Development and Full Body Validation of a 5th Percentile Female Finite Element Model

BY

Matthew Logan Davis

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Approved by:

F. Scott Gayzik, PhD, Advisor, Chair

Joel D. Stitzel, PhD

Andrew R. Kemper, PhD

Kerry A. Danelson, PhD

Ashley A. Weaver, PhD
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Abstract

To mitigate the societal impact of motor vehicle crash, researchers are using a variety of tools, including human body finite element models (FEMs). Such models are often developed to represent a 50th percentile male occupant (M50). However, in order to address the effects of size and sex-related geometrical changes, there is interest in developing models of other driving cohorts. As part of the Global Human Body Models Consortium (GHBMC) project, comprehensive medical image and anthropometrical data of the 5th percentile female (F05) were acquired for the explicit purpose of FEM development. This multi-modality dataset was leveraged for the development of a posture specific CAD model of the F05.

The CAD dataset was then utilized for mesh development of the GHBMC F05 occupant (F05-O) model. Structured hexahedral mesh was used in the majority of the body with the exception of abdomen. The number of contacts implemented in the model was minimized via node-to-node connections and element property assignment based on the underlying CAD data. Ultimately, the F05-O model consisted of 981 parts, 2.6 million elements, 1.4 million nodes, and had a mass of 51.1 kg.

As there is relatively little biomechanical data specific to anthropometries beyond the average male, small female model responses often require numerical transformation prior to validation. Therefore, a study was conducted utilizing volumetrically scaled human body models to quantitatively compare scaling techniques. This study found that no single technique was ideal for all impact scenarios. However, scaling based on a ratio of effective masses was found to be the most proficient at scaling a reference response to the target. An additional study utilizing human body models was also performed to
evaluate the CORA and ISO/TS 18571 objective evaluation techniques. Significant differences were found for the interpretation of these methods and a survey of subject matter experts suggests using the magnitude method in CORA and the ISO shape and phase methods may be the most intuitive approach to reporting objective ratings.

For full body validation, the model was simulated in 10 validation cases ranging from hub impacts to full body sled cases. In order to make comparisons to experimental data, which represent the mass of an average male, the model was compared to experimental corridors using two methods: 1) post-hoc scaling the outputs from the baseline F05-O model using the technique applied to the experimental dataset and 2) geometrically morphing the model to the body habitus of the average male to allow direct comparisons. This second step required running the morphed full body model in all 10 simulations for a total of 20 full body simulations. Overall, geometrically morphing the model was found to more closely match the target data with an average ISO score for the rigid impacts of 0.76 compared to 0.67 for the scaled responses. Based on these data, the morphed model was then used for model validation in the vehicle sled cases and attained an average weighted score of 0.69 for the two sled impacts.

The F05-O model was found to be robust and showed fair to good agreement with experimental biomechanical response data as per the results of ISO objective rating metrics. Based on the findings of this study, it was quantitatively demonstrated that full body morphing can be a more effective means model validation than post-hoc data scaling. As human body modeling is extended to address ever more challenging aspects of injury biomechanics, the GHBMC F05-O model is poised to provide needed insight to the biomechanics of small female occupants.
Chapter I: Introduction and Background

I. ROAD TRAFFIC INJURIES AND HUMAN BODY MODELING

Motor vehicle collision injury prevention remains a leading public health concern worldwide. In 2013, the World Health Organization (WHO) reported more than 1.2 million deaths and another 20-50 million non-fatal injuries as a result of motor vehicle accidents (W.H.O 2013). According to the WHO, road traffic injuries are the 8th leading cause of death globally and are predicted to become the 5th leading cause of worldwide death by 2030. The injury outcomes of vehicular crash also result in significant financial costs. In the United States alone, the National Highway Traffic Safety Administration (NHTSA) estimates the economic and societal costs of vehicle crash at $871 billion (Blincoe et al. 2014).

To mitigate the toll of these injuries, researchers are using a variety of tools to design and evaluate vehicle safety devices. One of the emerging tools for this purpose is computational human body modeling. The application and development of such models is a growing component of injury biomechanics. The last 20 years has seen a large increase in the number of human body models (HBMs) being developed at both the full body level (Hayes et al. 2014; Toyota 2010; Yang et al. 2006) and body regional level (DeWit and Cronin 2012; Li et al. 2010; Shin et al. 2012; Soni and Beillas 2015). Computational HBMs, such as finite element models (FEM), are appealing because they offer a cost-effective method to evaluate and design vehicle safety devices. They are also useful for providing deeper insight into the injury mechanisms of specific tissues during dynamic loading scenarios.
I. MOTIVATION FOR SMALL FEMALE MODELS

Traditionally, HBMs have been 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. A 2013 study by Sivak et al. observing gender trends in motorists found that there are now more licensed female drivers than male drivers in the United States (Sivak 2013). Thus, it is important to represent the changing demographic of road users when evaluating vehicle safety devices. For example, Summers et al. described a series of air bag related fatalities and serious injuries that raised concerns about the effectiveness of air bags for protecting occupants beyond the average male. In these air bag related fatalities, close proximity to the wheel was identified as a factor in all cases, with a secondary trend related to stature, gender, and age. Of the drivers who sustained fatal injuries in minor to moderate severity crashes with a deploying air bag, 78% were female. Within the group of fatally injured females, 82% were below average height (Summers et al. 2001). While driver stature has been reported as the most dominant effect on seated proximity to the steering wheel, small women are believed to be the largest anthropometrical population that sit closer to the steering wheel. This proximity to the steering wheel, combined with differences in body size, may lead to more severe injuries (Manary et al. 1998; Summers et al. 2001). From a regulatory perspective, this has been addressed by including a small female anthropomorphic test device (Hybrid III 5th percentile female) in certain regulatory crash modes to evaluate the ability of vehicle safety devices to protect a wider range of occupants.
I. SMALL FEMALE INJURY RISK

Previous research has shown that females are considered to be at a higher risk of sustaining injury during automotive accidents when compared to males (Evans 2001; Morris et al. 1998). As such, small females are generally considered to be one of the most vulnerable driving populations, especially in loading conditions such as out of position air bag loading (Duma et al. 1998; Jingwen et al. 2012). Therefore, researchers often apply models of these occupants as a means of conservatively estimating the performance of safety devices (Duma et al. 1998). Research in the literature suggests that females not only have a higher risk of injury to specific body regions, but have a higher risk of sustaining fatal or disabling injuries as a result of vehicle crash (Ulfarsson and Mannering 2004). With regards to specific body regions, small females have been reported to be more likely to sustain chest injuries with a maximum AIS of 2 or greater (Welsh and Lenard). Temming et al. also found that females are two times more likely to suffer neck distortion injuries compared to males (Temming and Zobel 1998). To reduce confounding effects of crash severity, Evans et al. reported that females are at a greater risk of injury compared to males in similar physical impacts (Evans 2001). However, the increased risk of injury also appears related to tolerable ΔV. Using the National Automotive Sample System (NASS) and the Cooperative Crash Injury Study (CCIS), Mackay et al. found that the median tolerable ΔV was considerably lower for females than for males at any given severity level on the Abbreviated Injury Scale (AIS) (Mackay and Hassan 1999). One of the proposed reasons for this increased risk is the preferred seating posture of small females closer to the steering wheel as a result of their stature (King and Yang 1995; Manary et al. 1998; Melvin et al. 1993). Increased injuries may
also be attributable to lower structural strength within females due to lower bone mineral content (Duma et al. 1999).

I. **EXISTING SMALL FEMALE FINITE ELEMENT MODELS**

Until recently, finite element models of the 5th percentile female (F05) have typically been developed by applying scaling techniques to existing 50th percentile male models, since scan data of such a specific target anthropometry are 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 (Serre 2006; Vezin and Verriest 2005). 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 (Happee et al. 2000). This model was developed using anthropometry from the RAMSIS anthropometry database. Kimpara et al. reported on the development of a small female model using geometry from the View Point Datalabs database (Kimpara et al. 2002). However, as this model was designed for computational efficiency, internal organs were not explicitly represented. The model was later updated to include soft tissue structures (Kimpara 2005). In 2003, Iwamoto et al. presented the Toyota Total Human Model for Safety (THUMS) American F05 (AF05) model (Iwamoto et al. 2003). The THUMS AF05 was originally produced by scaling the THUMS average male model to the AF05 using external anthropometry(Schneider et al. 1983). However, in order to account for gender differences, thoracic and pelvic regions of the model were developed ad hoc to represent the small female. Recently, the THUMS AF05 has been updated to include
explicitly represented internal organs with geometries derived from medical image data (Watanabe et al. 2012). Additionally, Klein et al. has developed a set of parametric whole body models to account for age, sex, and BMI. As the focus of this work was on evaluating injuries to the lower extremity, parametric models of the pelvis, femur, and tibia, as well as external body surface shape models were used to create full body morphs using the THUMS average male model as a baseline (Klein 2015).

One of the main limitations for each of the models described above is a lack of posture specific organ shape and placement. This is an important aspect for the development paradigm as recent studies have found significant differences in abdominal organ position and shape between supine and seated postures (Beillas 2009; Hayes et al. 2013). These changes in shape and position can have significant effects on the inertial response of the abdomen to blunt loading. As part of the Global Human Body Models Consortium (GHBMC) project, extensive medical images of a representative 5th percentile female volunteer in multiple modalities and postures were acquired. These medical images were acquired for the explicit purpose of developing 5th percentile female finite element models. A brief description of the GHMBC and their mission can be seen below.

I. GLOBAL HUMAN BODY MODELS CONSORTIUM (GHBMC)

The Global Human Body Models Consortium (GHBMC) is a consortium of automotive manufactures, universities, and government agencies whose goal is to consolidate world-wide research and development activities in human modeling into a single global effort to advance crash safety technology. The mission of the GHBMC is to develop and maintain high fidelity finite element HBMs for automotive crash simulation.
During Phase I of the project, the primary focus of the GHBMC was on the development and validation of a detailed model of the average male occupant (M50-O). As part of Phase II work, the GHBMC extended the family of models to include male and female seated occupants of various sizes and pedestrians including a six year old child (Figure 1).

Figure 1. GHBMC family of models

I. GHBMC SMALL FEMALE MODEL DEVELOPMENT

The development of the GHMBC small female model proceeded in three main stages: 1) development of the medical image database and CAD dataset, 2) development of the finite element mesh, and 3) model validation. The first objective of the project was to develop CAD data from medical images of a representative F05 subject. 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. This is a significant aspect for the development approach in order to capture posture-dependent organ locations. The strengths of each imaging modality were leveraged to develop small female anatomy that can be used to characterize relevant crash-induced injuries (CIIs). Included geometries were developed using a variety of segmentation techniques. The
developed geometries were also compared to literature when available for verification. The second objective was to discretize the CAD dataset for the development of the finite element mesh. Lastly, the model was validated by simulating biomechanical studies conducted on post mortem human subjects (PMHS).

When developing the M50-O model, the GHBMC established body region model (BRM) centers of expertise (COE) for the development and validation of regional finite element models using CAD geometries developed by the full body model (FBM) COE. These regions were then integrated by the FBM COE for full body validation. For development of the F05-O, the CAD and mesh development for each body region was conducted by the FBM COE. BRM COEs were consulted throughout the meshing and model assembly processes and provided valuable insight derived from lessons learned during the development of the M50-O model that became the foundation for F05-O development. In addition, the FBM COE worked closely with each BRM COE throughout the validation process to integrate model enhancements identified by BRM COEs during regional validation.

Similar to the M50-O development, the FBM COE was responsible for F05-O full body validation. However, due to the limited amount of data specific to the F05, the model responses had to be scaled in some cases to make accurate comparisons. As such, common scaling techniques were evaluated to determine the most appropriate approach for model validation (Eppinger 1976; Mertz 1984). After the responses were scaled to the average male corridors, the model was quantitatively evaluated using the ISO/TS 18571 standard objective rating metric. Following validation, the data obtained from this model will be a valuable tool for the development of vehicle safety devices. To date, the
data set used for the development of this model is the first of its kind, acquired with the explicit purpose of developing a full-body finite element model of the F05 for the enhancement of injury prediction. The GHBMC F05-O can be seen in Figure 2.
I. CHAPTER SUMMARIES

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

Height, weight, and gender have an effect on the size and shape of anatomical structures. In addition, organ location and shape have been shown to be posture dependent. This chapter presents the application of a multi-modality medical image dataset for the development of a 5th percentile female CAD model in a seated posture. The geometries developed in this chapter served as the source data for the GHBMC 5th percentile female finite element model.

Chapter III: Thoracoabdominal Organ Volumes for Small Women

In order to accurately simulate inertial properties during blunt impact, it is important to verify the volumes of thoracoabdominal organs in the model. This chapter reports the application of medical image segmentation using a sample of small women to determine the range of thoracoabdominal organ volumes for this population.

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

While computed tomography (CT) scans are a valuable tool for developing computational models of bones, limitations with clinical CT scans prevent the accurate reconstruction of thin cortical bone. This chapter describes a method for generating accurate CAD of long bones using clinical CT scans. The method was further applied to obtain continuous cortical thickness maps along the length of the long bones.
Chapter V: An Evaluation of Mass-Normalization Using 50th and 95th Percentile Human Body Finite Element Models in Frontal Crash

Human body finite element models are ideal tools for exploring the effects of body habitus on the biomechanical response during an accelerative event. This chapter describes the application of average male and large male finite element models to evaluate how body habitus influences injury risk and how mass normalization affects the characteristics of the model response. This work was a precursor to the more focused study of scaling techniques presented in Chapter VI.

Chapter VI: An Objective Evaluation of Mass Scaling Techniques Utilizing Computational Human Body Models

Scaling of human body response data is commonly used to compare subjects with differing morphology. While there are several techniques in the literature, it is unclear which is the most applicable to a wide range of impact types. This chapter presents a quantitative comparison of scaling techniques using human body models. In addition, two alternative techniques were proposed and evaluated.

Chapter VII: Comparison of Objective Rating Techniques vs. Expert Opinion in the Validation of Human Body Surrogates

Objective evaluation methods provide quantitative insight into how well human body models predict a biomechanical response. However, these ostensibly objective techniques can have differences in their algorithms that may lead to discrepancies when interpreting model performance. This chapter evaluates two commonly applied objective evaluation techniques and compares them to a survey of subject matter experts to
determine which technique, if either, compares more consistently with expert interpretation.

**Chapter VIII: Development and Full Body Validation of a 5th Percentile Female Finite Element Model**

This chapter describes the mesh development and full body validation of the GHBMC 5th percentile female occupant finite element model. The model was compared to experimental data in validation cases ranging from localized rigid hub impacts to full body sled cases. In order to make direct comparisons to experimental data, which represent the mass of an average male, the model was compared to experimental corridors using two methods: 1) *post-hoc* scaling the outputs from the baseline F05-O model and 2) geometrically morphing the model to the body habitus of the average male to allow direct comparisons.

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<td>II</td>
<td>A Multi-Modality Image Set for the Development of a 5th Percentile Female Finite Element Model</td>
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<td>III</td>
<td>Thoracoabdominal Organ Volumes for Small Women</td>
<td>Traffic Injury Prevention Medical Engineering and Physics</td>
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<tr>
<td>IV</td>
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<td>Techniques Utilizing Computational Human Body Models</td>
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<td>VII</td>
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<td>Stapp Car Crash Journal</td>
<td>Published</td>
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*Editor’s choice award at the Journal of Biomechanical Engineering

*John W. Melvin Student Award: 3rd Place
I. REFERENCES


Toyota (2010) Documentation of Total Human Model for Safety (THUMS) AM50 Pedestrian/Occupant Model


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

Matthew L. Davis 1,2, Brian C. Allen2, Carol P. Geer2, Joel D. Stitzel1,2, F. Scott Gayzik1,2

1Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
2Wake Forest School of Medicine, Winston-Salem, NC

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II. 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 5th percentile female (F05) were 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 5th percentile female, Anthropometry, Injury, Modeling, Segmentation
II. 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 at both the full body level (Hayes et al. 2014; Toyota 2010; Yang et al. 2006) and body regional level (DeWit and Cronin 2012; Li et al. 2010; Shin et al. 2012; Soni and Beillas 2015). 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 (W.H.O 2013). 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 (Gayzik et al. 2011; Gayzik et al. 2012).

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. In the late 1990’s, there was a series of
fatalities resulting from airbag related injuries in otherwise low to moderate severity frontal crashes. Upon investigation, it was found that 78% of the fatalities were female, and 82% of the females were less than 163 cm tall (below average height) (Summers et al. 2001). This indicated that vehicle safety features needed to be evaluated for drivers beyond the average sized male, and provided motivation to expand the methods of vehicle safety evaluation. From a regulatory perspective, this was addressed by including the response of a small female anthropomorphic test device (Hybrid III 5th percentile female) to evaluate the ability of vehicle safety devices to protect a wider range of occupants. In an effort to further the ability of computational models to provide comparable data, this study focuses on the development of a model of a 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 (Evans 2001; Morris et al. 1998). 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) (Mackay and Hassan 1999). 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 (Ulfarsson and Mannerling 2004). 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 (Evans 2001). More specifically, small females are also more likely to
sustain chest injuries with a maximum AIS of 2 or greater (Welsh and Lenard). 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 (King and Yang 1995; Manary et al. 1998; Melvin et al. 1993). Increased injuries may also be attributable to lower structural strength within females due to lower bone mineral content (Duma et al. 1999).

Until recently, finite element models of the 5th percentile female (F05) have typically been developed by applying scaling techniques to existing 50th percentile male models, since scan data of such a specific target anthropometry are 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 (Serre 2006; Vezin and Verriest 2005). 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 (Happee et al. 2000). 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 (Iwamoto et al. 2003). This model was developed by scaling the THUMS average sized male using anthropometric data of the small female occupant from the University of Michigan (Schneider et al. 1983). However, in order to account for gender differences, thoracic and pelvic regions of the model were developed *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.
(Kimpara et al. 2002). 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 (Kimpara 2005). 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 5th percentile female.

The objectives of this study are two-fold. The first is to present comprehensive image and anthropometrical data of the F05 subject selected in this study. 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 (Beillas 2009; Hayes et al. 2013). 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.
II. 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. The initial screening process was conducted via advertisements with limited anthropometric data self-reported during the phone screen. Once candidates were identified, 15 anthropomorphic measurements were acquired and compared to existing anthropometry values presented by Gordon et al (Gordon et al. 1989). 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.

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 (Gayzik et al. 2009; Gayzik et al. 2011). 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 2. Examples of the multi-modality image dataset can be seen in Figure 3.

CT scans were acquired using a GE LightSpeed, 16-slice scanner. Due to the fast acquisition time, high resolution, and strong contrast for bone, CT scanning was selected as part of the development process. All images were acquired with the scanner in helical mode, with the subject placed in both supine and quasi-seated postures. The inability to achieve a true seated posture was due to limitations presented by the scanner’s restrictive bore size (72 cm). Therefore, to place the subject in the quasi-seated position, custom foam and acrylic inserts were attached to the scanning table and were rotated about the Y-axis (per SAE J211) such that the subject fit through the bore. For all scans, the matrix size was 512 mm x 512 mm. The CT scan protocol was established after consulting with a board certified radiologist and was designed to minimize radiation dose. No contrast was used on the subject in this study. Females who were pregnant or did not report the use of a contraceptive were excluded from the study.

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. Scans for other regions were non-breath held. Scan acquisition parameters were TR = 5.26 ms, TE = 1.8 ms, flip angle = 10°, 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. Images were primarily collected in the transverse plane, with coronal images obtained for select anatomy in the head and abdomen. Upon completion of image acquisition, all images were reformatted to a matrix size of 512 x 512.

The uMRI scans were obtained using a 0.6 Tesla Fonar Upright MRI (Fonar, Inc., Melville, NY) and Sympusle v.7.0 software. 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 = 30°, 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 23°. The images taken in the standing posture were of the shoulder, thorax, abdomen and standing knee.

<table>
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<tr>
<td></td>
<td>MRI</td>
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<td>2</td>
</tr>
<tr>
<td></td>
<td>uMRI</td>
<td>320</td>
<td>1.5</td>
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<tr>
<td></td>
<td>CT</td>
<td>500</td>
<td>0.63</td>
</tr>
<tr>
<td>Thorax</td>
<td>MRI</td>
<td>480</td>
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<tr>
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<tr>
<td>Upper Extremity</td>
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External Anthropometry

A 7-axis 3D FaroArm 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 (Gayzik et al. 2012). The data were collected with an accuracy of ±0.68 mm for surface scanning and ±0.02 mm for digitized landmarks. 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 (Gayzik et al. 2012). 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 A 1) (Gayzik et al. 2012; Schneider et al. 1983). 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 A 2). Directly following landmark data collection, while the subject was still seated in the buck, a scan of the subject’s surface topography was acquired using the FaroArm with an integrated laser attachment. 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 acquired using similar techniques further described by Gayzik et al (Gayzik et al. 2012).

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.

**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 (Gayzik et al. 2011).

**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 quasi-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 limitation to the quasi-seated CT scan was its inability to accurately capture the correct curvature of the lumbar spine in the seated position. In order to overcome this limitation, the assembly data for the lumbar spine was taken from the seated uMRI scans, where a more realistic seated posture was possible due to the open bore nature of the scanner. By coupling the lumbar spine curvature with the thoracic and cervical spines from the seated CT, the full spine was able to be assembled in the seated position.

**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 were 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 segmented 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 uMRI 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 were preferable for initial segmentation due to 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.

