of total bone length for the proximal and distal ends respectively. The total average transition length when factoring in both distal and proximal transition regions was 10.9%. 

<table>
<thead>
<tr>
<th>Percentile</th>
<th>Bone</th>
<th>Bone Z Axis Length (mm)</th>
<th>Distance along bone (prox. to distal) with greatest measured ( \mu_{tc} )</th>
<th>Maximum Measured ( \mu_{tc} ) (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>5th</td>
<td>Humerus</td>
<td>296</td>
<td>58.5%</td>
<td>4.9</td>
</tr>
<tr>
<td></td>
<td>Ulna</td>
<td>217</td>
<td>55.3%</td>
<td>3.7</td>
</tr>
<tr>
<td></td>
<td>Radius</td>
<td>211</td>
<td>45.0%</td>
<td>3.5</td>
</tr>
<tr>
<td></td>
<td>Femur</td>
<td>410</td>
<td>39.8%</td>
<td>8.4</td>
</tr>
<tr>
<td></td>
<td>Tibia</td>
<td>333</td>
<td>62.2%</td>
<td>5.5</td>
</tr>
<tr>
<td></td>
<td>Fibula</td>
<td>312</td>
<td>66.4%</td>
<td>4.0</td>
</tr>
<tr>
<td>50th</td>
<td>Humerus</td>
<td>338</td>
<td>74.1%</td>
<td>6.7</td>
</tr>
<tr>
<td></td>
<td>Ulna</td>
<td>277</td>
<td>57.4%</td>
<td>5.1</td>
</tr>
<tr>
<td></td>
<td>Radius</td>
<td>260</td>
<td>46.4%</td>
<td>4.7</td>
</tr>
<tr>
<td></td>
<td>Femur</td>
<td>465</td>
<td>35.4%</td>
<td>7.5</td>
</tr>
<tr>
<td></td>
<td>Tibia</td>
<td>408</td>
<td>59.1%</td>
<td>7.5</td>
</tr>
<tr>
<td></td>
<td>Fibula</td>
<td>401</td>
<td>55.5%</td>
<td>4.3</td>
</tr>
<tr>
<td>95th</td>
<td>Humerus</td>
<td>364</td>
<td>20.3%</td>
<td>6.0</td>
</tr>
<tr>
<td></td>
<td>Ulna</td>
<td>297</td>
<td>17.2%</td>
<td>5.5</td>
</tr>
<tr>
<td></td>
<td>Radius</td>
<td>275</td>
<td>51.6%</td>
<td>4.6</td>
</tr>
<tr>
<td></td>
<td>Femur</td>
<td>529</td>
<td>29.9%</td>
<td>9.0</td>
</tr>
<tr>
<td></td>
<td>Tibia</td>
<td>439</td>
<td>52.8%</td>
<td>7.4</td>
</tr>
<tr>
<td></td>
<td>Fibula</td>
<td>424</td>
<td>39.2%</td>
<td>3.8</td>
</tr>
</tbody>
</table>
4.4 DISCUSSION

Long bones were chosen for this study due to their geometric qualities and their relative importance in biomechanical studies. These bones have a similar geometry with a single main axis enabling cut planes to be readily acquired normal to the main axis. There was also a clear demarcation of when $T_c$ values were no longer valid. As applied, the technique would not be suitable on other bones in the body for a number of reasons. For example, many bones in the body, including those of great interest in biomechanics like the ribs, have cortical thickness throughout the structure that fall below the scanner cut-off that was utilized in this work. Other methods in literature have instead taken approaches such as utilizing denuded cadaveric samples or micro CT scans that can produce resolutions that are not feasible for living subjects, or simply using literature data to set constant thicknesses [8,23] [5].

Figure 15 provides a detailed example of the trend found in all of the examined long bones. Moving from the proximal end of the bone toward the distal end, $T_c$ rises to a maximum at an average of 49.1% of the length of the bone and declines back to the scanner cut-off level. Nearly all of the diaphyseal bone was captured above 1.5 times the FWHM cut-off. According to Wolff’s law, this thickening corresponds with the location where the bone sustains the greatest moment from loading [47]. To the extent possible, literature data was leveraged for developing epiphyseal regions. However, this data is only available in a large supply for the M50, with very limited literature on the epiphyseal cortical thickness for the F05 and M95. In regions where there was no literature data, the periosteal surface was projected 1 mm.
Characteristic mid-shaft cortical thickness values were compared to data obtained from work by Virtama and Helela. This study was referenced several reasons. The study utilized a large number of samples taken from direct measurement of cadaveric bone (no imaging used) for all the bones evaluated. Two measurements were taken along the anterior-posterior (A-P) axis of the bone, with the inner and outer diameters subtracted to determine a combined cortical thickness (CCT). For the purposes of comparison to the present study, the combined cortical thickness was divided by 2 since CCT is, in effect, two lengths of cortical thickness. The CCT divided by 2 can be thought of as determining cortical thickness by averaging samples per cross section. While these values are useful for comparison of a 50th percentile male and female, they do no specifically address the cortical thickness of the 5th percentile female and 95th percentile male. Therefore, these subjects can be compared to the characteristic cortical thicknesses found in the study, but nothing can be inferred from the standard deviations found by Virtama with relation to the F05 and M95.