**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.
Figure 4. Schematic of model assembly process, a) Cervical spine masks from supine CT segmentation, b) 3D polygon surface data from segmentation placed in the quasi-seated CT image space, c) Cervical spine is transformed as a group into the model space and aligned with the skull and thoracic spine using bony landmarks (blue).

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

Figure 5. Skeletal structures segmented from supine MRI data (outlined in red) placed in the uMRI image space.
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 uMRI 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.

II. RESULTS

In total, 2 subjects passed the initial screening of height and weight. However, only one of these volunteers passed the secondary screening of external anthropometry measurements from Gordon et al. Ultimately, 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. 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 3. The recruited subject was also used for external surface and bony landmark data collection. 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 A 3.
Table 3. Anthropometric measurements of the 5th percentile female volunteer

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Subject</th>
<th>Target (Gordon et al. 1989)</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. Using this data, 3D geometries were developed for all skeletal structures and each organ that will be explicitly modeled. The skeleton consists of 182 individual bones. Explicit representations of 32 organs have also been developed, represented by brain, thoracic, and abdominal organs relevant 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 (obtained from a contrast-enhanced clinical CT scan), 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 4. 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. This was modeled as a single part and was conditioned to remove artifact (arising mainly from breathing). This shell of the volunteer developed from the external scan data was also compared to medical image data by body region for validation. 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 (Burmaster 1998; Gehan and George 1970). The assembled model can be seen in Figure 6.
### Table 4. 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>1039</td>
<td>Basal Ganglia</td>
<td>15.6</td>
</tr>
<tr>
<td>Right Lung</td>
<td>1196</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 internal organs that will be explicitly represented. Bony landmark data is shown as blue points.**

### II. 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. 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) (Gordon et al. 1989). 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. While other data sources, such as the CAESAR anthropometry database were considered, there were a number of requirements for the current study that were not contained in that study. By obtaining our own prospectively recruited data, we were able to obtain landmark and surface data in a controlled driving posture based on seating accommodation models. Additionally, medical image data from
the same subject used for the anthropometry component was another requirement for this dataset that was not previously available.

Across the 15 anthropometric measurements recorded, the waist circumference had the largest deviation from the target. While a large deviation from the target anthropology in this area can be problematic due to anticipated interaction with simulated countermeasures, like seatbelts or airbags, this is a body region that can be adjusted in the final model development. The deviation from the target waist circumference can be attributed to the amount of subcutaneous fat found in the volunteer. From a modeling perspective, disagreement in this measurement is manageable because subcutaneous fat tissue can be readily reduced in the CAD development stage without affecting the morphology of surrounding structures. In addition, for the final model, subcutaneous fat generally acts to serve as a passive transfer for energy in the dynamic loading simulations.

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.
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 (Geraghty et al. 2004). 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 relevant to crash induced injuries (CIIs). Typically, organs of interest for CIIs 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.

Using supine CT data, measurements of cortical bone thickness were also obtained through segmentation. Due to limits imposed by the full width half max of the scanner, thickness measurements were limited to bones with diaphyseal measurements greater than 2.75 mm. This excluded bones, such as the ribs, with cortical thickness values throughout the structure that fall below the scanner cut-off that was used in this work. In these cases, cortical thickness values will be applied in the modeling phase using techniques from the literature (Kim et al. 2012; Li et al. 2010; Li et al. 2010). Ultimately, cortical thickness measurements were limited to specific long bones (humerus, radius, ulna, femur, fibula, tibia and femur) and compared to published data. The dataset was compared to a review of radiographic measurements of cortical bone published by Virtama et al (Virtama and Helela 1969). 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 5. 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 5. Bone cortical thicknesses comparison to literature

<table>
<thead>
<tr>
<th>Bone</th>
<th>F05 Subject (mm)</th>
<th>Virtama et al. (Virtama and Helela 1969) Average (mm)</th>
<th>Minimum (mm)</th>
<th>Maximum (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Humerus</td>
<td>4.9</td>
<td>4.4</td>
<td>2.95</td>
<td>5.4</td>
</tr>
<tr>
<td>Radius</td>
<td>3.5</td>
<td>3.3</td>
<td>2.1</td>
<td>4.45</td>
</tr>
<tr>
<td>Ulna</td>
<td>3.7</td>
<td>3.5</td>
<td>2.5</td>
<td>5.8</td>
</tr>
<tr>
<td>Femur</td>
<td>8.4</td>
<td>8.5</td>
<td>6.7</td>
<td>11.65</td>
</tr>
<tr>
<td>Tibia</td>
<td>5.5</td>
<td>5.75</td>
<td>3.8</td>
<td>7.45</td>
</tr>
<tr>
<td>Fibula</td>
<td>4.0</td>
<td>3.3</td>
<td>2.15</td>
<td>4.4</td>
</tr>
</tbody>
</table>

One of the main limitations of this work was the sample size. However, the decision to take this approach was made pragmatically. The careful recruitment of one anthropometrically representative female allowed resources to be allocated for the collection of an extensive amount of both medical image data and external surface data that would not have been feasible on a larger sample of subjects. By taking this approach, medical image data was available to validate results of the developed geometries.

This work has attempted to not only present the development of a small female finite element model, but also to provide geometric data that otherwise is limited in the literature. Data on the organ volumes and cortical thickness values of the small female can be used for future modeling applications. Also, detailed anthropometric landmarks will be useful for future reconstruction of small female models. This will also facilitate more accurate model to model comparisons. However, 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 relevant to the modeling of CIIs, 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 (Gayzik et al. 2011). 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.

II. 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,
these data will be used as part of a larger effort in developing a detailed finite element model of the 5\textsuperscript{th} percentile female for human injury prediction in vehicular crash.

II. ACKNOWLEDGEMENTS

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).
II. APPENDIX A

Table A 1. Adjustable buck parameters (Gayzik et al. 2012)

<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>Heel 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 A 2. 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 2nd Metacarpal, 2nd Prox. Phalange</td>
</tr>
<tr>
<td>Med. Clavicle*</td>
<td>MedC</td>
<td>Clavicle</td>
<td>Second Metacarpal*</td>
<td>SM</td>
<td></td>
</tr>
<tr>
<td>Suprasternale (Manubrium)</td>
<td>SSM</td>
<td>Sternum</td>
<td>5th Metacarpal*</td>
<td>FM</td>
<td></td>
</tr>
<tr>
<td>Substernale (Xyphoid Process)</td>
<td>SSX</td>
<td>Sternum</td>
<td>Lat. 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>Med. 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, Tibia, Talus, Foot Complex 1st Metatarsal, 1st Prox. Phalange, Foot Complex 5th Metatarsal, 5th Prox. Phalange, 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></td>
</tr>
<tr>
<td>Posterior Superior Iliac Spine*</td>
<td>PSIS</td>
<td>Pelvis</td>
<td>Ball of Foot*</td>
<td>BF</td>
<td></td>
</tr>
<tr>
<td>Pubic Symphysis</td>
<td>PS</td>
<td>Pelvis</td>
<td>5th Metatarsal*</td>
<td>FMT</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 A 3. Coordinates of bony landmarks per SAE J1733 coordinate system

<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>
<tbody>
<tr>
<td>TH</td>
<td>-228.1</td>
<td>0.0</td>
<td>-656.4</td>
<td>85.6</td>
<td>0.0</td>
<td>-1477.8</td>
</tr>
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Figure A 1. F05 Brain CAD model showing detailed structures. C – cerebrum, SSS – superior sagittal sinus, TS – transverse sinus, LV – lateral ventricle, 3rd ventricle, 4th ventricle, BS – brainstem, BG – basal ganglia, CC – corpus callosum, TH – thalamus, CB – cerebellum, FC – falx cerebri, T – tentorium (transparent), F - Fornix

Figure A 2. Head and Neck region showing detailed rendering of 52 neck muscles. Neck muscles are symmetric about the midplane.
Figure A 3. Thorax and abdomen anterior and posterior views of contents. Bone data is slightly transparent. Heart, Lungs, great vessels, diaphragm, spleen, kidneys, pancreas, stomach, gallbladder, jejunum, colon, small intestine volume, and bladder are shown.

Figure A 4. Plex CAD data. The right side shows components of the F05 knee CAD in an oblique anterior view: FC – femur cartilage, TC – tibial cartilage, PC – patellar cartilage, PL – patellar ligament (transparent).
II. REFERENCES


Toyota (2010) Documentation of Total Human Model for Safety (THUMS) AM50 Pedestrian/Occupant Model


Chapter III: Thoracoabdominal Organ Volumes for Small Women

Matthew L. Davis \textsuperscript{1,2}, Joel D. Stitzel\textsuperscript{1,2}, F. Scott Gayzik\textsuperscript{1,2}

\textsuperscript{1}Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
\textsuperscript{2}Wake Forest School of Medicine, Winston-Salem, NC

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III. ABSTRACT

Objective: Thoracoabdominal injuries commonly occur as a result of motor vehicle crashes. In order to design occupant protection systems that reduce risk of injury, 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 small female by stature and weight because of their use in restraint system development. Geometric data on thoracoabdominal organs are needed to construct accurate representations of female occupants. This study aimed to gather information on organ volumes from clinical medical imaging studies of small females.

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, pancreas, gallbladder, lungs, and heart. Segmentations were conducted using semi-automatic techniques. Additionally, diametric measurements of the vena cava, aorta, trachea, and colon were obtained from the medical images at discrete locations using linear measurement tools.

Results: A total of 14 adult scans were selected with stature and weight ranges of 145.0 to 162.6 cm and 43.7 to 65.5 kg respectively. The following are the average thoracoabdominal organ volumes: liver (1224.5 ± 220.7 ml), spleen (151.6 ± 42.1 ml), left kidney (123.7 ± 20.1 ml), right kidney (115.4 ± 20.9 ml), heart (417.8 ± 36.6 ml), pancreas (54.1 ± 11.8 ml), and gallbladder (20.6 ± 13.4 ml). The average diameters were
19.7 ± 3.2 mm and 17.7 ± 5.1 mm for the vena cava and aorta respectively. The colon had an average diameter of 37.9 ± 7.1 mm.

**Conclusion:** Data characterizing the small female are important to validate the geometries used in computational models, including models derived from scaling techniques and those developed using subject specific medical imaging. The goal of this study was to use a sample of subjects anthropometrically representative of small females to evaluate the average volume for organs commonly injured in motor vehicle crashes. Based on these data, the right and left lungs were strongly correlated with stature and the heart was strongly correlated with weight. 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
III. 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 frequency and economic costs (Cavanaugh 1993; Klinich et al. 2008). When looking at injuries on the Abbreviated Injury Scale (AIS), thoracic trauma accounts for 39% of all AIS 3-6 injuries (serious to fatal) and abdominal trauma accounts for 16.5% of AIS 4+ (severe or worse) injuries and 20.5% of AIS 5+ (critical or worse) injuries (Elhagediab and Rouhana 1998; Kemper et al. 2009; Ruan et al. 2003). While there have been previous efforts to quantify abdominal loading in anthropomorphic test devices (ATDs), such as the frangible abdomen and fluid-filled abdomens, no current ATDs are equipped to represent or provide organ specific response to impact loading (Rouhana et al. 1990; Rupp et al. 2001; Viano and King 2000). Also, there are currently no established standards for inferring injury to these soft tissues. As a result, researchers are beginning to turn to computational models in an effort to obtain a greater understanding of the biomechanics and injury mechanisms of these organs.

Computational human body modeling is a growing field within biomechanics that has the capability to assess localized responses of internal organs. One such tool being used by researchers to study 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 (Yang et al. 2006). Traditionally, these models were explicitly developed to represent an average male (50th percentile in terms of stature 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. For example, an investigation of fatalities resulting from airbag-related injuries in otherwise low to moderate severity frontal crashes in the late 1990’s revealed that 78% of the fatalities were female, and 82% of the females were less than 163 cm tall (below average stature) (Summers et al. 2001). This information provided motivation to develop vehicle safety devices for a wider range of occupants and was addressed from a regulatory perspective by including the response of a small female anthropomorphic test device (Hybrid III 5th percentile female). Accordingly, this study aims to expand the literature related to the female driver below average of stature and weight.

Small females are generally believed to be 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 (Kimpara 2005; Manary et al. 1998; Melvin et al. 1993). 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 change in velocity for the crash duration was considerably lower for females than for males (Mackay and Hassan 1999). Work conducted by Tavris et al. also suggests that female occupants are more susceptible to injury during collisions (Tavris et al. 2001). 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 (Dischinger et al. 1995; Welsh and Lenard).
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, small female models have primarily been developed by scaling existing 50th percentile male models, since medical image data for such specific target anthropometry are limited. The techniques used to scale these models rely on data from external anthropometry databases (Iwamoto et al. 2007; Kimpara 2005; Vezin and Verriest 2005). 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. However, a similar comparison is also appropriate even when the model is developed using medical images of a specific subject because of the biodiversity of human subjects. Currently, there is no dataset that specifically addresses the volumes of thoracoabdominal organs for the small female. Because changes in organ volumes can affect the inertial response of the abdomen, it is important for internal organs to be accurately represented during model development.

Medical images are commonly used to assess organ volumes in vivo (Prassopoulos and Cavouras 1994; Prassopoulos et al. 1997; Schiano et al. 2000; Zhu et al. 1999). Currently, one of the largest datasets available for organ volume comparisons was published by Geraghty et al (Geraghty et al. 2004). In this study, the authors reported the normal distribution of female organ volumes for the 1st percentile organ volume to the 99th percentile. However, the reported volumes were corrected for stature and weight and therefore are not necessarily representative of a female subject of specific anthropometric measurements. For example, a 5th percentile female in terms of stature and weight may not necessarily have organs that fall in the 5th 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 representing females below average stature and weight. Data acquisition focused on organs that are relevant to crash induced injuries (CIIs), namely the liver, spleen, right kidney, left kidney, right lung, left lung, pancreas, gallbladder, and the heart. In addition, data were 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 small female.

III. 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, female patients whose stature and weight ranged from the 1st percentile to the 50th percentile were sought (145 -163 cm and 43 to 72 kg) (Fryar et al. 2012). Selected individuals were required to be free of abdominal or systemic injuries, have all organs present, and be between the ages of 18 and 50 to reduce the effects of age-related volume changes. Stature and weight values were pulled from patient records and all images were anonymized and approved for use under Wake Forest University’s Institutional Review Board (IRB# 00006511). Ultimately, 14 subjects were selected, comprised of 4 magnetic resonance imaging (MRI) scans and 10 contrast-enhanced computed tomography (CT) scans of the chest, abdomen, and pelvis. All images were acquired with the subject in the supine position. The CT scans were acquired for routine clinical evaluation with a patient population comprised of inpatients and outpatients. The MRI scans were completed as part of a larger dataset obtained explicitly for model development (Gayzik et al. 2011;
Gayzik et al. 2012). 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 injury 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 large areas of low attenuation indicating lesions. Within the selected scans, no abnormal features, such as blurring or movement artifact were seen. In total, the medical images of 14 subjects were used for data acquisition. The ranges of observed stature and weight were 145.0 to 162.6 cm and 43.7 to 65.5 kg respectively. Stature, weight, and slice thickness values for each subject can be seen in Table 6. Within this table, the stature, weight, and BMI percentiles are displayed according to anthropometry results from Fryar et al (Fryar et al. 2012). Percentile values were calculated from Z-scores based on the reported mean and standard deviation. Figure 8 compares the stature and weight data for each subject to BMI values of 20, 25, and 30.

![Figure 8. Weight and stature for selected subjects. Grey lines represent stature and weight percentiles.](image-url)
Table 6. Stature, weight, BMI and slice thickness for each subject

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+ Percentile values were calculated from Fryar et al.
*Denotes MRI scan, all others were CT

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 were collected on either a 1.5 Tesla Twin Speed scanner (GE, Milwaukee, WI) or a 3T Siemens Skyra (Gayzik et al. 2011). To increase the distinction of organ boundaries, 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.

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. Segmentations were limited to thoracoabdominal organs germane to CIIIs, namely the liver (n = 14), spleen (n = 14), right kidney (n = 14), left kidney (n = 14), pancreas (n = 13), gallbladder (n = 14), right lung (n = 10), left lung (n = 10), and heart (n = 13). 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 extraneous volume (Geraghty et al. 2004).

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 of large longitudinal and portal fissures of the liver were removed. 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.
Following segmentation, each organ was rendered in 3D and used for volume calculations using optimal resolution settings within Mimics. After rendering, the contours of each 3D surface were verified against the original mask. The volume calculations for the regions of interest were completed using algorithms within Mimics, employing a merging cubes algorithm and smoothing parameters to accurately evaluate the areas between slices. Examples of the rendered organs can be seen in Figure 9.

Additionally, diametric measurements were taken of the large intestine, aorta, vena cava, and trachea. All measurements were taken using linear measurement tools and were acquired in either the axial or coronal view to evaluate maximum diameter. In the aorta and the vena cava, the diameters were evaluated at three and five discrete locations respectively. For the vena cava, the measurements were taken at the point of its exit from the heart (n = 12), the inferior exit from the liver (n = 14), and at the level of the superior surface of L3 (n = 12). The aorta measurements were taken at the exit of the aorta from the heart (n = 13), peak of the aortic arch (n = 13), the inferior portion of the aortic arch (n = 13), at the level of the superior surface of T11 (n = 14), and at the level of the superior surface of L3 (n = 14). The L3 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 10.
Similar to the vasculature measurements, tracheal diameters were evaluated at 3 locations. The first measurement was taken 1 cm superior to the aortic arch. The remaining measurements were taken after the initial bifurcation of the trachea. For both the left and right branch of the major bronchi, the measurement was made midway between the tracheal bifurcation and the insertion into the lung. Lastly, for all but one subject, the colon was evaluated at 4 locations: the midpoints along the ascending, descending, transverse, and sigmoid colons.
III. RESULTS

In total, 116 organs were segmented and 196 diameter measurements of representative small female subjects were collected. The small female livers had an average volume of $1224.5 \pm 220.7$ ml. The average volumes of the left and right kidneys were $123.7 \pm 20.1$ ml and $115.4 \pm 20.9$ ml respectively. The gallbladder was most variable with regards to total volume, with an average volume of $20.6 \pm 13.4$ ml. The heart was the least variable with an average volume of $417.8 \pm 36.6$ ml. Volumetric data for each observed organ can be seen in Figure 11. Within the plot, the red line represents the median, and the edges of the boxes represent the 25th percentile ($q_1$) and 75th percentile ($q_3$). The whisker length (w) represents maximum and minimum values, with outliers being identified if they are greater than $q_3 + w(q_3 - q_1)$ or less than $q_1 - w(q_3 - q_1)$. Also, each plot includes the average organ volume with the standard deviation in parentheses.

Correlation analyses were also performed to compare organs to stature, weight, and BMI in addition to inter-organ volume effects. The correlation coefficients for each of these variables are displayed in Figure 12. Additionally, linear regressions were performed for each organ using stature and BMI as predictors of organ volume. The resulting regression equation coefficients, $R^2$ value, and root mean square error (RMSE) for each organ can be seen in the appendix in Table B 1. For each predictor, the
corresponding estimates for the 50\textsuperscript{th} percentile female (F50) and 5\textsuperscript{th} percentile female (F05) according to the stature and BMI (calculated from stature and weight) values from Fryar et al. have been reported.

![Correlation matrix for stature, weight, BMI, and each organ](image)

**Figure 12. Correlation matrix for stature, weight, BMI, and each organ**

With regards to lung segmentations, conditions such as atelectasis or incomplete representation in the medical image resulted in the exclusion of 4 subjects from analysis. Additionally, 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 4 vena cava measurements (out of 42) and 3 aortas measurements (out of 70). The average diameter of the vena cava was found to be 19.7 ± 3.2 mm. The average diameter of the aorta was 17.7 ± 5.1 mm. Tracheal and primary bronchi diameters were
15.3 ± 1.3 mm and 12.3 ± 1.6 mm respectively. The total average diameter of colon measurements was 37.9 ± 7.1 mm. With regards to segmental measurements of the colon, the ascending colon was the widest, with an average diameter of 42.3 ± 7.8 mm. Data for each diametric measurement can be seen in Figure 13.
III. DISCUSSION

Medical images have proven to be a valuable diagnostic tool for evaluating organ volumes in vivo (Geraghty et al. 2004; Zhu et al. 1999). Advances in software algorithms have aided both the speed and accuracy of obtaining organ specific volumes (Pekar et al. 2004). Also, as the resolution of medical images has increased, it has become possible to more accurately distinguish the border between organ parenchyma and surrounding 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.

Because of the importance of the small female in the regulation of automotive safety and the increasing use of human body computational models, it is important to understand how changes in stature and weight of small females can affect the volumes of internal organs. This knowledge is directly applicable to computational models, where volumetric changes in organs can affect the inertial response during blunt impact events. Based on the obtained data, stature and weight variations amongst small females do seem to play a role in total organ volume. With regards to stature, the left and right lungs have the strongest correlations with r values of 0.8 and 0.84 respectively. These correlations agree with previous studies that have shown linear correlation between lung volume and stature over small ranges of stature (Hepper et al. 1960). The left and right lungs were also strongly correlated with each other (r = 0.99). The spleen was found to have a moderately positive correlation to stature with an r value of 0.58. No other observed organ volumes had strong correlations with stature. Weight, however, was found to have
a strong correlation to heart volume \((r = 0.83)\), with several organs showing moderate correlation (spleen \((r = 0.64)\), left lung \((r = 0.4)\), right lung \((r = 0.42)\), and pancreas \((r = 0.42)\)). In terms of inter-organ volume correlations, strong correlation was found between the right and left kidneys \((r = 0.88)\). Moderate correlation was found between liver and spleen volumes \((r = 0.64)\), spleen and heart volumes \((r = 0.61)\), and spleen and lung volumes \((r = 0.63\) for right lung and \(r = 0.64\) for left lung). While the sample size of 14 is a limitation for definitively establishing standards for organ volumes based on stature and weight, this data will serve as a useful comparison for establishing models across the distribution of small females. Also, the results reported in Figure 12 are useful for displaying organs that do not seem to be significantly affected by stature and weight.

Because of their role as organ tethers, accurate modeling of large thoracoabdominal vasculature and the trachea is also important from an injury biomechanics perspective. Previous literature has shown that stature and weight do have a correlation with vessel diameter (Lopes-Berkas and Jorgenson 2011; Sandgren et al. 1999). Prince et al. reported that the mean diameter of the vena cava was \(20 \pm 3\) mm. Similar results were found for the small female in this study with a mean vena cava diameter of \(19.7 \pm 3.3\) 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. As seen in Figure 13C, the vena cava is larger at its exit from the liver than it is superior to the heart. It should also be noted that fluid intake can have an effect on the vessel diameter. In 2008, Kosiak et al. reported that the level of fluid intake can cause the vena cava diameters to fluctuate approximately 8% (Kosiak et al. 2008). The values for small female aortic diameters are also in line with previously published values and exhibit a
nearly linear decrease from the superior aorta to the measurement taken at L3 (Hager et al. 2002; Mao et al. 2008).

There are numerous applications for evaluation organ volumes of specific anthropometries. The organs evaluated in this study were selected due to their prevalence of injury as a result of motor vehicle crashes. In particular, the data obtained are applicable to the development of computational models of the thorax and abdomen of small women. 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 from scaling. In both cases, it is important that the proper inertial response of internal anatomy be represented. Accurately characterizing these geometries enables models to more meaningfully predict localized tissue responses. For example, in simulations involving motor vehicle accidents, correctly implementing thoracoabdominal anatomy is important to glean reasonable data from impacts involving airbag deployment or seatbelt interactions. This includes not only local responses but also how they can affect the global response of the model. Potentially, the data can 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 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 thorax and abdomen in dynamic impact environments. 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. This data was obtained to both characterize the thoracoabdominal region of the small female and also to provide insight as to how stature and weight affect the internal structures of this cohort. By examining these values for a variety of statures and weights, this dataset will provide a useful comparison for both current and future full-body models of the small female.

III. ACKNOWLEDGEMENTS

Funding for this study was provided by the Global Human Body Models Consortium, LLC (GHBMC) through GHBMC Project Number: WFU-005. The authors would like to thank Gabrielle Lynn Rawls for her assistance with data collection.
### III. APPENDIX B

Table B 1. Multivariate regression characteristics for organ volume prediction using stature and BMI as predictors

<table>
<thead>
<tr>
<th>Organ</th>
<th>Intercept</th>
<th>Stature</th>
<th>BMI</th>
<th>R²</th>
<th>Root Mean Square Error</th>
<th>F05 Estimate (ml)</th>
<th>F50 Estimate (ml)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liver</td>
<td>-225.13</td>
<td>5.55</td>
<td>26.27</td>
<td>0.08</td>
<td>229.6</td>
<td>1191.8</td>
<td>1386.4</td>
</tr>
<tr>
<td>Spleen</td>
<td>-819.18</td>
<td>5.07</td>
<td>8.19</td>
<td>0.56</td>
<td>30.4</td>
<td>125.9</td>
<td>224.6</td>
</tr>
<tr>
<td>L Kidney</td>
<td>60.07</td>
<td>-0.06</td>
<td>3.23</td>
<td>0.18</td>
<td>19.8</td>
<td>122.4</td>
<td>137.9</td>
</tr>
<tr>
<td>R Kidney</td>
<td>12.43</td>
<td>0.10</td>
<td>3.87</td>
<td>0.22</td>
<td>20.1</td>
<td>113.0</td>
<td>133.5</td>
</tr>
<tr>
<td>Heart</td>
<td>-397.85</td>
<td>3.71</td>
<td>10.69</td>
<td>0.69</td>
<td>22.4</td>
<td>397.5</td>
<td>493.2</td>
</tr>
<tr>
<td>L Lung</td>
<td>-8974.98</td>
<td>62.08</td>
<td>21.95</td>
<td>0.66</td>
<td>307.6</td>
<td>865.6</td>
<td>1683.0</td>
</tr>
<tr>
<td>R Lung</td>
<td>-9917.59</td>
<td>69.12</td>
<td>24.56</td>
<td>0.73</td>
<td>289.7</td>
<td>1041.6</td>
<td>1952.3</td>
</tr>
<tr>
<td>Pancreas</td>
<td>-69.75</td>
<td>0.50</td>
<td>2.04</td>
<td>0.19</td>
<td>11.6</td>
<td>50.7</td>
<td>66.6</td>
</tr>
<tr>
<td>Gallbladder</td>
<td>109.27</td>
<td>-0.62</td>
<td>0.3</td>
<td>0.1</td>
<td>13.8</td>
<td>22.5</td>
<td>16.9</td>
</tr>
</tbody>
</table>
III. REFERENCES


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

Matthew L. Davis $^{1,2}$, Nicholas A. Vavalle$^{1,2}$, Joel D. Stitzel$^{1,2}$, F. Scott Gayzik$^{1,2}$

$^1$Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
$^2$Wake Forest School of Medicine, Winston-Salem, NC

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IV. ABSTRACT

Computed tomography scans are a valuable tool for developing computational models of bones. The objective of this study is to present a method to generate computer aided design (CAD) representations of long bones from clinically based CT scans. A secondary aim is to apply the method to six long bones (humerus, ulna, radius, femur, tibia, fibula) from a sample of 3 individuals who match anthropometric targets of a 5th percentile female, a 50th percentile male, and a 95th percentile male (by height and weight). Periosteal and endosteal bone surfaces were segmented and used to calculate the characteristic cortical thickness, $T_c$, at one mm increments along the bone axis. In the epiphyses where the value of $T_c$ fell below the scanner threshold, the endosteal bone layer was replaced using literature values projected inward from the periosteal surface. On average, $74.7 \pm 7.4\%$ of the bone geometry was above the scanner cut-off and was therefore derived from the CT scan data. The thickness measurement was also compared to experimental measurements of cadaveric bone and was found to predict average cortical thickness with an error of 3.1%. This method presents a possible solution for the characterization of characteristic thickness along the length of the bone and may also aid in the development of orthopaedic implant design and subject specific finite element models.

Keywords: Biomechanics, full width half max, subject specific bone model, cortical thickness, CAD, area moment of inertia
IV. INTRODUCTION

Numerical models of bone have become an important part of orthopedic biomechanics in recent years (Huiskes and Hollister 1993). Such models have been used in the development and evaluation of orthopedic implants (Harrysson et al. 2007; Mellal et al. 2004), surgical guidance (Taddei et al. 2003), and subject specific finite element models (FEMs) for biomechanics research (Anderson et al. 2005). In particular, FEMs combined with greater computing power are an integral part to understanding injuries to the human body (Li et al. 2010; Park et al. 2013; Yang et al. 2006). To achieve biofidelic 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 the mechanical response of bone and, if fracture is predictable in a given model, the timing, location, and pattern of fracture (Clark et al. 2007; Cowin 2001; Li et al. 2010; Zdero et al. 2010). An overestimation of cortical thickness will cause an overestimation of stiffness and result in an unrealistic response of the model to loading (Cowin 2001; Currey 2002).

Many research studies have used CT to determine the cortical thickness of bones (Anderson et al. 2005; Au et al. 2008; Louis et al. 1995; Newman et al. 1998; Prevrhal et al. 1999; Prevrhal et al. 2003; Rizzo et al. 2011; Treece et al. 2010; Viceconti et al. 1998), primarily for the purpose of developing computer aided design (CAD) representations and FEMs of these structures (Anderson et al. 2005; Au et al. 2008; Keyak et al. 1990; Viceconti et al. 1998). However, it has been shown in both computational (Dougherty and Newman 1999) and experimental studies (Newman et al. 1998; Prevrhal et al. 1999; Prevrhal et al. 2003) that the limit thickness where the
compact bone can 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 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 (Viceconti et al. 1998).

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 impact design. 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. The current study focused on a method to generate thickness estimates along the entire main axis of each selected long bone.

The aim of this study is to present a method to generate CAD representations of selected long bones. 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. For this study, CAD development was performed on 3 individuals representing a range of target anthropometries. The sizes were selected to correspond to the three main driving adult driving populations evaluated by the National Highway Traffic Safety Administration using anthropomorphic test devices 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.
IV. METHODS

CT images of a living 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. The individuals were scanned as part of a larger effort in developing global standard finite element models of the human body, and details of the scanning protocol can be found in previous publications (Gayzik et al. 2011). 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 (Gayzik et al. 2012; Gayzik et al. 2011; Gordon et al. 1989). 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.

CT is the primary modality used to determine cortical thicknesses because of its ability to produce a visible contrast between cancellous and cortical bone. The periosteal and endosteal surfaces of each bone were identified using by selecting pixels exceeding 226 Hounsfield Units (HU) within a commercial image processing software package (Mimics v. 15.0, Materialise, Leuven, Belgium). Prior to thresholding, an attenuation histogram was observed to verify starting Hounsfield units (Au et al. 2008; Gayzik et al. 2011; Sinha 2009). The range of values for thresholding was based on values recommended for adult cortical bone from Mimics and also based on line thresholding.
Following thresholding, a gradient magnitude filter was employed and manual correction was used to fix any missing or erroneous pixels. The error associated with these segmentation techniques is characterized by the voxel size of the image sets, namely 0.38 mm$^3$. To reduce the error involved in this approach, care was taken to follow published guidelines in place for the segmentation of bony tissue. Manual segmenting was used to separate bones in joints and in regions too thin for thresholding. The polygonal surfaces obtained from the initial segmentations were then conditioned using relaxation and the removal of single-point spikes in the polygon mesh. All relaxation functions were performed using the relaxation feature within Geomagic Studio (v12, Geomagic, Rock Hill, SC). This function has three parameters, smoothness level, strength, and curvature priority. Default values of smoothness were used, but curvature priority was maximized in order to maintain the curvature of the geometry. Strength had a large influence on the overall smoothing, and was minimized in all cases so that volume of the geometries had a volume change of less than 1%. Spikes were removed based on angle deficiency. In each case, the parameter for angle deficiency was set to ensure that volume change was less than 1%. Once conditioned, the polygon surfaces were used to create Non-Uniform Rational B-Spline (NURBS) surfaces using Studio software that were water tight to a tolerance of 0.01mm. NURBS surfaces were used because they produce smaller file sizes than discrete polygonal surfaces for the surfaces in use in this work (Noorani 2006). 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.
To determine the average cortical thickness ($T_c$) at planes along the length of the bone, a visual basic script was written in a CAD software package (Rhinoceros v. 4, McNeel, Seattle, WA). $T_c$ was calculated at a finite number of planes perpendicular to the long axis (Z axis) of the bone (Figure 14-A). 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 14-B). The location of the contour line pairs was identified by the distance from the basis plane along the Z-axis.

$T_c$ was calculated at each cross section along the bone axis using the inner and outer circumference per Equation 1, (Figure 14-C). Using this method, $T_c$ values were determined along the main axis of the bone in one mm increments. Equation 1 was used due to changes in the relative distance between these contours.
Figure 14. 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.

\[
T_c = \frac{A}{\frac{1}{2}(p_{outer} + p_{inner})}
\]

where \( A \) = cross sectional area between the outer and inner contours (mm²) 
\( p_{outer} \) = outer contour perimeter length (mm) 
\( p_{inner} \) = inner contour perimeter length (mm)

Dougherty et al. provided a relationship between FWHM and FOV for scans with an FOV of up to 42 cm (Dougherty and Newman 1999). Good agreement was found between the FWHM of the scanner used in the present study and the data provided by Dougherty. Extrapolating from that data and based on an FOV of 50 cm, the limit thickness for cortical bone reconstruction for the current study was considered 2.75 mm.
The $T_c$ along the Z axis was calculated for each bone and 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 from the cutoff through the proximal and distal ends of the bone. Cortical thickness from literature sources were used in the epiphyseal regions by projecting inward from the periosteal surface. In regions where no data was available, 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 (Gayzik et al. 2011).

Within Studio, deviation spectral analysis and tangential splines were used to join the segmented (diaphyseal) and prescribed (epiphyseal) regions. The result is the projected surface connected by a transition region (Figure 15). Upon completion, each bone was imported back into the image space and compared to the original scan data and matched with typical differences in volume and surface areas of the components of less than 1% (Figure C 1).
Two qualities of the Z transition region were assessed: the $G_1$ continuity condition over the region and the Z axis length of the region. $G_1$ continuity was verified within Studio by performing checks for tangential continuity at the boundary of each surface plane for each long bone region. 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 15). 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. The transition length was calculated as the distance along the Z axis from the scanner cut-off.
to the average of the $R_R-R_i$ 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.

Two approaches were applied as an initial verification of the presented methods, one computational and the other experimental. For computational verification, a 270 mm cylinder with known thicknesses was developed using a CAD software program (Rhinoceros v. 4, McNeel, Seattle, WA). The $T_c$ was then evaluating using the methods described above.

The method was also compared to data obtained from physical measurements of cortical thickness. For this comparison, three post-mortem human femurs were obtained (2 male, 1 female). These cadaveric femurs were obtained with corresponding CT images (slice thickness and in plane resolution of 0.625 mm and 0.703 mm respectively), but had the added benefit that they were able to physically evaluated. Femur samples were gathered with the assistance of the Wake Forest School of Medicine Orthopedic Surgery research department. Following procurement, CT images of each femur were obtained. The medical images were then used to develop the corresponding CAD data. Next, $T_c$ was measured for each cadaveric femur using the methods described above.

Following the computational analysis, discrete locations on the physical bone were evaluated manually using calipers (accuracy of ± 0.025 mm). Two discrete cross-sections along the shaft (165mm and 100mm from proximal) were sectioned using an oscillating bone saw. These locations were selected because they corresponded to locations that fell within the scanner’s thickness reconstruction cut-off. Specifically, the 165 mm location was chosen because it roughly estimated the region of maximum
cortical thickness for the bones. The second location had a thinner cortical thickness across all femurs. Once cross-sections had been obtained, cortical thickness measurements were evaluated at 45 degree increments around the circumference of the bone and averaged. For the epiphyseal region of the bone, the periosteal surface of the bone CAD was evaluated by comparing maximum femoral head circumference to the experimental femurs.

IV. RESULTS

The final CAD models were composed of G1-continuous, concentric NURBS surfaces representing the periosteal and endosteal cortical layers. 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 16.
Figure 16. 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 7. On average, 74.7 ± 7.4% of the original CAD data on the long axis was retained based on the $T_c$ analysis. This value corresponds to regions of bone that were above the scanners cut-off and had regions of cortical thickness greater than 2.75mm. The resulting cortical thickness compares favorably to the literature in both diaphyseal and epiphyseal regions (Virtama and Helela 1969). 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>Bone</th>
<th>Bone Z Axis Length (mm)</th>
<th>Distance along bone (prox. to distal) with greatest measured $T_c$%</th>
<th>Maximum Measured $T_c$ (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Humerus</td>
<td>296</td>
<td>58.5%</td>
<td>4.9</td>
</tr>
<tr>
<td>Ulna</td>
<td>217</td>
<td>55.3%</td>
<td>3.7</td>
</tr>
<tr>
<td>Radius</td>
<td>211</td>
<td>45.0%</td>
<td>3.5</td>
</tr>
<tr>
<td>Femur</td>
<td>410</td>
<td>39.8%</td>
<td>8.4</td>
</tr>
<tr>
<td>Tibia</td>
<td>333</td>
<td>62.2%</td>
<td>5.5</td>
</tr>
<tr>
<td>Fibula</td>
<td>312</td>
<td>66.4%</td>
<td>4.0</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Bone</th>
<th>Bone Z Axis Length (mm)</th>
<th>Distance along bone (prox. to distal) with greatest measured $T_c$%</th>
<th>Maximum Measured $T_c$ (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Humerus</td>
<td>338</td>
<td>74.1%</td>
<td>6.7</td>
</tr>
<tr>
<td>Ulna</td>
<td>277</td>
<td>57.4%</td>
<td>5.1</td>
</tr>
<tr>
<td>Radius</td>
<td>260</td>
<td>46.4%</td>
<td>4.7</td>
</tr>
<tr>
<td>Femur</td>
<td>465</td>
<td>35.4%</td>
<td>7.5</td>
</tr>
<tr>
<td>Tibia</td>
<td>408</td>
<td>59.1%</td>
<td>7.5</td>
</tr>
<tr>
<td>Fibula</td>
<td>401</td>
<td>55.5%</td>
<td>4.3</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Bone</th>
<th>Bone Z Axis Length (mm)</th>
<th>Distance along bone (prox. to distal) with greatest measured $T_c$%</th>
<th>Maximum Measured $T_c$ (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Humerus</td>
<td>364</td>
<td>79.7%</td>
<td>6.0</td>
</tr>
<tr>
<td>Ulna</td>
<td>297</td>
<td>17.2%</td>
<td>5.5</td>
</tr>
<tr>
<td>Radius</td>
<td>275</td>
<td>51.6%</td>
<td>4.6</td>
</tr>
<tr>
<td>Femur</td>
<td>529</td>
<td>29.9%</td>
<td>9.0</td>
</tr>
<tr>
<td>Tibia</td>
<td>439</td>
<td>52.8%</td>
<td>7.4</td>
</tr>
<tr>
<td>Fibula</td>
<td>424</td>
<td>39.2%</td>
<td>3.8</td>
</tr>
</tbody>
</table>
In order to characterize geometrical changes in the bone around its circumference, area moment of inertia calculations were performed at 1 mm increments using the contours. X and Y axis moments were calculated about the centroid of each contour by position the x-axis along the anterior-posterior direction of the bone and the y-axis along the medial-lateral direction of the bone (Figure 17 and Figure 18).

With regards to verification of the methods, the initial evaluation of the script was conducted using the CAD cylinder with variable thickness. The script was found to correctly calculate the thickness along the length of the cylinder (Figure C 2). From the experimental measurements, the average error of the computational measurement was 3.1% below the measured thickness, but within one standard deviation in all cases. A summary of these measurements can be seen in Table 8. With regards to outer surface measurements of the femoral head, the average deviation between the physical and computational measurements was less than 1% (Table 9).

<table>
<thead>
<tr>
<th></th>
<th>Femur 1</th>
<th></th>
<th>Femur 2</th>
<th></th>
<th>Femur 3</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Cortical Thickness</td>
<td>Cortical Thickness</td>
<td>Cortical Thickness</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Degree</td>
<td>(mm)</td>
<td>(mm)</td>
<td>(mm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>D = 165</td>
<td>D = 100</td>
<td>D = 165</td>
<td>100</td>
<td>165</td>
</tr>
<tr>
<td>45</td>
<td>5.03</td>
<td>6.35</td>
<td>4.04</td>
<td>3.81</td>
<td>8.03</td>
</tr>
<tr>
<td>90</td>
<td>6.17</td>
<td>9.91</td>
<td>5.18</td>
<td>5.59</td>
<td>6.71</td>
</tr>
<tr>
<td>135</td>
<td>8.38</td>
<td>6.22</td>
<td>7.47</td>
<td>3.30</td>
<td>7.06</td>
</tr>
<tr>
<td>180</td>
<td>7.37</td>
<td>4.80</td>
<td>4.47</td>
<td>5.08</td>
<td>8.92</td>
</tr>
<tr>
<td>225</td>
<td>5.99</td>
<td>4.19</td>
<td>6.96</td>
<td>3.99</td>
<td>7.04</td>
</tr>
<tr>
<td>270</td>
<td>7.59</td>
<td>5.46</td>
<td>5.99</td>
<td>3.96</td>
<td>9.78</td>
</tr>
<tr>
<td>315</td>
<td>4.50</td>
<td>6.22</td>
<td>6.10</td>
<td>4.06</td>
<td>6.86</td>
</tr>
<tr>
<td>St. Dev.</td>
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<td>1.70</td>
<td>1.19</td>
<td>0.79</td>
<td>1.10</td>
</tr>
<tr>
<td>Average</td>
<td>6.91</td>
<td>6.19</td>
<td>5.66</td>
<td>4.36</td>
<td>7.78</td>
</tr>
<tr>
<td>CAD</td>
<td>6.70</td>
<td>5.85</td>
<td>5.38</td>
<td>4.32</td>
<td>7.72</td>
</tr>
</tbody>
</table>

*D = distance from proximal end (mm)
Figure 17. X-axis area moment of inertia.
Figure 18. Y-axis area moment of inertia.
Table 9. Comparison of the computational method to experimental circumference measurements of the femoral head.