Another point to note is that a cross section of a long bone (illustrated in Figure 13-C) can contain large circumferential variation in the cortical bone thickness when taken on different planes. Therefore, slight changes in the 2-point measurement location can have a large effect on the average thickness value. For that reason, the method used in the present study is more robust for calculating the average of the bone’s thickness. Note that the perimeter and area based measurement method produced a $T_c$ of the tibia in good agreement with the previous study. In all, good agreement was found between cortical thicknesses and literature values. The advantage of the present approach is that
the thickness values are determined along the shaft’s length (Figure 15). This would provide useful data that could be applied to future modeling efforts of key populations.

The bone models of the 50th percentile male presented in this work have been used along with a complete human body CAD data set [31] in the development of a full human body FEM with the purpose of providing injury prediction in automobile accidents [8,31,48-51]. Also, the bone models of the 5th percentile female are being implemented in a full body FEM of the small female that is currently being developed. The subjects were matched to a selection of 15 anthropomorphic targets for their relative sizes, were deemed normal by a collaborating radiologist and were in generally excellent health [32]. This is an advantage because models are frequently developed from data where specimen age, pathology, and physical parameters are not ideal.

The techniques in this paper can also be extended to other medical and biomechanical applications. For example, data of cortical thickness along the shaft of long bones would be useful for regenerative medicine studies attempting to facilitate both bone repair and tissue-engineered bones. Computational bone modeling is also used in a variety of applications including the development of patient specific orthopedic implants and patient specific finite element models. Though the use of computational models for biomechanical assessment of orthopedic implants was first introduced in 1972 [52], the biofidelity of such models has been greatly enhanced with introduction of accurate computational geometries from computed tomography. [53] The methods in this paper provide a means to address the limitation of geometry reconstruction. In addition, due to the large volume of clinical CT scan data available to researchers at academic medical
centers, applications of these methods are readily expandable to examine larger populations.
4.5 Conclusion

Accurate geometrical reconstruction of cortical bone is critical for biomechanical modelers and designers since bones are the primary load path for external forces. The thickness of cortical bone has a direct result on the predicted timing and location of fracture and forces required to fracture bone. Such prediction is essential for determining the severity of injury, the effectiveness of countermeasures, and even the kinematics of the body during a dynamic event.

A method for developing CAD data for long bones based on clinical CT images has been presented and literature data on bone thicknesses were reviewed. The $T_c$ in 1 mm increments along the main axis of 6 long bones from 3 subjects are provided in the appendix. A large volume of CT data is typically available in patient databases, so the application of this approach increases the potential for its use in model development. The subjects used in this study were recruited for a related study and closely matched the anthropometry of a 5th percentile female, 50th percentile male, and a 95th percentile male respectively. This study has shown that supplemental data from literature can be used to augment the data that is available in clinical scans. The framework presented is flexible enough to improve accuracy further through implementation of more specific literature data.
DECLARATIONS

Funding: Funding for this work was provided by the Global Human Body Models Consortium.

Competing interests: None declared.

Ethical Approval: Approval for the collection of the data used in this study was granted by the Institutional Review Board of Wake Forest University School of Medicine, IRB #5705

Acknowledgments

The authors would like to thank Dustin Crouch for his work developing the area calculation script, Brad Thompson for his pilot work on the M50, and Dan Moreno for his overall assistance related to the GHBMC project.
4.6 References


[52] Prendergast P. J., Finite element models in tissue mechanics and orthopaedic implant design.

Chapter V: Conclusion

The presented work details the initial development phase of the Global Human Body Model Consortium’s (GHBMC) small female finite element model. The focus of this phase of the work was on initial subject recruitment, medical image acquisition, model assembly, and geometrical validation. As part of the effort to validate the model geometries, specific datasets were developed to characterize the small female, including thoracoabdominal organ volumes germane to crash induced injuries and cortical bone thickness. These datasets were acquired to fill the need for data specifically representative of 5th percentile females. To date, computer aided designs (CAD) have been assembled for all bony structures and organs that will be explicitly represented in the subsequent finite element model. Model assembly utilized the comprehensive multi-modal medical image and external anthropometry dataset to accurately represent a vehicle occupant.