<table>
<thead>
<tr>
<th></th>
<th>Femur 1</th>
<th>Femur 2</th>
<th>Femur 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>CAD</td>
<td>156.6</td>
<td>155.9</td>
<td>144.1</td>
</tr>
<tr>
<td>Physical</td>
<td>157.1</td>
<td>155.2</td>
<td>144.4</td>
</tr>
</tbody>
</table>

IV. DISCUSSION

In the past, researchers have used a number of techniques to develop geometrical representations of cortical bone. For example, some such as Rizzo et al. and Buie et al. have used imaging techniques such as micro-CT and peripheral quantitative CT (pQCT) to evaluate localized and regional cortical bone thicknesses (Buie et al. 2007; Rizzo et al. 2011). In 2006, Bessho et al reported on the development of CT-based finite element models of the proximal femur (Bessho et al. 2007). In that case, the authors used global thresholding and a closing algorithm for extracting the geometry of the bone. Rather than explicitly model cortical bone from CAD, the authors developed cortical bone within their finite element model as shells with prescribed thicknesses. Au et al. developed CAD of the knee joint by automatically prescribing contours to the inner and outer surfaces of cortical bone (Au et al. 2008). These were done on a series of images smoothed using Gaussian operators and identified using gradient-based edge detection. However, all of these techniques require either micro-CT of the samples or making assumptions in the thin region of bone. The work presented in this study attempts to use clinical CT data in the region of bone above the inherent scanner limitations and a mixture of physical measurements and micro-CT measurements found in the literature to develop thin regions. This approach was chosen to implement the strengths of both techniques for model development.
Long bones were chosen for this study due to their geometric qualities and their relative importance in biomechanical studies. However, 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, narrowing the field of view, or using micro CT scans that can produce resolutions that are not feasible for living subjects. (Li et al. 2010; Li et al. 2010) (Anderson et al. 2005). However, clinical scan databases are appealing because of their overall size and the variability of the subject pool they represent.

Figure 16 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 (Wolff 1892). To the extent possible, literature data was leveraged for developing epiphyseal regions. However, this data is only available on aggregate, with very limited literature on the epiphyseal cortical thickness for specific genders, ages or body habitus. While this approach results in a CAD model that is not truly subject specific, the approach provides a method for using clinical CT datasets where segmentation data is no longer accurate. Because these regions cannot be characterized due to PSF limitations of the scanner, it was deemed more scientifically sound to use data that has been published in the literature.
Characteristic mid-shaft cortical thickness values were compared to data obtained from work by Virtama and Helela (Figure 16). This study was referenced because it utilized a large number of samples taken from direct measurement of cadaveric bone 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). 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 $T_c$ found in the study, but nothing can be inferred from the standard deviations found by Virtama with relation to the F05 and M95. For further validation, data compared favorably to a literature review of cortical thickness values presented by Gayzik et al (Gayzik et al. 2011).

Another point to note is that a cross section of a long bone (illustrated in Figure 14-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. Area moment of inertia data were also gathered along the length of the bone to show geometrical properties of the bone related to bending resistance (Figure 17 and Figure 18). Area moment values also compare favorably to existing biomechanical data in the literature (Augat et al. 1996; Bouxsein et al. 1994; Kennedy 2004; Myers et al. 1993; Rittweger et al. 2000). However, these data are typically only reported at a discrete location along the bone making a comparison of the full geometry difficult. The
advantage of the present approach is that the thickness values and area moment of inertia calculations are determined as a continuum along the shaft’s length.

The techniques in this paper can also be extended to other medical and biomechanical applications. Computational bone modeling is 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 (Prendergast), the biofidelity of such models has been greatly enhanced with introduction of accurate computational geometries from computed tomography (Skinner Hb Fau - Kim et al.). As demonstrated by Zdero et al, changes in cortical thickness can cause large changes in the stiffness of bone in response to mechanical loading. 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 (Zdero et al. 2010). The data collected using the presented technique for measuring cortical bone thickness outline a potential method for assigning cortical thicknesses to bones used in blunt impact simulations.

IV. CONCLUSION

Accurate geometrical reconstruction of cortical bone is critical for biomechanical modelers and designers since bones are the primary load path for external forces. 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. 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 5\textsuperscript{th} percentile female, 50\textsuperscript{th} percentile male, and a 95\textsuperscript{th} 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.

IV. 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

IV. ACKNOWLEDGEMENTS

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 Daniel Moreno for his overall assistance related to the GHBMC project.
IV. APPENDIX C

Figure C 1. A) Initial segmentation of cortical bone. B) Overlay of initial segmentation (dark blue) and resulting smoothed NURBS surface (transparent). C) Contour of final CAD compared to medical image.

Figure C 2. Thickness measurements from the CAD cylinder.
IV. REFERENCES


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Chapter V: An Evaluation of Mass-Normalization Using 50th and 95th Percentile Human Body Finite Element Models in Frontal Crash

Matthew L. Davis 1,2, Nicholas A. Vavalle1,2, F. Scott Gayzik1,2

1Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
2Wake Forest School of Medicine, Winston-Salem, NC

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V. ABSTRACT

Human body finite element models (FEMs) are ideal tools to explore the effects of body habitus on the biomechanical response of a given subject in a vehicle crash. This study aims to investigate the differences between a large male (M95) FEM and an average male (M50). The models are identical aside from their respective morphologies. The same generic frontal crash driver-side buck was used with each model with an acceleration pulse from a late model NCAP crash. The HIC$_{15}$ of the M50 and M95 models were 491 and 806. The Brain Injury Criteria values for the M50 and M95 models were 0.50 and 0.64, respectively. Neck Injury Criteria for the M50 and M95 models were 0.44 and 0.41, respectively. The percent chest deflections were 21.1% for M50 and 17.3% for M95. Equal stress, equal velocity scaling was used to scale M95 outputs to the M50. Mass scaling was found to increase correlation of signal phases, but had diminished effect on magnitude and shape. Given the global trend of increased size of occupants, this study provides insight into the effects of body habitus on the occupant kinematics and injury risk in frontal crash.

**Keywords** Finite Element (FE), Human Body Modelling (HBM), injury biomechanics, large male, model morphing.
V. INTRODUCTION

According to the United States Census Bureau, there are approximately 113 million adult males in the USA (Baudrit and Trosseille 2015), meaning that there are over 5 million that are at least the 95th percentile stature. This segment of the population is under-represented in the United States New Car Assessment Program (US NCAP); currently, no 95th percentile male anthropometric test devices (ATDs) are included in the protocols. Further, the USA adult population increased in both height (3 cm increase) and weight (11 kg increase) between 1960 and 2002 (Ogden et al. 2004). It has been shown that tall occupants can be left unprotected by side airbag systems and can strike the vehicle structure instead of the airbag (Loftis et al. 2011). This study aims to present injury risk differences between an average male and large male, using two validated human body finite element models (FEMs) in an identical simulated crash condition.

Two models from the Global Human Body Models Consortium (GHBMC) were used in this study – the average male model (M50) and the large male model (M95). A regional model approach was used in the development and validation of the M50 model. Head (Mao et al. 2013; Takhounts et al. 2013; Yanaoka and Dokko 2013), neck (DeWit and Cronin 2012; Fice et al. 2011; Mattucci et al. 2012; Mattucci et al. 2013), thorax (Li et al. 2010; Li et al. 2010), abdomen (Soni and Beillas 2013), pelvis (Kim et al. 2012) and lower extremity (Shin and Untaroiu 2013; Untaroiu et al. 2013; Yue and Untaroiu 2014) models were developed and validated by consortium universities and were then integrated into one full body model (Hayes et al. 2014; Park et al. 2013; Vavalle et al. 2013; Vavalle et al. 2013; Vavalle et al. 2013). The full body M95 model was morphed
from the M50 model and was subsequently validated in 7 conditions (Vavalle et al. 2014).

The objective of the current study is to evaluate the effects of body habitus on the outcomes of a representative consumer crash test. Both the M50 and M95 were simulated in a simplified driver buck designed for frontal impacts with a United States New Car Assessment Program (US NCAP) acceleration applied. The effects of body habitus were tested in several ways, including kinematic response and injury risk assessment. Finally, a quantitative curve comparison was done to evaluate how closely the response of the models matched after scaling.

However, due to the stature and weight differences between the M50 and M95 models, measured variables from simulations must be normalized to a reference (in this case, the M50). Two common approaches for obtaining normalized and scaled response data are the impulse-momentum (Mertz 1984) and equal-stress equal-velocity (ESEV) methods (Eppinger 1976; Yoganandan et al. 2014). The impulse-momentum method takes into account the type of test and effective mass and length characteristics of specific body regions involved. Because of this, the impulse momentum is considered to be a more specific evaluation than ESEV. However, the ESEV method relies only on the mass ratio between the M50 and the M95. As such, this approach is applied in the current study, because all body regions in this case are coupled. ESEV sidesteps the potential error which could be introduced when determining effective masses and characteristic lengths of coupled body regions.
V. METHODS

Human Body Models

The Global Human Body Models Consortium (GHBMC) average male (M50) (Vavalle et al. 2014; Vavalle et al. 2013; Vavalle et al. 2013) and large male (M95) (Vavalle et al. 2014) LS-Dyna (971 R6.1.1, LSTC, Livermore, CA) models were used for this study. For both models, target stature and weight values were determined by those used in development of the M50 and M95 Hybrid III ATDs. The development approach for the M50 model can be found in the literature (Gayzik et al. 2011), but briefly, a multi-modality approach was used to collect imaging data of a living subject and subsequently made into CAD. The imaging approach leveraged the strengths of MRI, upright MRI, CT and external laser scanning to obtain highly detailed information of the subject in a seated posture. The CAD was delivered to collaborating universities where regional meshing and development was completed. The full body model was integrated from the five regional models and validated for mass distribution (Vavalle et al. 2013) and simulation responses (Vavalle et al. 2014; Vavalle et al. 2013), in addition to the regional validation that was performed by each university (DeWit and Cronin 2012; Li et al. 2010; Mao et al. 2013; Shin et al. 2012; Soni and Beillas 2015). In total, the average male model is composed of 1.3 million nodes, 2.2 million elements and 984 parts and represents a weight of 76.8 kg and height of 174.9 cm.

The large male model was morphed directly from the average male model using a radial basis function interpolation with a thin-plate spline basis function and relaxation algorithm (RBF-TPS) (Vavalle et al. 2014). The method smoothly interpolates between a reference geometry and target geometry using a relatively reduced set of landmarks.
across the whole body (Bookstein 1989; Donato and Belongie 2002). The RBF-TPS method required homologous landmarks on the reference (M50) and target (M95) geometries and the FE nodal locations of the reference model. The imaging data of both subjects were leveraged to acquire the input landmarks. For the development of the M95, homologous landmarks were established on the outer flesh and select abdominal organs (liver, kidneys, and spleen) of both the M50 and M95 to accurately capture the external anthropometry of the model while also representing the internal anatomy of the M95. Spline equation coefficients were calculated from the homologous landmarks and applied to the reference nodal locations to determine the target nodal locations. The relaxation algorithm was applied to improve element quality by eliminating the requirement that the spline functions pass through the landmarks. Anthropometric and anatomical verifications were performed to ensure that the relaxation did not hinder model fidelity to the target population size. The application of the RBF-TPS method produced a model that was not just a scaled version of the average male, but rather one that matched anthropometric and anatomical targets of a large male. One advantage of this morphing method is that the target model contains the same number of nodes, elements and parts as the reference model. All other modelling considerations also remained the same between the two models, but the M95 model represents a mass of 103.3 kg and stature of 189.5 cm. Direct comparisons between the models were facilitated since any given node in one model was in a homologous location on the other model.

Simulation

A simplified driver-side frontal crash buck design was used in this study. The buck contained several important features to allow for a reasonable approximation of a vehicle
interior. The seat was modelled using a low density foam material property (density = 4.06e-8 kg/mm$^3$, elastic modulus = 3.05e-2 GPa, and stress strain curve, nominally 11.3 KPa at 50% strain) with a shell coating of fabric material. The knee bolster also used a low density foam material property with discrete elements connected to the buck structure to tune the response. The airbag used in the model was a hybrid jetting type that fired 6 ms after the start of the crash event. Nitrogen gas was used to inflate the bag which was modeled with a fabric material (0.55 GPa modulus, fabric thickness 0.3 mm). The mass flow rate of gas into the airbag was determined empirically and provided by the GHBMC. At its fully inflated state, the bag had a radius of 540 mm. Delta-V values of the simulated crashes were in the range indicating the use of a Stage 1+2 bag (Gabler and Hinch 2008). The belt stiffness was equivalent to 10 kN of force at 5.23% strain and included a pretensioner that fired 9 ms after the start of the event and a retractor that fired 17 ms after the start of the event. Further, the shoulder-belt force was limited to 3.75 kN with a force-limiter. No changes were made to the restraint system parameters for the large male model.

The acceleration pulse applied to this buck was derived from the floor accelerometer of the US NCAP test #7147 (NHTSA. Updated 2012). The pulse was applied to the rigid floor of the buck and prescribed motion in the three linear degrees of freedom (DOF) with constraints applied for rotational degrees of freedom. The test was a frontal crash test into a barrier at 56.2 km/h. The total delta-v of this impact was 65.6 km/h, as seen in the velocity profile of the simulation in Figure 19.
To accommodate the size of the M95 model the seat was moved rearward by 145 mm and downward by 39.5 mm. For both simulations the models were settled into the seat, using gravity, for 100 ms. The belt was subsequently fitted on the settled model using the built-in belt-fitting capabilities of LS-PrePost v4.2 (LSTC, Livermore, CA). The crash event was then simulated to 150 ms.

All simulations were computed using LS-Dyna v6.1.1 MPP (LSTC, Livermore, CA). The Wake Forest University Distributed Environment for Academic Computing (DEAC), a heterogeneous Linux-based high-performance computing system, was used. Both models were simulated on 48 processors.

**Model Output Methods**

Head accelerations were obtained from the kinematics of the skull using the history function in LS-PrePost. Head rotational velocities from the models were collected using a constrained interpolation method. A node at the centre of gravity (CG) of the head was constrained to nine nodes on the skull using the *CONSTRAINED_INTERPOLATION card in LS-Dyna. This allowed for a deformable skull that was capable of constraining
the CG node. Other constraint methods typically require a rigid body. A cross-section plane at the occipital condyles (OCs) was used to obtain neck forces and moments (White et al. 2013). To calculate chest deflections, nodal coordinates were taken on a node of the sternum and a node of the T8 spinous process. These were then subtracted to obtain deflections, and normalised by the initial value to give percent deflections. Femur forces were obtained using cross-section sets that were approximately 2/3 of the length from the proximal femur to the distal femur. All outputs were obtained in local coordinate systems commensurate with the SAE J211 standards (SAE 2014). The trajectories of the inboard (right) side of the models relative to the buck were tracked by averaging a cluster of five nodes at each of the head, shoulder and hip, which are representative of typical kinematic marker locations.

Several measurements were taken from the interactions between the human body models and vehicle structure, as well. The belt forces at the upper shoulder and outer lap-belt locations were taken from the seatbelt element force outputs. The upper shoulder-belts were force-limited in both cases, and this output was used to confirm the function of this feature. The forces between the airbag and the body were examined using the contact forces. Finally, the distance between a node on the forehead and a node on the top of the steering wheel was measured.

**Injury Risk Evaluation**

For both models, a quantitative injury risk assessment was performed for individual body regions based on injury risk criteria in the literature. These values were also used to evaluate the effect of body habitus on potential injury outcomes in an otherwise identical
sled pulse. For the head, neck and lower extremity regions, data were obtained using the accelerometer and section-plane techniques described above.

Head injury risk was evaluated by applying two techniques that utilised different model outputs. Existing regulations from the National Highway Traffic Safety Administration (NHTSA) specify performance limits for a Head Injury Criteria (HIC) (Eppinger et al. 1999; Mertz et al. 1997; Prasad and Mertz 1985). An industry standard, HIC$_{15}$ is a measure of the maximum average translational acceleration over any 15 ms duration of an impact event. All outputs used in the calculation of HIC$_{15}$ scores were obtained from acceleration of the head CG node of both the M50 and M95 models. Head injury risk was also evaluated using the Brain Injury Criteria (BrIC) developed by Takhounts et al. (Takhounts et al. 2013). This injury criterion differs from HIC in that prediction is not based on acceleration data. Instead, prediction is determined by the directional dependence of maximum angular velocities. To calculate BrIC for both models, angular velocity data was obtained from the head CG node that used the constrained interpolation method.

Neck injury risk was assessed using the Neck Injury Criteria (Nij). Nij is evaluated by linearly combining the normalised axial load and normalised flexion/extension moment about the occipital condyle (Eppinger et al. 1999; Prasad and Daniel 1984). In both axial load and moments, the critical neck intercept values used for normalisation were applied per the suggested values for the M50 and M95. Axial load and neck moment data were obtained from the section plane placed at the occipital condyle to mimic data acquisition from ATD tests.
The risks of injury to thoracic structures in the models were evaluated using two displacement criteria. The first was the compression criteria established by Kroell et al. (Kroell et al. 1971). The compression criterion is a linear equation that uses maximum chest compression percentage as a predictor for the level of AIS severity. Here, chest compression is calculated by the maximum chest displacement divided by initial chest depth. The second method used for evaluation was the chest deflection criteria, where maximum chest deflection was related to AIS injury risk based on risk curve equations used in the US NCAP (Laituri et al. 2005; Sohr and Heym 2009; Transportation 2006).

As the M50 and M95 models were simulated in a frontal sled pulse with knee contact to the bolster, femur tolerance loads were used to assess risk of injury to the lower extremity. These values were obtained from the models using section planes in the femur corresponding to the location of data acquisition in ATDs (Eppinger et al. 1999). To use the M95 data in the injury risk equation, femur loads from that model were scaled by the femur cross-sectional area scale factor established by Mertz et al. (Mertz et al. 1989). M50 data were not scaled since the injury risk equation was developed for an average male.

ISO Comparison with Eppinger Scaling

Part of ISO/TS 18571, a technical standard regarding the quantitative assessment of dynamic data, was used to quantitatively compare the model output curves. The standard uses a weighted average of four metrics – corridor, phase, magnitude and slope – to derive a total score describing the fit of data to a reference curve. Scores can range from 0 to 1, with 1 being the best. The score is categorised into excellent, good, fair, or poor Total ISO Ratings, based on the thresholds 0.94, 0.80 and 0.58, respectively. For the
purposes of this study, only the phase, magnitude and slope scores were calculated. Further explanation of this is included within the discussion section.

The phase, magnitude and slope scores are calculated using the methods of Enhanced Error Assessment of Time Histories (EEARTH) (Zhan et al. 2011), which compares a model curve to a single reference curve. The phase score uses the time shift of the maximum cross correlation in a linear regression. Zero time shift receives a phase score of 1 and the maximum allowable time shift of 20% receives a phase score of 0. The magnitude and slope scores are calculated from the time-shifted data. Dynamic time warping (DTW) is used in calculating the magnitude score to parse out the effects of phase on magnitude differences. The one-norm of the difference between the shifted and warped curves are calculated for each time point and normalised by the one-norm of the shifted and warped reference curve. The magnitude score is calculated using a linear regression between no difference (magnitude score 1) and a maximum allowable difference of 0.5 (magnitude score 0). The slope score is calculated from the time-shifted data without DTW applied, since DTW compromises the time-dependence of slope. The average slope is calculated in 1 ms intervals and the same one-norm procedure used for the magnitude calculation is performed. The slope score is calculated as the linear regression between zero difference (slope score 1) and a maximum allowable difference of 2.0 (slope score 0).

Typically, the reference curves represent average experimental data and the curves being assessed are from a model or ATD. However, in the case of this study, the reference curves were the M50 model data and the M95 model data were being assessed. The M95 data were scaled to the M50 data using the ESEV methods developed by
Eppinger (Eppinger 1976). This approach assumes linear relationships among time, length, and mass (Equations 1-3), where T is the unit of time, L is the unit of length, and M is the unit of mass.

\[ T_{M50} = \lambda_t T_{M95} \]  
(1)

\[ L_{M50} = \lambda_l L_{M95} \]  
(2)

\[ M_{M50} = \lambda_m M_{M95} \]  
(3)

ESEV also assumes that both models have identical density and modulus of elasticity. By applying these assumptions, and substituting equations 1-3 into the functions for density and elastic modulus, normalization factors can be found for time, deflection, acceleration, force, and moment (Equations 4-8). Full derivation of these normalization factors can be found in the literature (Eppinger 1976; Yoganandan et al. 2014).

\[ Time \text{ Normalization Factor} = \lambda^\frac{1}{3} \]  
(4)

\[ Deflection \text{ Normalization Factor} = \lambda^\frac{1}{3} \]  
(5)

\[ Acceleration \text{ Normalization Factor} = \lambda^{-\frac{1}{3}} \]  
(6)

\[ Force \text{ Normalization Factor} = \lambda^\frac{2}{3} \]  
(7)

\[ Moment \text{ Normalization Factor} = \lambda \]  
(8)

In order to make observations on the efficacy of mass scaling between these models, the ISO comparison was made between unscaled M95 data to M50 as well as scaled M95 data to M50, with the unscaled data acting as a baseline against which the scaled scores were compared. The head accelerations (x, y and z), neck axial force, neck flexion/extension moment, chest deflection, and left and right femur forces were
compared using this method. The hypothesis was that, after mass scaling, model responses would be improved as compared to before scaling and would be closer to scores of 1.

V. RESULTS

Both models completed the simulation without numerical error. The peak hourglass energy, as a percentage of the internal energy and total simulation energy, was 7.3% and 0.5% in the M50 simulation and 12.0% and 1.0% in the M95 simulation. Hourglass energies were also observed on a part by part basis, and largest contributors were shown to be stable in the model (Takhounts et al. 2003). Further, there were no discontinuities in the hourglass energy curve throughout either simulation. Discontinuities in this curve can be indicative of a numerical instability.

Kinematics

The simulations can be seen in Figure 20, where time-lapse images of both models in the crash event are included. The trends of the model responses appear similar, with the exception that the additional mass of the M95 brings the model further towards the steering column, increasing the load on the airbag. This is confirmed in the contact forces between the airbag and body, where the peak of the M50 is 8.3 kN and the peak of the M95 is 9.2 kN. While the shoulder-belt forces were held constant by the force-limiter, the lap-belt forces between the two models differed. The peak lap-belt force in the M50 model was 6.6 kN, whereas the peak in the M95 model was 9.1 kN. A graph showing the restraint force information is shown in Figure 21.
Figure 20. Time-lapse images of the M50 and the M95 in the frontal NCAP simulation.
Figure 21. Restraint force data from the M50 and M95 models. Airbag forces are plotted on the left, while belt forces are plotted on the right.

The trajectories of the head, shoulder and hip were found to differ between the two models, as seen in Figure 22. In all three locations the M95 model displayed greater forward motion than the M50 model. The total magnitude of forward excursion was found to be less in the hip area, which was restrained by the seat and lap-belt. The head displacement in the X direction was 154.3% greater in the M95 and the shoulder displacement was 139.5% greater. In the hip, the X direction motion of the M95 was 139.9% greater than the M50. The total Z motion in the hip was the smallest gross displacement of any of the regions for both models. To shed light on possible injury risk, the relative distance between the head of the models and the steering wheel was also observed. The distance between a homologous node on the forehead of each model and
the top of the steering wheel was measured. The M95 model started approximately 125 mm further from the wheel than the M50 model and, at its closest, was 49.6 mm closer. This can be seen in Figure 23.

![Figure 22. Head, shoulder and hip trajectories of the average and large male for the duration of the simulation.](image1)

![Figure 23. Distance from the forehead of each model to the top of the steering wheel.](image2)

Injury Risk

The HIC<sub>15</sub> of the M50 and M95 models were 491 and 806, respectively, with corresponding injury risk values included in Table 10. This equates to a HIC<sub>15</sub> that is roughly 1.6 times larger in the M95 model than the M50 model. The value for the M95 was also above the suggested HIC<sub>15</sub> performance limit of 700. The BrIC values for the M50 and M95 models were 0.50 and 0.64, respectively. Nij values for the M50 and M95 models were 0.44 and 0.41, respectively, which in both cases is below the performance limit of 1.0. In both the M50 and M95, the appropriate critical intercepts were used to account for the difference in body habitus. The percent chest deflections were 21.1% for M50 and 17.3% for M95, lower for the larger model. Based on these values, the chest
compression criterion predicts an AIS injury level of 1 for both the M50 and M95. Within Table 10, injury probability to the thorax can be seen in terms of peak chest deflection. Peak femur forces for the left and right leg in the M50 model were 4.1 kN and 6.7 kN, while they were 6.0 kN and 6.9 kN in the M95 model. The largest difference was seen in the HIC$_{15}$ values of the models. Data on the probability of injury for the head, neck, thorax and lower extremity can be seen in Table 10.

<table>
<thead>
<tr>
<th>Criteria</th>
<th>M50</th>
<th>M95</th>
</tr>
</thead>
<tbody>
<tr>
<td>HIC$_{15}$</td>
<td>Model Value</td>
<td>Threshold</td>
</tr>
<tr>
<td></td>
<td>491.3</td>
<td>700</td>
</tr>
<tr>
<td>Nij</td>
<td>0.44</td>
<td>1.0</td>
</tr>
<tr>
<td>Thoracic Chest Deflection (mm)</td>
<td>36.9</td>
<td>63</td>
</tr>
<tr>
<td>Femur Force, Left (kN)</td>
<td>4.1</td>
<td>10.0</td>
</tr>
<tr>
<td>Femur Force, Right (kN)</td>
<td>6.7</td>
<td>10.0</td>
</tr>
</tbody>
</table>

**ISO Comparison**

The ISO/TS 18571 comparison, with corridor scores omitted, can be seen in Table 11, with both the native M95 data, as a baseline, and mass scaled M95 data. Overall, the scaled M95 data received scores that would have been commensurate with fair ratings given the full complement of ISO standard scoring. On average, mass scaling had the largest effect on the phase score, improving the average value by 0.13. The slope score was improved by 0.08. Mass scaling was found to have the least effect on the magnitude score. The highest scores tended to be in the mass scaled phase correlations, which ranged from 0.73 to 1.00. The magnitude scores were in the middle of the three scores, on average, with a range of 0.00 to 0.79. The outlier in this group was the hip z
displacement, which received the only 0.00 score out of any of the categories and was 0.56 below the next lowest magnitude score. Also, gross Z motion (upwards or downwards) was constrained by the seat- and lap-belt, and was small in comparison to the fore-aft direction. The lowest scores, on average, occurred on the slope, which ranged from 0.21 to 0.79.

Table 11. ISO/TS 18571 comparison between the models. For the purposes of this comparison, all M95 data labelled as scaled were done so using the Eppinger method, where $\lambda=0.74 \ (M50 = 76.8 \text{ kg} / M95 = 103.3 \text{ kg})$

<table>
<thead>
<tr>
<th>Signal</th>
<th>Phase Score</th>
<th>Magnitude Score</th>
<th>Slope Score</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unscaled M95</td>
<td>Scaled M95</td>
<td>Unscaled M95</td>
</tr>
<tr>
<td>Head CG Accel. X</td>
<td>0.64</td>
<td>0.90</td>
<td>0.89</td>
</tr>
<tr>
<td>Head CG Accel. Y</td>
<td>0.92</td>
<td>0.80</td>
<td>0.62</td>
</tr>
<tr>
<td>Head CG Accel. Z</td>
<td>0.66</td>
<td>0.88</td>
<td>0.68</td>
</tr>
<tr>
<td>Neck Force Axial</td>
<td>0.60</td>
<td>1.00</td>
<td>0.93</td>
</tr>
<tr>
<td>Neck Flex./Ext. Moment</td>
<td>0.80</td>
<td>0.98</td>
<td>0.78</td>
</tr>
<tr>
<td>Chest Deflection</td>
<td>0.94</td>
<td>0.73</td>
<td>0.57</td>
</tr>
<tr>
<td>Hip X Displacement</td>
<td>0.82</td>
<td>0.93</td>
<td>0.61</td>
</tr>
<tr>
<td>Hip Z Displacement</td>
<td>0.78</td>
<td>0.90</td>
<td>0.00</td>
</tr>
<tr>
<td>Left Femur Force</td>
<td>0.83</td>
<td>0.97</td>
<td>0.77</td>
</tr>
<tr>
<td>Right Femur Force</td>
<td>0.81</td>
<td>1.00</td>
<td>0.79</td>
</tr>
</tbody>
</table>
V. Discussion

The GHBMC average male and large male models were compared in a frontal US NCAP simulation. A simplified driver-side buck was used in the study to apply the crash pulse collected from US NCAP test #7147 in the NHTSA Online Database. The pulse and buck were identical between simulations, the only difference being the human body model and seat position. Since the M95 was morphed from the M50, this study focuses on the effect of this morphing on full body kinematics and injury risk.

The models displayed differences in outputs and injury risks, with the M95 model at an increased risk for injury to the head. A number of factors related to the size and stature differences between models led to these findings. The additional mass provided by the large male model accounted for a 9.2% increase in peak kinetic energy within the system. Increased stature required a more rearward seat track for the M95 model. Thus, inflation of the airbag occurred prior to model contact in the case of the M95. In comparison, the M95 contacted the airbag 54 ms into the simulation, whereas the M50 model made contact at 30 ms. The proximity of the chest to the airbag in the M50 model likely contributed to the larger chest deflection, and therefore injury risk observed. The distance of the M95 model from the airbag, coupled with the force limiting belt, led to more forward excursion before loading the countermeasures, yielding a shorter duration loading pulse (Figure 21). This shorter time of engagement increased forward head excursion (Figure 22 and Figure 23; Table 10) and led to greater linear and rotational head acceleration, as evidenced by increases in HIC and BrIC. Also, the M95 experienced knee bolster impact prior to engaging the airbag (44.5 ms), which may have contributed to the increase in forward excursion of the head and shoulders. While knee
bolster contact happened prior to airbag engagement for the M95, the M50 model struck the knee bolster earlier in the simulation (37.5 ms) because of its initial position closer to the wheel. Also, due to adjustments made to seat position, both models had similar locations of engagement with the airbag and knee bolster.

The model curves were shown to be quantitatively different through the use of ISO/TS 18571, a recently released standard for quantitative assessment of dynamic data sets. The M95 model is a morphed version of the M50 model. As such, these models lend themselves to investigating mass scaling effects since two assumptions of Eppinger’s method are perfectly held: that the reference and target subjects have the same modulus and density. No material model adjustments were made after morphing the M95 model, only nodal locations are different between the two models. Regarding the assumption of geometric similitude, the M95 is not an exact scaled version of M50 but morphed to match a typical 95th percentile male subject. However, the similitude is thought to be on par with PMHS subjects, and this mass scaling method is widely applied despite subject-to-subject variability. Given the arguments above, the ISO comparison could be regarded as an examination of the geometric similitude assumption. However, the fixed size of the buck is a confounding factor in this study, since the larger model interacted at different times and durations with the countermeasures, potentially altering the time histories in ways that scaling alone could not account for (as seen in the hip kinematics). Despite that limitation, mass scaling increased the quantitative comparison scores between models. Ideally, a mass scaling method would yield ISO scores approaching 1, with the target model (M50, in this study). The results indicated that scaling had the greatest effect on phase, with smaller effects seen on magnitude and shape. However, post hoc scaling of
data is, in itself, a model with assumptions and approximations. Future work will focus on using this approach to more closely evaluate the relative pros and cons of scaling techniques.

The full ISO/TS 18571 comparison uses constant-width corridors as a means of considering subject-to-subject variability. The corridor score was not included in this analysis since model outcomes are deterministic and variability would not be expected. Because of this, total ISO Scores and their corresponding ISO Ratings were not assigned to the curve comparisons.

The study was also limited in that vehicle deformation was not considered. A deceleration pulse was applied to the floor of the buck without the application of floor pan intrusion or potential effects on the steering column. For example, from the report of this NCAP test, the brake pedal was 50 mm closer to the front of the seat track in post-crash measurements and the centre of the steering-wheel hub was 11 mm closer. Future work could focus on full vehicle crash simulations with the human body models included, to explore the effects of body habitus on injury risk in a more real-world setting.

V. CONCLUSION

This work presented a comparative study between an average male and a large male human body finite elements model. The models were simulated in a US NCAP frontal impact test and biomechanical data and injury risks were evaluated. Significant differences were found between the models, with the large male at a higher risk for head injury and slightly lower risk for chest injury. Quantitative comparison methods were used, to assess the efficacy of mass scaling methods. Equal stress, equal velocity mass scaling resulted in modest gains in curve comparison, on average, providing a fair
response match. The mass scaling was found to have the greatest impact on reducing phase differences between models. These models can be leveraged within the injury biomechanics community to better understand the effects of body habitus on injury risk and design safety features tailored to larger occupants.

V. ACKNOWLEDGMENTS

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V. REFERENCES


Chapter VI: An Objective Evaluation of Mass Scaling Techniques Utilizing Computational Human Body Models

Matthew L. Davis $^{1,2}$, F. Scott Gayzik$^{1,2}$

$^1$Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
$^2$Wake Forest School of Medicine, Winston-Salem, NC

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VI. ABSTRACT

Biofidelity response corridors developed from Post Mortem Human Subjects are commonly used in the design and validation of anthropomorphic test devices and computational human body models (HBMs). Typically corridors are derived from a diverse pool of biomechanical data and later normalized to a target body habitus. The objective of this study was to use morphed computational HBMs to compare the ability of various scaling techniques to scale response data from a reference to a target anthropometry. HBMs are ideally suited for this type of study since they uphold the assumptions of equal density and modulus that are implicit in scaling method development. In total, six scaling procedures were evaluated, four from the literature (Equal-Stress Equal-Velocity, ESEV, and three variations of Impulse Momentum) and two which are introduced in the paper (ESEV using a ratio of effective masses, ESEV-EffMass, and a kinetic energy approach). In total, 24 simulations were performed, representing both pendulum and full body impacts for three representative HBMs. These simulations were quantitatively compared using the ISO-TS18571 standard. Based on these results, ESEV-EffMass achieved the highest overall similarity score (indicating it is the most proficient at scaling a reference response to a target). Additionally, ESEV was found to perform poorly for two degree-of-freedom systems. However, the results also indicated that no single technique was clearly the most appropriate for all scenarios.
VI. INTRODUCTION

Physical test data from Post-Mortem Human Subjects (PMHS) are commonly used to develop biofidelity response corridors for the purpose of describing a human response to a blunt impact. Typically, cadaver tests involve a sample of human subjects with large variations in their physical characteristics (i.e. height, weight, shape, age, etc.). This anthropometric diversity can lead to widely varying response data during dynamic impacts. As such, the data from these experiments are frequently normalized to a target anthropometry (i.e. the average male). This normalization is a mathematical procedure in which the response data is adjusted to account for the variation in the physical characteristics of the human subjects (Moorhouse 2013). The goal of normalization is to reduce variability within a dataset so that it can be used to characterize the response of a selected population. These response corridors are then frequently applied in the design and validation of human body surrogates, both anthropomorphic test devices (ATDs) and computational human body models (HBMs). Among human body surrogates, computational tools, such as the finite element method, have become a valuable supplement to physical testing because they offer a cost-effective way to evaluate and design safety systems within a dynamic impact environment. The last 20 years has seen a large increase in the number of human body models being developed and validated at both the full body level (Hayes et al. 2014; Iwamoto et al. 2007; Yang et al. 2006) and body regional level (DeWit and Cronin 2012; Li et al. 2010; Shin et al. 2012; Soni and Beillas 2015). To compare to the general population, such models are often developed to represent an average (50th percentile by height and weight) male occupant.
In order to predict the response of a greater percentage of the population, there is interest in developing such models to represent subjects of varying body habitus. Unfortunately, there is relatively little experimental data representing specific anthropometries beyond the average male. Therefore, the validation of such models is a challenge due to the limited amount of experimentally derived corridors for comparison. In order for the experimental data to be applicable to specific anthropometries, it is important that the data for comparison first be scaled to represent the target anthropometry of the corridors (i.e. the average male). While often used interchangeably with normalization, scaling is the process by which a response can be transformed from one standard to another (Yoganandan et al. 2014). For example, a response representing a small female can be scaled to the response of an average male. Therefore, scaling techniques are often employed to compare HBM outputs to a normalized average male response for the purpose of validation (Vavalle et al. 2014).

While several scaling methods have been reported in the literature, two of the most commonly applied techniques are equal-stress equal-velocity scaling (Eppinger 1976) and impulse momentum scaling (Mertz 1984). Both of these techniques apply certain assumptions related to the physical characteristics of the PMHS or HBM for which the response data is being compared. By assuming identical elastic modulus and geometric similitude, both procedures can be used to derive scaling factors for engineering variables of interest such as time, displacement, acceleration, and force. Despite the widespread use of these techniques, there is little published work comparing the techniques and their applications to different loading environments. A recent study by Moorhouse used metadata from PMHS experiments to evaluate the performance of these techniques in a
number of impact cases (Moorhouse 2013). While this work shed valuable light concerning the ability of these techniques to normalize experimental data to a set of biofidelity response corridors, it is unclear if differences are caused by experimental limitations, such as variance in subject morphology, as opposed to the scaling procedures specifically. Therefore, the objective of this study is to use computational HBMs to compare the ability of these techniques to scale response data from a reference to a target anthropometry. In addition, two modified scaling methods based on these well-known techniques are introduced and evaluated. HBMs are well positioned to conduct this study because volumetrically scaled versions of the same model satisfy the assumptions of identical elastic modulus and geometric similitude that are the foundation of the scaling techniques evaluated.

VI. METHODS

All simulations were performed using the Global Human Body Models Consortium’s (GHBMC) 50th percentile male (M50) as a baseline. Although large male (M95) and small female (F05) GHBMC models have been developed, the M50 model was volumetrically scaled using global scale factors of 0.859 and 1.104 to achieve a model mass representative of a 5th percentile female (F05VS, 50 kg) and a 95th percentile male (M95VS, 102.3 kg) respectively (Davis et al. 2014; Vavalle et al. 2014). Therefore, these models were not meant to match specific external anthropometric measurements, but rather a target full body mass. By taking this approach and leaving the constitutive material models unmodified, the models maintain geometric similitude and identical elastic modulus. A side-by-side comparison of the three models can be seen in Figure 24.
Figure 24. Comparison of human body models. A) F05VS. B) M50. C) M95VS.

The models were then simulated in rigid hub impacts, analogous to the classic two degree of freedom mechanical system, and full body rigid wall impacts, which are analogous to one degree of freedom impacts. In each case, simulations were performed that were used during the validation of the GHBMC M50 model to ensure reasonable biomechanical response as a baseline (Vavalle et al. 2015; Vavalle et al. 2013). The rigid hub simulations consisted of isolated impacts to the thorax, abdomen, and pelvis. Furthermore, two simulations were performed for each rigid hub impact, one where identical impactors were used for each body habitus and one where the impactors were volumetrically scaled with the same global scale factors used to scale the models. The scaled impactor simulations were performed so that the same relative surface area of the body was impacted for each model. In these simulations, the density of the rigid impactor was adjusted so that the input energy was the same for each model. The rigid wall impacts consisted of a lateral drop test and a lateral sled type simulation. All simulations were run using LS-Dyna v6.1.1, rev. 78769 on a Linux Red Hat 6 high
performance computing system (the Distributed Environment for Academic Computing, or DEAC cluster) maintained at Wake Forest University.

The chest impact used a 23.4 kg cylindrical hub impactor with a 15 cm diameter and a nominal impact velocity of 6.7 m/s. This impact was simulated by aligning the impactor against the sternum at the 4th intercostal space. Thoracic deflection was described as the change in chest depth throughout the simulation (Kroell et al. 1974). Forces were reported as the contact force of the impactor.

The abdominal impact involved a 2.5 cm diameter, 48 kg bar with an initial velocity of 6.0 m/s. This was a free-back impact and occurred at the level of the umbilicus (approximately L3) (Hardy et al. 2001). Compressible abdominal depth was defined as the abdominal depth from the anterior surface of L3 to the anterior surface of the body normal to L3, i.e., the L3 depth. This method was chosen because it is the effective limit to abdominal compression. The force of the impact was measured as the contact force of the rigid bar.

The pelvic impact used a square-faced impactor weighing 16 kg impacting with 800 J of energy. This required giving the impactor a 10 m/s velocity in the lateral direction. The pelvis impactor was aligned to contact the trochanter and iliac crest at 90 degrees, according to the literature (Bouquet 1998). Pelvic deflection was measured as the relative displacement between the outer hip and a node on the center of the sacrum. Similar to the other rigid impact simulations, the contact force of the impacting plate was used to obtain force data.

The lateral drop test was simulated to specify a 1 meter free fall onto an instrumented rigid surface (Stalnaker 1979). For this study, contacts were developed to
measure the force response of the thorax and pelvis. Lateral thoracic deflection was measured using two symmetric nodes on the lateral portions of the thoracic flesh. Similar to the isolated pelvic impact, pelvic deflection during the drop test was measured as the relative displacement between the outer hip and a node on the center of the sacrum.

The lateral sled tests were modeled as a 6.7 m/s lateral impact using a Heidelberg-type sled (Cavanaugh et al. 1990; Cavanaugh et al. 1993). The impact environment included a flat rigid wall as a backrest, a Teflon seat, and five rigid impacting plates located at the shoulder, thorax, abdomen, pelvis, and knee. Torso forces were obtained as the sum of the shoulder, thorax, and abdomen forces. The pelvis force was measured as the contacting force at the pelvis plate. Deflection data were obtained using the same methods described for the lateral drop test.