While the outlined data and techniques have focused on the assembly and validation of anatomical structures germane to the modeling of CIIs (bony structures and organs), other structures designed to facilitate passive load transfer and promote accurate kinematics will also be modeled. Based on previous work for the development of the GHBMC average male dataset, roughly 96 muscles and 26 ligaments, tendons, and other cartilaginous tissues are targeted for inclusion in the final model. With regards to model development, cartilaginous tissue will be created during the meshing phase where possible, reducing the need for CAD representations. Upon completion of geometry development, the model will be meshed and used for a series of regional and global validation tests. Ultimately, the geometries developed for this model will be used for full
body simulations in dynamic crash environments. The goal of these simulations will be to provide researchers and engineers with a tool to increase motor vehicle safety.
Chapter VI: Scholastic Vita

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EDUCATION

Virginia Tech – Wake Forest University, Winston-Salem, NC
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Thesis: Development of a CAD Dataset of the 5th Percentile Female
Advisors: Dr. F. Scott Gayzik (Primary Advisor), Dr. Joel D. Stitzel,
Dr. Kerry Danelson
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B.S. Biomedical Engineering May 2012

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RESEARCH OBJECTIVES

- To use computational modeling for injury biomechanics research in reducing the
  social and economic costs of unintentional injury
- To complete a Ph.D. in Biomedical Engineering and continue on to a research
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HIGHLIGHTED SKILLS

- Computational biomechanics research: Dynamic nonlinear finite element analysis,
analytical programming, and human body model development and validation
- Experimental biomechanics research: Tissue and organ-level quasi-static testing,
  compression of tissue simulants, and data analysis
JOURNAL PUBLICATIONS


2. Davis ML, Stitzel JD, Gayzik FS. “Characterization of Abdominal Organ Volume and Large Vasculature Diameter for the 5\(^{th}\) Percentile Female.” *Submitted to Traffic Injury Prevention, May 2014*.

3. Davis ML, Vavalle NA, Stitzel JD, Gayzik, FS. “A technique for developing CAD geometry of long bones using clinical CT data.” *In preparation, May 2014*.

4. Vavalle NA, Davis ML, Stitzel JD, Gayzik FS. “Quantitative Validation of a Full Human Body Finite Element Model in Dynamic Pendulum Impacts.” *In preparation, May 2014*


PEER-REVIEWED CONFERENCE PAPERS


SELECTED CONFERENCE ABSTRACTS AND SCIENTIFIC EXHIBITS

1. Davis ML, Stitzel JD, Gayzik FS. “A Multi-Modality Image Set for the Development of a 5\(^{th}\) Percentile Female Finite Element Model.” Accepted for podium presentation at the International Research Council on Biomechanics of Injury’s Annual Meeting, Berlin, Germany, September 2014.


3. Davis, ML, Hayes AR, Gayzik FS, Moreno DP, Stitzel JD. “The Development of Volumetric Organs from a Multi-Modality Image Dataset for Use in a Small Female Full-Body Finite Element Model.” Poster at the VT-WFU School of Biomedical Engineering and Sciences Symposium, May 2013


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1. LS-DYNA Training, Troy, MI, August 2012
2. LS-DYNA Contacts Training, Winston-Salem, NC, January 2014
3. HyperMorph Training, Winston-Salem, NC, January 2014
4. HyperWorks Training, Winston-Salem, NC, September 2014

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**Computational Modeling**
2. Development of the Global Human Body Model Consortium’s 5th percentile female CAD dataset

**Orthopaedic Biomechanics**
1. Compared the biomechanical effects of fibular allograft to calcium phosphate fixation for split-depression tibial plateau fractures
2. Evaluated the effects of varying levels of vibration amplitude on the structural and mechanical properties of ligaments and tendons
3. Assessed the ability of low-magnitude-high-frequency vibration to accelerate healing of MCL tears
4. Evaluated the ability of desferrioxamine to accelerate healing of non-union bone fractures

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1. Evaluated the effects of varying levels of vibration amplitude on the structural and mechanical properties of ligaments and tendons
2. Assessed the ability of low-magnitude-high-frequency vibration to accelerate healing of MCL tears
3. Evaluated the ability of desferrioxamine to accelerate healing of non-union bone fractures

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Conference Chair’s Award

*University of North Carolina at Chapel Hill Undergraduate Awards* 2010 - 2012
Dean’s List
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UNC Biomechanics of Human Movement Symposium

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FEA Pre/Post Processing Software: LS-PrePost, Hyperworks, Oasys
Image Analysis Software: Mimics, Image J, Amira
CAD: Rhinoceros, Geomagic Studio, Solidworks

**PROFESSIONAL MEMBERSHIPS**

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Treasurer, VT – Wake Forest Chapter 2013 – Present
REFERENCES

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