In total, 24 simulations were performed: three rigid hub impacts with three models each and two impactor size configurations as well as two rigid wall impacts with three models each. Similarly, three reference/target configurations were evaluated for each scaling technique: F05VS data scaled to M50, F05VS data scaled to M95VS, and M95VS data scaled to M50. Because the majority of experimental biomechanics data in the literature represents the M50 response, this scaling paradigm was developed to evaluate the two common scaling directions, a smaller specimen up to the M50 and a larger specimen down to the M50. In addition, scaling from F05VS to M95VS was included to evaluate the performance of the scaling techniques over what is essentially the full range of adult anthropometry. Therefore, this was done to evaluate each method in a variety of mass increments and scaling directions. A full test matrix for this study can be found in Table D1.
Equal-Stress Equal-Velocity

The first of six methods evaluated was the mass-based scaling procedure using total body mass, also known as the equal-stress equal-velocity technique (ESEV\textsubscript{1}, the subscripts are introduced here to assist the reader). This technique scales response data based solely on a mass ratio of the target models mass to the reference model to which the response is being scaled. The main assumption of this technique is that there are linear relationships between the length, mass, and time units (Equations (2)-(4)).

\[
L_{\text{target}} = \lambda_l L_{\text{ref}} \tag{2}
\]
\[
M_{\text{target}} = \lambda_m M_{\text{ref}} \tag{3}
\]
\[
T_{\text{target}} = \lambda_t T_{\text{ref}} \tag{4}
\]

Where L, M, and T refer to the length, mass, and time units respectively. The subscripts l, m, and t represent the specific scale factors (λ) for each dimension. Further, the ESEV\textsubscript{1} technique assumes identical density and elastic modulus between the reference and target.

\[
\rho_{\text{target}} = \rho_{\text{ref}} \tag{5}
\]
\[
E_{\text{target}} = E_{\text{ref}} \tag{6}
\]

Where ρ refers to density and E refers to elastic modulus. Based on these assumptions, a dimensional analysis can be performed to derive the scaling factors for other engineering variables. Applying the units of density and elastic modulus yields the following:

\[
\frac{M_{\text{target}}}{L_{\text{target}}^2} = \frac{M_{\text{ref}}}{L_{\text{ref}}^2} = \frac{\lambda_m M_{\text{target}}}{\lambda_l L_{\text{target}}^2} \tag{7}
\]
\[
\frac{M_{\text{target}} L_{\text{target}}}{T_{\text{target}}^2 L_{\text{target}}^2} = \frac{M_{\text{ref}} L_{\text{ref}}}{T_{\text{ref}}^2 L_{\text{ref}}^2} = \frac{\lambda_m M_{\text{target}} \lambda_l L_{\text{target}}}{\lambda_l T_{\text{target}}^2 \lambda_l^2 L_{\text{target}}^2} \tag{8}
\]
By isolating the scale factors from this dimensional analysis, factors can be found to scale the dimensions of time, deflection, acceleration, force, and moment. Full derivation of the scale factors shown in equations (9)-(13) can be found in the literature (Eppinger 1976).

\[ Time \text{ Scale Factor} = \lambda^\frac{1}{3} \]  
\[ Deflection \text{ Scale Factor} = \lambda^\frac{1}{3} \]  
\[ Acceleration \text{ Scale Factor} = \lambda^{-\frac{1}{3}} \]  
\[ Force \text{ Scale Factor} = \lambda^\frac{2}{3} \]  
\[ Moment \text{ Scale Factor} = \lambda \]

**Impulse Momentum**

Next, three variations of impulse momentum were evaluated (Mertz 1984). This procedure differs from the ESEV1 technique in that it scales human response data based on both a mass ratio and a stiffness ratio.

\[ m_{target} = \lambda_m m_{ref} \]  
\[ k_{target} = \lambda_k k_{ref} \]

Where \( m \) refers to the model mass and \( k \) refers to stiffness. Unlike the ESEV1 technique, which uses a simple ratio of total body mass to develop the scale factors, the impulse momentum procedure accommodates specific body region characteristics and the type of impact testing for determining normalizing factors. This is accomplished by calculating the effective mass of the impacted body region using response data and an impulse momentum analysis. The methods presented by Mertz et al. describe the impulse momentum technique in terms of a one degree-of-freedom system (Equation (16)).
approach was used for the rigid wall simulations used in this study. However, Viano et al. extended the analysis to experiments using a two-degree-of-freedom mass-spring model (Viano 1989). This is applicable to isolated impacts such as a pendulum with a finite mass impacting a specific region of the human body. As such, the pendulum impacts evaluated in this study were scaled using the two degree-of-freedom approach. The impulse momentum analysis for a two body impact is used to establish the effective mass of an impacted body region (Equation (17)).

\[
M_{eff-1DOF} = \frac{\int_{t_0}^{t} F \, dt}{V} = \frac{I_i}{V}
\]

(16)

\[
M_{eff-2DOF} = \frac{\int_{t_0}^{t} M_{pend}A_{pend} \, dt}{\int_{t_0}^{t} A_{model-i} \, dt} = \frac{I_i}{V}
\]

(17)

Where \( F \) is the force measured on the impacting plate, \( M_{eff} \) is the effective mass of the respective model, \( M_{pend} \) is the mass of the pendulum, \( A_{pend} \) is the acceleration of the pendulum, \( A_{model-i} \) is the body region acceleration of the respective model, \( I_i \) is the impulse of the respective model, and \( V \) is the change in velocity during the impact. The effective mass ratio \( (R_m) \) can then be calculated as the ratio of the target effective mass to the reference effective mass:

\[
R_m = \frac{M_{eff-target}}{M_{eff-ref}}
\]

(18)

By assuming geometric similitude and an identical elastic modulus between the two models, the standard stiffness ratio \( (R_k) \) can be considered a ratio of a characteristic length of the body region being impacted. An example of this would be chest depth in a frontal thoracic impact.
\[ R_k = \frac{L_{\text{target}}}{L_{\text{ref}}} \]  

(19)

Where L is the characteristic length. Following the development of these ratios, scale factors can be derived for time, deflection, acceleration, and force. However, it is important to note that the scale factors are different for one degree-of-freedom impacts and two degree-of-freedom impacts. They are made by implementing the equations of motion for either a one or two degree-of-freedom spring mass system. Following the derivation, this approach yields the scale factors shown in Equations (20)-(23) for one degree-of-freedom impacts and Equations (24)-(27) for two degree-of-freedom impacts (Yoganand et al. 2014).

Time Scale Factor = \[ \sqrt{\frac{R_m}{R_k}} \]  

(20)

Deflection Scale Factor = \[ \sqrt{\frac{R_m}{R_k}} \]  

(21)

Acceleration Scale Factor = \[ \sqrt{\frac{R_k}{R_m}} \]  

(22)

Force Scale Factor = \[ \sqrt{R_m \cdot R_k} \]  

(23)

Time Scale Factor = \[ \sqrt{\frac{R_m}{R_k}} \cdot \frac{M_{\text{pend}} + M_{\text{eff-ref}}}{M_{\text{pend}} + M_{\text{eff-target}}} \]  

(24)

Deflection Scale Factor = \[ \sqrt{\frac{R_m}{R_k}} \cdot \frac{M_{\text{pend}} + M_{\text{eff-ref}}}{M_{\text{pend}} + M_{\text{eff-target}}} \]  

(25)

Acceleration Scale Factor = \[ \sqrt{\frac{R_k}{R_m}} \cdot \frac{M_{\text{pend}} + M_{\text{eff-ref}}}{M_{\text{pend}} + M_{\text{eff-target}}} \]  

(26)

Force Scale Factor = \[ \sqrt{R_m \cdot R_k \cdot \frac{M_{\text{pend}} + M_{\text{eff-ref}}}{M_{\text{pend}} + M_{\text{eff-target}}}} \]  

(27)
Hereafter, this technique will be referred to as IM-CharLength\(_2\). This naming refers to the characteristic length approach used to calculate the stiffness ratio. However, in some experimental cases, the length parameter may be absent and cannot be located in an anthropometry table. By extending the assumptions of geometric similitude to the full body scale, it is possible to estimate the stiffness ratio by using the cubed root of the effective masses. Otherwise, the methods described above remain the same, but equation (19) is replaced with equation (28). This approach will be referred to as IM-CR\(_3\).

\[
R_k = \frac{\sqrt[3]{M_{\text{eff-target}}}}{\sqrt[3]{M_{\text{eff-ref}}}}
\]  

(28)

While the assumption of characteristic length, either through the linear measurement or the cubed root of the effective mass, is commonly used, Moorhouse recently proposed a technique that replaces the characteristic length estimate with an estimate of the effective stiffness calculated from response data (Moorhouse 2013). As long as deflection data for the impact is recorded alongside the force data, the method can be used to calculate an effective stiffness based on the following spring equations:

\[
\int Fdx = \frac{1}{2} k_{\text{eff}} x_{\text{max}}^2
\]

(29)

\[
k_{\text{eff}} = \frac{2 \int Fdx}{x_{\text{max}}^2}
\]

(30)

Where \(k_{\text{eff}}\) is the effective stiffness, \(F\) is the force during the impact, and \(x\) is the deflection during the impact. By calculating the effective stiffness for both the reference and target, a ratio of the effective stiffness can be calculated. This effective stiffness ratio (\(R_{\text{eff-k}}\)) can then be substituted for \(R_k\) in the impulse momentum equation described above. This approach will hereafter be referred to as IM-EffStiff\(_4\).
\[ R_{\text{eff-k}} = \frac{k_{\text{eff-target}}}{k_{\text{eff-ref}}} \]  \hspace{1cm} (31)

**Equal Stress Equal Velocity with Effective Mass**

While the ESEV\(_1\) technique is useful due to its simplicity and applicability to both one and two degree-of-freedom systems, the fact that the scale factor is based solely on total body mass can be a limitation. The biodiversity of human subjects can lead to confounding factors that are difficult to characterize using this approach. For example, a tall, thin subject may have the same total body mass as a short, overweight subject. In this case, the two subjects may have nearly identical scale factors indicated when in reality the dynamic response to impact can be significantly different. However, by calculating the effective mass from experimental response data in the impulse momentum procedure, some of the physiological variation across specimens can potentially be accounted for. Then, by applying the effective mass approach to the ESEV\(_1\) procedure, a more representative mass ratio can be obtained for deriving scale factors. Therefore, a 5th scaling method was developed that implemented aspects of both the mass normalization and impulse momentum normalization techniques. For this method, the effective masses calculated from impulse momentum were used to develop the mass ratio used in the ESEV\(_1\) analysis (ESEV-EffMass\(_5\)). Following the evaluation of the effective mass ratio (Equation (32)), the ratio (\(\lambda_{\text{EffMass}}\)) is then substituted into equations (9)-(13) to determine the scale factors for each engineering variable.

\[
\lambda_{\text{EffMass}} = \frac{\int_{t_0}^{t} M_{\text{pend}} A_{\text{pend-target}} dt}{\int_{t_0}^{t} A_{\text{target}} dt} = \frac{\int_{t_0}^{t} M_{\text{pend}} A_{\text{pend-ref}} dt}{\int_{t_0}^{t} A_{\text{ref}} dt} = \frac{M_{\text{eff-target}}}{M_{\text{eff-ref}}} \]  \hspace{1cm} (32)
Work-Energy Theorem

The final technique evaluated used the ESEV$_1$ approach with the mass ratio derived from the work-energy theorem. In this technique, the pendulum force and deflection response were used to relate the net work done to the model to the change in kinetic energy within the model. This was done to create a surrogate for the effective mass calculation employed in the impulse momentum procedure.

\[
\frac{\Delta v}{\Delta t} = \frac{F}{m} \quad (33)
\]

\[
\Delta v \frac{\Delta x}{\Delta t} = \frac{F}{m} \Delta x \quad (34)
\]

\[
\frac{1}{2}mv_f^2 - \frac{1}{2}mv_i^2 = F\Delta x \quad (35)
\]

\[
\Delta \left( \frac{1}{2}mv^2 \right) = \int Fdx \quad (36)
\]

\[
m_i = \frac{2 \int F_i dx_i}{\Delta v^2} \quad (37)
\]

\[
\lambda_{KE} = \frac{2\int F_{target} dx_{target}}{\Delta v^2} \quad \frac{M_{KE-target}}{M_{KE-ref}} = \frac{M_{KE-target}}{M_{KE-ref}} \quad (38)
\]

Where \( v \) is the impactor velocity, \( F \) is the impactor force throughout the impact, \( m \) is the mass, \( x \) is the deflection, and \( t \) is the time of the event. Once the ratio, \( \lambda_{KE} \), has been derived, the value was substituted into equations (9)-(13) to determine the scale factors for the engineering variables of interest. This scaling approach was termed ESEV-KE$_6$.

ISO Comparison

Following the data scaling, the results from each of the techniques described above were quantitatively compared to the target time history outputs using the Enhanced Error Assessment of Response Time Histories (EEARTH) method. EEARTH is part of the
ISO/TS 18571 technical standard for the quantitative assessment of dynamic data. The output of ISO is a score ranging from 0 to 1, where 0 is a poor fit and 1 is an identical match. The standard employs a weighted average of four metrics (corridor, phase, magnitude, and slope) to derive a total to compare the fit of data to a reference curve (Zhan et al. 2011). While the ISO evaluation does have a corridor component, in the current application, the corridor assessment was not applied because only a single scaled curve is being compared to the reference curve. Therefore, for the purposes of this study, only the phase, magnitude, and slope components in the EEARTH portion of ISO were calculated (Davis et al. 2015).

VI. RESULTS

All models completed the simulations without numerical error. To illustrate the impacts evaluated in this study, a time lapse of each simulation can be seen in Figure D 1. In this figure, all simulations shown were performed with the unscaled M50 HBM. As an example for the typical result after applying the scaling techniques to a two degree-of-freedom impact, the scaled results of the frontal chest impact can be viewed graphically in Figure 25. Time history results for the frontal chest impact can be found in the appendix (Figure D 2 and Figure D 3). Scaled results for the frontal abdomen impact and the lateral pelvis impact can be found in the appendix in Figure D 4 and Figure D 9. With regards to rigid wall impacts, the torso and pelvis response for the drop test can be seen in Figure 26 and Figure 27 respectively. Note that for the rigid wall impacts, the IM-CR3 one degree-of-freedom scale factors become equivalent to ESEV-EffMass5. Therefore, in the each of the rigid wall impacts, the traces for IM-CR3 and ESEV-
EffMass are identical. Time history plots for the drop test and scaled responses for the lateral sled impacts can be found in the appendix in Figure D 10 through Figure D 19.

Figure 25. Force-deflection results for the frontal chest impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure 26. Torso force-deflection results for the drop test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff for F05 scaled to M50. IM-CharLength had nearly identical scale factors as IM-CR for M95 scaled to M50.
Figure 27. Pelvis force-deflection results for the drop test. In each case, the magenta line is scaled to the target blue line.
The average EEARTH results for each scaling technique applied to the rigid hub impactors with experimentally defined impactors can be seen in Figure 28. In the figure, the EEARTH scores for deflection are on top and the scores for force are on the bottom. As a baseline, the horizontal line in each dataset represents the unscaled EEARTH score. This unscaled EEARTH score was derived by comparing the raw outputs of the respective models. Overall, the ESEV-EffMass$_5$ technique had the highest average EEARTH scores with a score of 0.94 across all simulations. For these rigid impacts, the unmodified ESEV$_1$, IM-CharLength$_3$, and ESEV-KE$_6$ approaches had the lowest EEARTH scores with average scores of 0.89, 0.88, and 0.85 respectively. In each case, these scores were lower than the original baseline measure of unscaled outputs. In terms of the impulse momentum variations, the IM-EffStiff$_4$ technique improved the response of the signal with an average EEARTH score of 0.93. With regards to specific traces, ESEV-EffMass$_5$ had the highest scores for deflection with an average EEARTH score of 0.96. For force, however, the IM-EffStiff$_4$ and ESEV-EffMass$_5$ had the highest average EEARTH scores (0.92 and 0.91 respectively). These results follow the trends exhibited.
when observing the ability of the scaling techniques to match peak deflection and peak force of the target. Overall, for the experimentally defined rigid impacts, the ESEV-EffMass₅ most closely matched the peak deflection of the target with an average percent error of 3.9%. With regards to peak force, IM-EffStiff₄ and IM-CR₆ had the lowest percent errors with averages of 4.82% and 4.87% respectively. A summary of average percent peak errors for each scaling methodology can be found in Table 12.

The average EEARTH results for each scaling technique applied to the rigid hub impactors with *volumetrically scaled* impactors can be seen in Figure 29. The scaled results from these impacts followed a similar trend as to the simulations with unmodified impactors. The ESEV-EffMass₅ once again had the highest average EEARTH score for the deflection-time traces (0.97) and again most closely matched target peak deflections with an average percent error of 2.17%. However, in this case, the IM-EffStiff₄ had the highest scores for force-time (0.93) and similarly was the closest match in terms of average peak force (3.43%). When force and deflection scores are averaged, the highest overall score was 0.93 for both ESEV-EffMass₅ and IM-EffStiff₄. A summary of each technique for both experimentally defined and volumetrically scaled impactors can be seen in Table 13.
Table 12. Average percent peak errors (absolute values used) for deflection and force. Each score is in reference to the target curve of the scaling operation. All scores are presented as a percent (%).

<table>
<thead>
<tr>
<th>Impactor Type</th>
<th>Signal</th>
<th>Unscaled</th>
<th>ESEV</th>
<th>ESEV-EffMass</th>
<th>IM-CharL</th>
<th>IM-CR</th>
<th>IM-EffStiff</th>
<th>ESEV-KE</th>
</tr>
</thead>
<tbody>
<tr>
<td>2 Degrees of Freedom</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Experimentally Defined</td>
<td>Defl</td>
<td>9.60</td>
<td>6.05</td>
<td>3.90</td>
<td>11.02</td>
<td>8.63</td>
<td>5.98</td>
<td>8.92</td>
</tr>
<tr>
<td></td>
<td>Force</td>
<td>8.44</td>
<td>25.92</td>
<td>12.32</td>
<td>6.60</td>
<td>4.87</td>
<td>4.82</td>
<td>33.09</td>
</tr>
<tr>
<td>Scaled</td>
<td>Defl</td>
<td>10.38</td>
<td>5.21</td>
<td>2.17</td>
<td>12.63</td>
<td>9.45</td>
<td>7.49</td>
<td>9.58</td>
</tr>
<tr>
<td></td>
<td>Force</td>
<td>9.39</td>
<td>24.46</td>
<td>10.74</td>
<td>5.59</td>
<td>4.01</td>
<td>3.43</td>
<td>35.07</td>
</tr>
<tr>
<td></td>
<td>Force</td>
<td>41.89</td>
<td>17.49</td>
<td>10.24</td>
<td>11.75</td>
<td>10.24</td>
<td>12.06</td>
<td>10.23</td>
</tr>
<tr>
<td>Drop Test</td>
<td>Defl</td>
<td>28.41</td>
<td>15.21</td>
<td>11.47</td>
<td>9.93</td>
<td>11.47</td>
<td>12.69</td>
<td>5.00</td>
</tr>
<tr>
<td></td>
<td>Force</td>
<td>42.90</td>
<td>18.95</td>
<td>11.89</td>
<td>13.35</td>
<td>11.89</td>
<td>10.77</td>
<td>5.83</td>
</tr>
</tbody>
</table>

Table 13. Average EEARTH scores for each scaling technique for both experimentally defined and volumetrically scaled impactors.

<table>
<thead>
<tr>
<th></th>
<th>Average EEARTH Score</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Deflection</td>
</tr>
<tr>
<td></td>
<td>Experimentally Defined</td>
</tr>
<tr>
<td>Unscaled</td>
<td>0.90</td>
</tr>
<tr>
<td>ESEV</td>
<td>0.93</td>
</tr>
<tr>
<td>ESEV-EffMass</td>
<td>0.96</td>
</tr>
<tr>
<td>ESEV-CharLength</td>
<td>0.88</td>
</tr>
<tr>
<td>IM-CR</td>
<td>0.91</td>
</tr>
<tr>
<td>IM-EffStiff</td>
<td>0.94</td>
</tr>
<tr>
<td>ESEV-KE</td>
<td>0.90</td>
</tr>
</tbody>
</table>

A different trend emerged when analyzing the one degree-of-freedom simulations. Unlike the two-DOF systems, all techniques improved the correlation. In each of these cases, there was less differentiation between techniques than in the isolated impacts. In the lateral drop test, the ESEV₁ (0.88), ESEV-EffMass₅ (0.87), IM-CR₃ (0.87), and IM-EffStiff₄ (0.87) techniques had similar average EEARTH scores. A similar trend was seen for the lateral sled impacts, with the ESEV₁ having the highest average score (0.87) and the ESEV-EffMass₅ and IM-CR₃ techniques each averaging scores of 0.86. As the scores indicate, there was less qualitative difference between the techniques when
comparing peak deflections and forces. However, as can be seen in Table 12, the ESEV-KE technique was found to have the lowest percent error for matching target peak deflection and force. Detailed size, shape, and phase data for each simulation with all scaling directions in both deflection and force can be seen in Appendix D, Table D 2 - Table D 11.

![Graph showing comparison scores for deflection and force in different body areas.](image)

**Figure 30.** Average deflection and force EEARTH scores for both one degree-of-freedom impacts

**VI. DISCUSSION**

This study presents a methodology to evaluate the performance of various scaling techniques in an idealized setting. By using one HBM volumetrically scaled to different body masses, the primary underlying assumptions of equal elastic modulus and geometric similitude are upheld. Also, by volumetrically scaling impactor size while maintaining input energy, the dynamic loading environments uphold the assumptions of the techniques. These simplifications remove the confounding factors that would be present
when conducting this type of evaluation on a set of cadaver data or in complex loading environments. However, simulations were also performed with unmodified impactors. These were conducted to more closely match a real world experimental setting where only one impactor size may be used. Additionally, by impacting full body simulations into what are essentially infinite masses (rigid wall drop test and lateral sled into steel plates), the scaling techniques of interest were evaluated in a variety of loading conditions and directions.

When isolating the overall average EEARTH score for deflection, the ESEV-EffMass5 obtained the highest score (0.92). From a practical standpoint, the ESEV-EffMass5 technique also had the lowest overall percent error when comparing peak deflections to the target curves (7.34%). When looking at force-time histories however, several techniques performed well (IM-EffStiff4 = 0.88, IM-CR3 = 0.88, and ESEV-EffMass5 = 0.87). However, when looking at the overall ability to match peak forces, the IM-EffStiff4 and IM-CR3 techniques had the lowest percent errors (7.04% and 7.09% respectively). In terms of total EEARTH score, the overall highest average score when combining force and deflection were the ESEV-EffMass5 and IM-EffStiff4 techniques (0.89). These techniques have the advantage of taking regional response data into account, thereby making the techniques more suited for application to subjects of differing body habitus. However, the ESEV-EffMass5 technique combines the simplicity of the ESEV1 method while avoiding the potential limiting factor of a full body mass ratio. Also, it avoids the potential for an over constraint of the scaled result that is possible when both force and deflection response data are used (as in the IM-EffStiff4 and
ESEV-KE$_6$ techniques). A summary of total average EEARTH scores can be seen in Table 14.

<table>
<thead>
<tr>
<th></th>
<th>Total Average Deflection</th>
<th>Total Average Force</th>
<th>Total Average EEARTH Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Unscaled</td>
<td>0.80</td>
<td>0.82</td>
<td>0.81</td>
</tr>
<tr>
<td>ESEV</td>
<td>0.91</td>
<td>0.85</td>
<td>0.88</td>
</tr>
<tr>
<td>ESEV-EffMass</td>
<td>0.92</td>
<td>0.87</td>
<td>0.89</td>
</tr>
<tr>
<td>IM-CharLength</td>
<td>0.87</td>
<td>0.86</td>
<td>0.86</td>
</tr>
<tr>
<td>IM-CR</td>
<td>0.89</td>
<td>0.88</td>
<td>0.88</td>
</tr>
<tr>
<td>IM-EffStiff</td>
<td>0.90</td>
<td>0.88</td>
<td>0.89</td>
</tr>
<tr>
<td>ESEV-KE</td>
<td>0.84</td>
<td>0.76</td>
<td>0.80</td>
</tr>
</tbody>
</table>

While EEARTH scores provide a valuable quantitative synopsis of how scaling techniques compare to the target data by analyzing various characteristics of the signal, it can be difficult to determine if there are significant differences in techniques based on EEARTH scores alone. Therefore, the outputs from EEARTH must also be paired with engineering judgment, other metrics and practical considerations. These include the ability to match peaks, which are often used by biomechanists as a correlate for the overall response or injury risk, and what experimental data is available to calculate the factors used in the techniques. For example, with regards to pendulum type impacts, the ESEV$_1$, IM-CharLength$_2$, and ESEV-KE$_6$ techniques were indicated to be the worst performers as per EEARTH outputs, with average EEARTH scores for multiple simulations that were actually lower than the unscaled data. With regards to IM-CharLength$_2$, there were several cases where the technique scaled deflection in the opposite direction from the target. In the case of ESEV$_1$ and ESEV-KE$_6$, the scale factors for 2 degree-of-freedom systems often led to overestimation of peak forces compared to target with average percent errors of 25.19% and 34.08% respectively. In terms of the
ESEV₁ technique, this result agrees with findings from Moorhouse, where the ESEV₁ methods actually widened response corridors when applied to isolated tests such as the rigid impacts observed in this study (Moorhouse 2013). Because this runs counter to the purpose of scaling in the first place, the findings indicate against their use for two-DOF systems. This type of analysis is paramount in addition to the objective evaluation as a way to assess the practical implications of the results.

Further, in physical experiments of PMHS, the controls inherent to this study of volumetrically scaled test subjects would not be in place. Therefore, a limitation of the traditional ESEV₁ technique is that two subjects that have the same mass but significantly different body morphology would have a scale factor of one, despite the fact that they may display differences in response data. This limitation is exacerbated in isolated tests, where these differences in morphology can greatly affect the response of a localized impact. However, when looking at one-DOF impacts into a rigid or fixed plate, the use of a full body mass ratio for scaling was found to perform quite well as seen from the results described above. While there was less differentiation between methods for the one degree of freedom impacts, the traditional ESEV₁ did obtain the overall highest average EEARTH score, indicating that it is more appropriate for these infinite mass type impacts than localized impacts. However, an analysis of individual components of the EEARTH rating metric indicates that these scores may be skewed as a result of strong phase scores, while overall magnitude scores (and peak comparisons) tended to be lower than other techniques (Table D 8 - Table D 11).

When looking at individual components of the EEARTH score, the ESEV-EffMass₅ and ESEV-KE₆ techniques had the highest average magnitude scores (0.93) for the
deflection traces. For force-time histories, the IM-CR\textsubscript{3} and IM-EffStiff\textsubscript{4} had the highest average EEARTH scores (0.95). For both deflection and force time history signals, these results agree with the evaluation of percent peak errors reported above. These component level evaluations of the EEARTH analysis are important because they can elucidate how the techniques are performing with regards to each aspect of the curve rather than an aggregate total. This is especially true in cases like the rigid wall impacts, where large differences in peak force and deflection between the reference and target can lead to large scale factors when using techniques based on the effective mass or kinetic energy. As can qualitatively be seen in Figure 26 and Figure 27, ESEV-KE\textsubscript{6} performs well when only considering structural response, i.e. a force-deflection response with no time aspect. However, because of the magnitude of the scale factor required to match the peak values of deflection and force, the time axis was significantly affected as well which led to lower phase scores in some circumstances. Since this phase score is given equal weight in the overall EEARTH score, this can skew the results. The time history traces in the appendix can be referenced for further examples of how scaling techniques can affect signal phasing. From a practical perspective, there are cases, such as an evaluation of model viscoelasticity, where phase is an important aspect of model response and should be weighted equally in the analysis. However, in certain test conditions, the overall stiffness or damping response of the model may be more important, indicating that techniques which perform best in matching peak deflection and force should be implemented, and timing is of secondary importance. Therefore, it is important to evaluate the effects of this phasing and determine if structural response may be more important than the scaled timing of the event.
The impacts used in this study were selected because they represented an array of loading conditions, both frontal and lateral, to multiple bodies regions and 1-DOF and 2-DOF systems. In each case, the dynamic environment was modeled to represent a controlled impact where the only differences across simulations were the masses of the HBMs. Therefore, confounding factors such as input energy and timing of contact that may be present in loading environments like a vehicle interior were avoided. Based on the results of these simulations, several conclusions can be drawn. Firstly, the results indicated that one single method may not be appropriate for all simulations. Secondly, the ESEV-EffMass\textsubscript{5} achieved the highest scores when scaling both one and two degree-of-freedom impacts. However, techniques such as the IM-EffStiff\textsubscript{4} also performed well and showed versatility across multiple impact conditions. Thirdly, ESEV\textsubscript{1} was found to perform poorly in isolated (two degree of freedom) impacts, actually lowering the correlation, and should be substituted for one of the other tested methods. It was however on par with other methods in full body (one degree of freedom) impacts.

The scaling techniques used in this study were evaluated in a controlled setting where each of the inherent assumptions of equal modulus and geometric similitude were upheld. While the model masses evaluated in this study effectively push the limits of these techniques, the target masses were selected to represent ranges existing in the literature. In an experimental setting, biodiversity within a given PMHS dataset does not allow the idealized subject pool present in this study. Therefore, while these techniques are optimally applied to subjects with a relatively similar body habitus, researchers often apply these techniques to develop corridors from a wide variety of specimens. In this study, confounding factors such as age and gender related changes in modulus and body
mass distribution were not considered and are reserved for future work. These techniques were not evaluated to address biomechanical response differences from an adult specimen to a child specimen, for example.

Also, while scaling in this study has been focused on deflection and force time-histories, another important characteristic to consider when scaling full body impacts is kinematics. For example, the appropriate scaling of head translation may be of interest when comparing models of differing morphology in a frontal crash scenario. In the future, additional simulations will be performed to evaluate the scaling techniques in more complex loading conditions, such as vehicle sleds, where these types of variables can be explored. Also, while the use of volumetrically scaled models enables the techniques presented in this study to be observed in a controlled manner, impacts in the real world occur to a diverse population. Therefore, the assumptions of geometric similitude could be relaxed to see how these techniques perform when there are morphological variations between the subjects.

**FUNDING**

The authors would like to acknowledge the Global Human Body Models Consortium, LLC, for funding and support. All simulations were run on the DEAC cluster at Wake Forest University, with support provided by Drs. Damian Valles and Timothy Miller.

Dr. Gayzik is a Member of Elemance, LLC, which provides academic and commercial licenses of the GHBMC-owned human body computer models.
VI. APPENDIX D

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Figure D 1. Simulation time-lapse of the M50 for each impact condition
Figure D 2. Force-time results for the frontal chest impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 3. Deflection-time results for the frontal chest impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 4. Force-deflection results for the frontal abdomen impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 5. Force-time results for the frontal abdomen impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 6. Deflection-time results for the frontal abdomen impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 7. Force-deflection results for the lateral pelvis impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 8. Force-time results for the lateral pelvis impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 9. Deflection-time results for the lateral pelvis impact with impactor volumetrically scaled with the model. In each case, the magenta line is scaled to the target blue line.
Figure D 10. Torso force-time results for the drop test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff for F05 scaled to M50. IM-CharLength had nearly identical scale factors as IM-CR for M95 scaled to M50.
Figure D 11. Torso deflection-time results for the drop test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff for F05 scaled to M50. IM-CharLength had nearly identical scale factors as IM-CR for M95 scaled to M50.
Figure D 12. Pelvis force-time results for the drop test. In each case, the magenta line is scaled to the target blue line.
Figure D 13. Pelvis deflection-time results for the drop test. In each case, the magenta line is scaled to the target blue line.
Figure D 14. Torso force-deflection results for the lateral sled test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff for F05 scaled to M50.
Figure D 15. Torso force-time results for the lateral sled test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff for F05 scaled to M50.
Figure D 16. Torso deflection-time results for the lateral sled test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff for F05 scaled to M50.
Figure D 17. Pelvis force-deflection results for the lateral sled test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff and IM-CR*
Figure D 18. Pelvis force-time results for the lateral sled test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff and IM-CR
Figure D 19. Pelvis deflection-time results for the lateral sled test. In each case, the magenta line is scaled to the target blue line.

*Note: IM-CharLength had nearly identical scale factors as IM-EffStiff and IM-CR*
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Table D 11. Drop Test Pelvis Results

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VI. REFERENCES


Chapter VII: Comparison of Objective Rating Techniques vs. Expert Opinion in the Validation of Human Body Surrogates

Matthew L. Davis 1,2, Bharath Koya 1,2, Jeremy M. Schap 1,2, Fang-Chi Hsu 2, F. Scott Gayzik 1,2

1Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
2 Wake Forest School of Medicine, Winston-Salem, NC

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VII. ABSTRACT

Objective evaluation (OE) methods provide quantitative insight into how well human body models (HBMs) predict a biomechanical response. Two techniques for this purpose are CORA and the ISO/TS 18571 standard. These ostensibly objective techniques have differences in their algorithms that may lead to discrepancies when interpreting model performance. The objectives of this study were 1) to apply both techniques to a biomechanical dataset from a HBM, and compare the scores and 2) conduct a survey of subject matter experts (SMEs) to determine which OE method compares more consistently with SME interpretation. The GHBMC average male HBM was used in five simulations of biomechanics experiments, producing 58 time history curves. Because both techniques produce phase, magnitude, and shape scores, 174 pairwise comparisons were made. ISO had lower average scores for each component rating metric than CORA, indicating a stricter evaluation. Correlations between CORA and ISO were strongest for phase ($R^2=0.66$) and weakest for shape ($R^2=0.27$). Statistical analysis revealed significant differences between the two OE methods for each component rating metric. SMEs (n=40) were then surveyed to provide intuitive scoring of how well the computational traces matched the experiments. SME interpretation was found to statistically agree with the ISO shape and phase metrics, but was significantly different than the ISO magnitude rating. SME interpretation agreed with the CORA magnitude rating. The finding of the study suggests a mixed approach to reporting objective ratings, using the magnitude method in CORA and the ISO shape and phase methods.
VII. INTRODUCTION

The use of computational modeling has become an important aspect of the development process in the automotive and defense industries. Prior to production, products are often tested using a variety of computer programs to evaluate their performance. These simulations evaluate aspects of the design process ranging from structural crash-worthiness (Tabiei and Wu 2000) to occupant protection and injury risk mitigation (Davis et al. 2015). A growing component of these types of analyses includes the use of computational human body surrogates. These simulations can include a variety of models, including rigid body models (Untaroiu et al. 2010), anthropomorphic test devices (ATDs) (Gaewsky et al. 2015), or full human body models (HBMs) (Davis et al. 2015; Vavalle et al. 2013). Due to the reduced cost of running these simulations, as well as the large amount data that can be extracted from them, these types of simulations offer a valuable supplement to physical testing. However, in order for these models to yield meaningful data, they must be carefully validated. Typically, models are rigorously validated against experimental data prior to implementation. How closely a model matches an experiment is a key piece of information for modelers. For example, HBMs are commonly compared against mean response and corridor biomechanical data obtained from Post-Mortem Human Subject (PMHS) testing (Vavalle 2012). For the sake of validation, a quantitative comparison that leads to an unambiguous interpretation of the model performance taking into account the biological variation of specimens is paramount when characterizing the biofidelity of a model. These objective comparisons offer a robust means of evaluating the performance of a model throughout the course of development.
Objective Evaluation (OE) methods seek to replace the subjectivity inherent in the validation process with a numerical score that provides quantitative insight into how well a human surrogate predicts a biomechanical response. While there are many techniques for this purpose (Rhule et al. 2002; Sprague and Geers 2004; Vavalle et al. 2013), two commonly applied methods are Gehre et al.’s CORA software (Gehre et al. 2009) and the ISO/TS 18571 standard (Barbat et al. 2013). The advantage of these techniques is that their algorithms evaluate individual components of the curve to provide a more complete comparison of time-history signals. While both techniques evaluate similar aspects of the signals, there are several differences between the inherent algorithms of the methods that can lead to different interpretations of the results. It is important to understand how these differences can be interpreted and the effect they can have on model validation. While these techniques are broadly used (Davis et al. 2015; Gehre and Stahlschmidt 2011; Poulard et al. 2015), they have not been directly compared.

As such, the objectives of this study are two-fold: 1) compare the results of CORA and ISO/TS 18571 OE techniques applied to a set of biomechanical data derived from a human body finite element model, and 2) conduct a survey of subject matter experts (SMEs) to determine which of these OE methods, if either, compares more consistently with SME interpretation. The goal of this work is to evaluate how results from these techniques can influence the interpretation of model validity, and if these interpretations agree with real world expert interpretation.

**VII. METHODS**

The Global Human Body Models Consortium (GHBMC) average male occupant (M50-O v4.4) finite element model was selected for use in this study. The model was
developed based on a multi-modality medical image and external anthropometry dataset of a volunteer representing a 50\textsuperscript{th} percentile male in terms of height (174.9 cm) and weight (78.6 ± 0.77 kg). The development and application of this dataset was described by Gayzik et al. (Gayzik et al. 2011). Once developed, the model underwent validation simulations at both the regional (DeWit and Cronin 2012; Li et al. 2010; Shin et al. 2012; Soni and Beillas 2015) and full body levels (Hayes et al. 2014; Toyota 2010; Yang et al. 2006). More information on the development of the model can be found in the GHBMC M50-O user’s manual (GHBMC 2011).

**Simulations**

To obtain outputs representing a range of impact conditions and directions, the model was run through five simulations representing physical biomechanics experiments. These simulations consisted of both localized, rigid hub impacts and full body sled cases. The rigid hub simulations included an oblique thoracoabdominal impact (Viano 1989), a frontal abdominal impact (Hardy et al. 2001), and a lateral pelvis impact (Bouquet 1998). The full body sled cases represented a lateral impact into fixed steel plates (Cavanaugh et al. 1990; Cavanaugh et al. 1993) and a frontal sled test configuration (Shaw et al. 2009). All simulations were run using LS-Dyna v6.1.1, rev. 78769 on a Linux Red Hat 6 high performance computing system (the Distributed Environment for Academic Computing, or DEAC cluster) maintained at Wake Forest University.

The thoracoabdominal impact employed a 23.4 kg cylindrical hub impactor with a 15 cm diameter and a nominal impact velocity of 6.7 m/s (Viano 1989). The impact location was 7.5 cm below the xipohoid process at 60° from anterior. Model data were
compared to the mean experimental force vs. time signal in order to evaluate the OE techniques.

The abdominal impact consisted of a 2.5 cm diameter, 48 kg bar impacting at 6.0 m/s. This was a free-back impact occurring at the level of the umbilicus (approximately L3) (Hardy et al. 2001). The force of the impact was measured as the contact force of the rigid bar. Model data were compared to the mean experimental force vs. time curve.

The pelvic impact simulated a square-faced impactor weighing 16 kg impacting with 800 J of energy. This required giving the impactor a 10 m/s velocity normal to the sagittal plane. The pelvis impactor contacted the trochanter and iliac crest at 90°, according to the literature (Bouquet 1998). Similar to the other rigid impact simulations, the contact force of the impacting plate was used to obtain force data. This contact force was compared to the mean experimental force vs. time curve to facilitate OE technique comparison.

The lateral sled test was modeled as a 6.7 m/s impact using a Heidelberg-type sled (Cavanaugh et al. 1990; Cavanaugh et al. 1993). The impact environment included a flat rigid wall as a backrest, a Teflon seat, and five rigid impacting plates located at the shoulder, thorax, abdomen, pelvis, and knee. Torso forces were obtained as the sum of the shoulder, thorax, and abdomen forces. The pelvis force was measured as the contacting force at the pelvis plate. For both the torso and pelvis outputs, the model responses were compared to the mean experimental force vs. time data.

The frontal sled case was modeled as per Shaw et al. (Shaw et al. 2009). This simulation represented a frontal impact with an overall change in velocity of 40 kph. The simplified buck used in the simulation was modeled as a rigid body. Belt properties were
developed to match experimental conditions (26 kN of force at 7% strain) and no pretensioners or load-limiters were included. A foam knee bolster was also included to restrict motion of the lower extremities in the model. Prior to simulation, the model was gravity settled for 100 ms to obtain realistic flesh contours within the buck. With regards to outputs, both kinetic and kinematic responses were obtained for comparison to experimental values. With the exception of chest deflection data, all kinematics were reported in the global coordinate system. Reaction forces at the knee bolster and foot rests were also recorded in the global coordinate system and then transformed into a local coordinate system per the literature (Ash et al. 2012). Resultant belt force data were obtained to represent the responses of the upper and lower shoulder belt and the outer lap belt. All data extracted from the model were compared against the average of experimental PMHS tests (Shaw et al. 2009).

**Objective Evaluation**

While the model validity and accurate representation of the described biomechanical simulations is paramount, the goal of this study is to see how, when presented with identical comparison cases, the CORA and ISO techniques interpret model performance. To facilitate this comparison, all model data were output in binary files from LS-Dyna and were recorded at a sampling rate of 10 kHz. Post-processing of the data was performed in OASYS T-His (Ove Arup SYStems, Solihull, UK) and Matlab R2013 (MathWorks, Natick, MA). Force data were filtered using an SAE CFC 600 filter and kinematic data were not filtered.

In order to effectively source discrepancies between the two OE techniques, it is important to understand how each component of the rating metric is calculated. Detailed
descriptions, including the inherent algorithms, of each technique can be found in the literature ((ISO) 2013; Thunert 2012). However, as a foundation for comparison, each component of the CORA and ISO techniques are briefly described.

**CORA Metric**

The CORA rating metric is a set of algorithms comprised of two independent sub-rating schemes: a corridor score and a cross-correlation score (Gehre et al. 2009). A complete description of this technique can be found in the literature (Thunert 2012). The software was developed to calculate the level of correlation between two non-ambiguous signals and return a total score ranging from 0 to 1, where a 1 would display good correlation and a 0 would be a poor match based on defined tolerances. The default settings as recommended by the software provider were used in this analysis, with the exception of the phase interval, which is described below.

The corridor rating is designed to evaluate the deviation between the signals using a set of fixed-width or user-defined (i.e. experimentally reported) inner and outer corridors. If the model curve is within the inner corridor, the resulting score is a 1. If the model curve falls between the inner and outer corridor, the result is between 0 and 1 based on an interpolation score. If the signal is outside of the outer corridor, the result is a 0. While this technique gives a valuable global picture of model performance, a disadvantage of this approach is that phase differences between the model signal and the experimental data can lead to poor scores.

The cross correlation method analyzes three aspects of the signal in order to reduce the relative disadvantages of using only the corridor score: phase, shape, and magnitude. First, the algorithm attempts to eliminate differences in phasing by shifting the model
curve by multiples of Δt. Then, for each shifted state, the program calculates a cross-correlation value. The maximum cross-correlation over a user defined range of allowable time shift is then used as a basis for determining the three components of the cross-correlation rating. For calculating the phase rating, if the model signal was shifted less than a user defined minimum, the rating receives a score of 1. If the curve is shifted more than a specified maximum, the score is zero. For phase shifts between the specified minimum and maximum, the score is determined based on a regression relationship. Following the time shift, the magnitude rating is computed by comparing the square of the areas between the curves and the time axis. The final magnitude rating is then determined as a ratio between the two areas raised to a user defined exponent. Lastly, as shape similarities between the two signals are presumed to be inherently related to the maximum cross-correlation, the shape rating of the signal is calculated using the maximum cross-correlation value.

ISO Metric

Similar to CORA, the goal of the ISO metric was to combine a number of different rating metrics to robustly evaluate the correlation between two signals. Initially, the ISO established technical committee evaluated the CORA corridor technique and the Error Assessment of Response Time Histories (EARTH) (Sarin et al. 2010) techniques to combine a corridor and cross correlation rating. Ultimately, the committee established an overall metric based on the CORA corridor algorithm, and an updated version of the EARTH score referred to as the Enhanced EARTH metric (EEARTH) (ISO 2013).

Similar to the CORA cross-correlation metric, the total EEARTH rating is built on the individual phase, magnitude, and shape components. However, while the general
components of the EEARTH metric are similar to CORA, there are unique features within the algorithms that differentiate the two. The phase metric of the EEARTH rating is used to assess phase lag between the model and test curves. Using a pre-defined maximum allowable percentage time-shift, the model curve is iteratively shifted left with discrete time step intervals and the cross-correlation between the truncated curves is calculated. Next, the test curve is shifted left over discrete time step intervals and the same calculation is performed. If the time shift is greater than or equal to the maximum allowable time shift, the score is 0. If the maximum cross-correlation value occurs with no time shift, the score is 1. For time shifts in between these values, the rating is calculated using a regression method (ISO 2013; Barbat et al. 2013). The time shifted and truncated curves are then used to calculate the magnitude score.

Similar to CORA, the EEARTH magnitude rating measures differences in amplitude between the two curves. However, the EEARTH magnitude rating applies an algorithm known as dynamic time warping (DTW) prior to measuring discrepancies between the signals. The function of DTW is to expand and compress the time axis to align key components of the curve (such as local maxima and minima). This is all based on minimizing a local cost function (ISO 2013). Once the curves have been shifted, truncated, and warped, the magnitude error is calculated as a ratio of the difference in amplitude between the two signals based on a vector norm calculation. If the difference between the signals is less than the pre-defined threshold, the magnitude rating is 1. If the amplitude difference is greater than the maximum allowable magnitude error threshold, the score is 0. For values in between, the score is calculated using a regression function.
Lastly, the shape rating is calculated based on the test curve and the shifted, truncated model curve with no DTW applied. The two curves are divided into time intervals and the average slope is calculated at each interval. The shape score is then determined by calculating the ratio of the difference in slope between the model and test curves to the test curve. If there is no difference between the model and test curve, the shape score is 1. If the difference exceeds a pre-define threshold, the score is 0. Values in between are calculated based on a regression function.

**Application of OE Methods**

The kinetic and kinematic time-history traces obtained from the models were run through CORA v3.5 and ISO. When applying CORA, suggested default values were used for all parameter controls except for the phase range. For evaluating the phase shift, the allowable time shift range was changed from 3 to 12 percent to a range of 5 to 15 percent. For the application of ISO, all recommended weights and parameters set forth by the standard were applied. A total of 58 time history traces were obtained from the simulations. This provided a diverse sample of signals allowing for a robust comparison of the techniques. Because both techniques produce a phase, magnitude, and shape score, 174 (58 x 3) pair-wise comparisons were made. The corridor score of each technique was not included in the pair-wise analysis because the underlying algorithm is the same for both CORA and ISO. For each component of the cross-correlation rating metric, the comparative scores were cross-plotted and used to evaluate correlations. In this analysis, the coefficient of determination, $R^2$, was used to highlight differences in the two techniques. Statistical tests for significant differences between the two were also
determined using a Wilcoxon matched-pairs signed rank test and a significance value of $\alpha = 0.05$.

**Survey of Subject Matter Experts**

The survey component of this study was approved by the Wake Forest School of Medicine's Institutional Review Board (IRB #39944). To evaluate how the OE techniques compare to real world interpretation, a survey was distributed to subject matter experts (SMEs) to obtain a scoring of how well the computational traces match the experiments in terms of phase, magnitude and shape. The objective was to compare the SME based scores to the quantitative results obtained from CORA and ISO to determine which, if either, is more in line with SME assessment. Participants were asked to complete a one-time, electronic survey designed to provide their interpretation of how well a subsample of 15 time history traces from the full dataset compared to experimental traces. Participation was limited to individuals with training or expertise in computational modeling and model validation techniques. Participants were contacted for inclusion in the study via email and all responses were anonymized prior to analysis. However, demographic information including work title, affiliation, years of experience in biomechanics, and years of experience in model validation/signal analysis were requested. In order to answer the research question, the survey was sent to 69 SMEs with an expected participation of 50%. The sample size for the survey was determined by calculating the minimum number of survey questions and participants needed to detect a Cronbach’s alpha of 0.9 assuming type 1 error of 0.05, a two-sided test, and a 80% power. Prior to evaluating the curves, participants were introduced to the terminology used in the study to ensure a reasonable baseline. However, no coaching was conducted
in order to ensure that participants were not led to focus on specific curve attributes for evaluation. Each participant was asked to rank the phase, magnitude, and shape for the subsample of 15 curves, presented in a randomized order, on a scale of 0-100 that could be directly mapped to the scale implemented in CORA and ISO. The results were then analyzed using a 1-sample t-test that tested whether the mean from the sample was the same as CORA or ISO independently for phase, shape, or magnitude.

VII. RESULTS

All simulations normally terminated without numerical error. In each case, simulations were visually inspected for localized areas of instability and were found to be stable. To illustrate the impacts evaluated in this study, a time lapse of each simulation can be seen in Figure 31.

The time history signals for each impact condition were exported and compared to the experimental data using both CORA and ISO. Because of the variety of data obtained from these simulations, the OE methods were evaluated using signals that ranged from very good to poor correlation.
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<td>![Image]</td>
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</tr>
</tbody>
</table>

Figure 31. Simulation time-lapse of the M50 for each impact condition
The overall average scores for the two algorithms, including the corridor metrics, were 0.60 and 0.56 for CORA and ISO respectively. No signal-based weighting approach was used for off-axis signals which may produce low correlation scores, but are also low magnitude compared to the resultant (e.g. shear vs. normal loading). Overall scores typically emphasize the dominant signals (Davis et al. 2016; Pietsch et al. 2016). For both techniques, the corridor and cross-correlation scores were computed with the recommended weight factors (see Equations (39) and (40) where \(Z_{\text{corr}}\) stands for corridor score and \(Z_{\text{CrossCor}}\) stands for cross-correlation score). With regards to magnitude, CORA rated the curves with an average score of 0.49 ± 0.27, whereas ISO rated the signals lower in general with an average rating of 0.38 ± 0.36. For phase, the average CORA score was 0.72 ± 0.38 and the average ISO score was 0.69 ± 0.30. Lastly, shape scores were rated as 0.71 ± 0.34 in CORA and 0.61 ± 0.16 in ISO. Therefore, each of the components of the ISO cross-correlation score were lower on average compared to CORA, indicating a stricter rating of the signals.

\[
\begin{align*}
\text{CORA Score} &= 0.5 \times Z_{\text{corr}} + 0.5 \times Z_{\text{CrossCor}} \quad (39) \\
\text{ISO Score} &= 0.4 \times Z_{\text{corr}} + 0.6 \times Z_{\text{CrossCor}} \quad (40)
\end{align*}
\]

Cross-plots for each component metric of CORA and ISO can be seen in Figure 32. In these plots, the respective magnitude, phase, and shape scores were aggregated from each simulation and compared using linear regression. As such, each point on the plot represents the CORA and ISO scores for a single time history trace. The phase scores for each metric were found to have the strongest correlation with an \(R^2\) value of 0.66. Shape scores were found to have the weakest correlation with an \(R^2\) value of 0.27. With regards
to statistical comparison, the differences between CORA and ISO were found to be statistically significant for each component rating metric with p values of 0.003, 0.002, and 0.016 for phase, magnitude, and shape respectively.

Figure 32. Correlation analysis of each component rating metric

Survey Responses

In total, 40 responses were collected from the survey solicitation. Participants were primarily from academia (72%), followed by industry (15%) and government (13%). More than 33% of participants had 10+ years of biomechanics and signal analysis experience, with more than 60% having 5+ years of experience.
The average response for the phase, magnitude, and shape characteristics for each of the 15 curves can be seen in Figure 33. In Figure 33, the bars represent the average of all survey responses for a particular curve and rating metric. Overall, volunteers rated the magnitude scores lowest with an average score of 0.52 across all 15 curves. The magnitude rating also had the largest variation in responses with respect to the average with a coefficient of variation of 0.38, indicating the widest variation in SME assessment. The phase rating was given the highest scores with an average of 0.70. The phase rating also had the lowest average of coefficient of variation (c_v = 0.28), indicating the greatest agreement in SME assessment.

Figure 33. Survey responses for each curve
When comparing the SME responses to the ISO standard, the null hypothesis that the SME responses and ISO results were the same was rejected for magnitude \( (p<0.001) \), but was not rejected for the shape and phase metrics \( (p\)-values of 0.79 and 0.10 respectively). With regards to CORA, the null hypothesis was rejected for the shape and phase metrics \( (p\)-values of 0.005 and <0.001 respectively), but was not rejected for the magnitude rating \( (p = 0.79) \). From a real world perspective, this indicates that SME responses agreed with the ISO interpretation for phase and shape, but did not agree with the ISO magnitude rating. Conversely, the SME responses agreed with the CORA interpretation for magnitude, but not the phase and shape ratings.

For further evaluation, the percent difference in peak value was compared to the average volunteer magnitude rating for each of the 15 curves in the survey. The percent difference in peaks was moderately correlated to the magnitude ratings with a Spearman’s rho of -0.58. The average magnitude ratings were also compared to the percent difference in area under the curve for each of the curves evaluated in the survey. In this case, area under the curve showed strong correlation to the magnitude ratings with a Spearman’s rho of -0.79.

**VII. DISCUSSION**

As objective evaluation techniques are increasingly applied to the validation of computational human surrogate models, it is important to investigate how variations in different rating metrics can affect the overall estimation of model validity. The goal of this study was to apply both the CORA and ISO objective rating metrics to the same set of data derived from simulations of the GHBMC M50-O finite element model. While we strive for objective evaluations, we also note that the totality of how an engineer may
view a signal is probably beyond what can be encapsulated in three nominally orthogonal measures (magnitude, phase and shape). Thus in this work we sought to find which algorithms are more likely to be in agreement with evaluations made by experts in the field. As such, the CORA and ISO interpretations were also compared to real world interpretations from SMEs.

The cross-plots depicted in Figure 32 show general trends for each component metric. For example, phase scores for both ISO and CORA had higher scores on average compared to the magnitude and shape ratings. However, more interesting are cases where one technique assigns a score of nearly 1 to a curve, and the other technique assigns the same curve a score closer to 0. Using these techniques as they were intended, this means the user is to interpret that one technique says the signal is a good match to the experimental data, but the other technique says the model does not represent the real world test.

This indicates that both techniques have limitations, and therefore must be combined with engineering judgment prior to drawing final conclusions regarding a model’s validity. However, as the survey responses show, overall interpretation of the shape and phase response from SMEs tended to agree with the outputs from the ISO technique. For magnitude, the CORA method tended to more closely agree with the SMEs. This indicates that using the area of the signal may be a more intuitive means of assessing magnitude ratings. This finding also agrees with the strong correlation between the SME magnitude ratings and the percent difference in area for the curves in the survey. However, as the CORA magnitude rating uses a squared area ratio to assign a score, differences in polarity between the experimental and model curves can lead to artificially
high scores. Therefore, when comparing signals with flipped polarity, this limitation in the CORA magnitude rating should be addressed in the future by including a polarity correction factor.

Overall, the CORA magnitude rating and ISO phase and shape ratings were found to provide the most intuitive scores when comparing model and experimental curves (Table 15). However, both techniques tend to give higher scores on average to the phase rating compared to the other component rating metrics. This can lead to biased total scores when phase is equally weighted with magnitude and shape. A similar trend was seen in SME interpretations, with an average phase score that was 30% and 17% higher than the subsampled magnitude and shape scores respectively. Therefore, in some applications, it may be necessary for the user to increase the exponent governing phase scores between 0 and 1 to make the regression equation either quadratic or cubic. Also, either isolated or in combination, the interval over which phase shift is permitted could be reduced to more strictly govern the phase rating. In general, this would more strictly govern the overall rating and enable researchers to more easily discern potentially required model updates, such as viscoelastic adjustments.

<table>
<thead>
<tr>
<th>Component Rating Metric</th>
<th>OE Technique in Agreement with SMEs</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Magnitude</td>
<td>CORA</td>
<td>0.79</td>
</tr>
<tr>
<td>Phase</td>
<td>ISO</td>
<td>0.1</td>
</tr>
<tr>
<td>Shape</td>
<td>ISO</td>
<td>0.79</td>
</tr>
</tbody>
</table>

Also, when reporting final quantitative evaluations for validation, researchers commonly report the average of each component rating metric to give a global view of the model response. However, it may be necessary to discriminate between signals of
varying magnitude to give a clearer picture of model behavior. For example, orthogonal signals (responses on the x, y, and z axes) often have motion on a primary axis (ex. X-axis) or primary plane of motion (ex. X-Y plane). In these cases, there are one or two off-axis responses that do not have the same scale as the primary motion. Therefore, in certain applications it may be appropriate to apply a weighting calculation to apply more weight to the scores of plots with greater magnitude. Davis et al. proposed an approach to weight objective evaluations based on a weighting factor derived from the peak values of the experimental mean traces (Davis et al. 2016).

How closely a model matches an experiment is a key piece of information for modelers. The OE methods evaluated in this work seek to replace the subjectivity inherent in this process with a numerical score, yet it is clear from the results that ostensibly objective methods can produce different interpretations for the same data. Future regulatory measures may require a threshold OE score to be met for given model, thus it is imperative to understand how various techniques compare. This study provides a framework to critically compare results from each method, and highlights the relative strengths and weaknesses of each.
VII. REFERENCES


Toyota (2010) Documentation of Total Human Model for Safety (THUMS) AM50 Pedestrian/Occupant Model


Chapter VIII: Development and Full Body Validation of a 5th Percentile Female Finite Element Model

Matthew L. Davis 1,2, Bharath Koya 1,2, Jeremy M. Schap 1,2, F. Scott Gayzik 1,2

1Virginia Tech – Wake Forest University Center for Injury Biomechanics, Winston-Salem, NC
2Wake Forest School of Medicine, Winston-Salem, NC

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VIII. ABSTRACT

To mitigate the societal impact of vehicle crash, researchers are using a variety of tools, including finite element models (FEMs). As part of the Global Human Body Models Consortium (GHBMC) project, comprehensive medical image and anthropometrical data of the 5th percentile female (F05) were acquired for the explicit purpose of FEM development. The F05-O (occupant) FEM model consists of 981 parts, 2.6 million elements, 1.4 million nodes, and has a mass of 51.1 kg. The model was compared to experimental data in 10 validation cases ranging from localized rigid hub impacts to full body sled cases. In order to make direct comparisons to experimental data, which represent the mass of an average male, the model was compared to experimental corridors using two methods: 1) post-hoc scaling the outputs from the baseline F05-O model and 2) geometrically morphing the model to the body habitus of the average male to allow direct comparisons. This second step required running the morphed full body model in all 10 simulations for a total of 20 full body simulations presented. Overall, geometrically morphing the model was found to more closely match the target data with an average ISO score for the rigid impacts of 0.76 compared to 0.67 for the scaled responses. Based on these data, the morphed model was then used for model validation in the vehicle sled cases. Overall, the morphed model attained an average weighted score of 0.69 for the two sled impacts. Hard tissue injuries were also assessed and the baseline F05-O model was found to predict a greater occurrence of pelvic fractures compared to the GHBMC average male model, but predicted fewer rib fractures.
VIII. INTRODUCTION

Vehicular crash injury prevention remains a leading public health concern worldwide. In 2013, the World Health Organization reported more than 1.2 million deaths and another 20-50 million non-fatal injuries as a result of motor vehicle accidents (W.H.O 2013). The injury outcome of vehicular crash also results in significant financial costs. In the United States alone, the National Highway Traffic Safety Administration (NHTSA) estimates the economic and societal costs of vehicle crash at $871 billion (Blincoe et al. 2014). In order to design occupant protection systems that reduce risk of injury, researchers are using a variety of tools, including computational Human Body Models (HBM). The application and development of such models is a growing component of injury biomechanics. The last 20 years has seen a large increase in the number of HBMs being developed at both the full body level (Hayes et al. 2014; Toyota 2010; Yang et al. 2006) and body regional level (DeWit and Cronin 2012; Li et al. 2010; Shin et al. 2012; Soni and Beillas 2015). Computational HBMs, such as finite element models (FEM), are appealing because they offer a cost-effective method to evaluate and design vehicle safety devices (Yoganandan et al. 2014). They are also useful for providing deeper insight into the injury mechanisms of specific tissues during dynamic loading scenarios.

Historically, HBMs are developed to represent an average male (50th percentile in terms of height and weight). There are several reasons for this design paradigm for model development. First, the average male has been used as an indicator for average human response. Therefore, the majority of biomechanics datasets available for model validation implement normalization procedures to mathematically represent the average
male. However, real world motor vehicle crashes involve occupants of various gender, age, and size. As an example, an investigation of fatalities in the 1990’s resulting from airbag related injuries in otherwise minor to moderate severity frontal crashes found that 78% of the fatalities were female, and 82% of the females were below average height (less than 163 cm) (Summers et al. 2001). It was also determined that close proximity to the wheel at the time of airbag deployment was a factor in all of the air bag related fatalities.

Field data has shown that females have a higher risk of sustaining injury during motor vehicle crashes compared to males. For example, Evans et al. found that, for ages 20 to 35, female risk exceeds male risk of suffering a fatal crash in similar severity impacts by 28% (Evans 2001). Similarly, a study of single-occupancy two-car crashes found that women from age 20 to 35 have a 22% higher fatality risk (Evans and Gerrish 2001). Ulfarsson et al. used data from the Master Accident Record System (MARS) to investigate differences between male and female injury severity in single and two-vehicle accidents in a variety of vehicle types, including passenger cars, sport-utility vehicles (SUVs) and minivans. Overall, female drivers were found to have a higher chance of possibly sustaining an injury relative to male drivers (Ulfarsson and Mannering 2004). The aggregated data also suggested the possibility for differences in injury severity between male and female drivers.

There are several potential reasons for the increased injury risk for female drivers. 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 (King and Yang 1995; Manary et al. 1998; Melvin et al. 1993). Schoell et al. also showed the sex related differences in
geometry could contribute to the increased risk. In this work, the authors isolated geometric differences from other confounding factors, such as potential material differences, by morphing the rib cage geometry of the Global Human Body Models Consortium’s (GHBMC) 50th percentile male model (M50) to female specific anatomy based on a population based statistical model (Schoell et al. 2015). Increased injuries may also be attributable to lower structural strength due to lower bone mineral content within females (Duma et al. 1999). From a regulatory perspective, these potential differences in injury risk as a function of body habitus and gender were addressed by including the response of a small female anthropomorphic test device (Hybrid III 5th percentile female) to evaluate the ability of vehicle safety devices to protect a wider range of occupants. In order to further the ability of computational HBMs to provide comparable data, this study focuses on the model development of a female driver in the 5th percentile of height and weight (F05).

To date, several component level and full body models of the F05 have been developed. Since scan data of such a specific target anthropometry are limited, finite element models of the F05 have, until recently, been developed primarily via scaling algorithms applied to existing M50 models. To scale these models, anthropomorphic relationships are typically established using external anthropometry measurements or databases. For example, as part of a study evaluating the effect of head size on intracranial pressures following trauma to the head, Kleiven et al. developed a model representing the F05 head by scaling geometries obtained via the visible human database (National Institute of Health) (Kleiven and von Holst 2002). Van Rooij et al. also described the development of a finite element model of the F05 upper extremity to
evaluate air bag loading. In this case, the authors created a finite element mesh representing the upper extremity of the F05 by scaling data from the European HUman MOdel for Safety (HUMOS) (Robin 2001; Van Rooij et al. 2003). Similar approaches have also been applied to the development of full body F05 models. One of the earliest human body models of the small female was developed by Happee et al (Happee et al. 2000). This multi-body model was developed by scaling lengths and joint forces derived from the RAMSIS anthropometry database. However, to maintain computational efficiency, the model was implemented with flexible bodies rather than finite elements. With regards to F05 human body finite element models, the HUMOS2 F05 was developed by applying scale factors to an M50 model (Serre 2006; Vezin and Verriest 2005). In the case of the HUMOS2, European databases of anthropometry measurements were used to create relationships between external measurements and internal dimensions from a dataset of bi-plane X-ray images. Therefore, the internal dimensions of the model were based on relationships from external measurements. Kimpara et al. developed an early version of a small female thorax by adjusting female geometry available in the View Point Datalabs database (Kimpara et al. 2002). However, this was a computationally efficient model that did not include explicitly represented internal organs. This model was later updated to include soft tissue structures for the investigation of the small female response in omni-directional impacts (Kimpara 2005). In 2003, Iwamoto et al. published on the development of the Toyota Total Human Model for Safety (THUMS) American F05 (AF05) model (Iwamoto et al. 2003). This model was originally produced by applying anthropometric data to scale the THUMS average sized male to the body habitus of the AF05 occupant with thoracic and pelvic regions
implemented to represent female-specific geometry (Schneider et al. 1983). The THUMS AF05 has recently been updated to include explicitly represented internal organs with geometries derived from medical image data (Watanabe et al. 2012). Additionally, recent work by Klein et al. has developed a set of parametric whole body models to account for age, sex, and BMI. As the focus of this work was on evaluating injuries to the lower extremity, parametric FE models of the pelvis, femur, and tibia, as well as external body surface shape models were used to create full body morphs using the THUMS average male model as a baseline (Klein 2015).

While much research has been done in this area, there was a need in the literature for a full body small female FEM that was developed using a comprehensive approach employing data obtained \textit{a priori} from a representative individual in multiple postures. This is an important aspect for the development paradigm as recent studies have found significant differences in abdominal organ position and shape between supine and seated postures (Beillas 2009; Hayes et al. 2013). In addition, recent work by Davis et al. was presented for establishing the range of thoracoabdominal organ volumes in small women that can be used for verification of the internal anatomy (Davis et al. 2015). As part of the Global Human Body Models Consortium (GHBMC) project, a comprehensive multi-modality medical image and anthropometrical dataset of the F05 were acquired. This dataset is unique in that it was acquired in a variety of postures for the specific purpose of FEM development. Previous work has detailed the development of a posture specific CAD geometry model of the F05 occupant (F05-O) based on these datasets (Davis et al. 2014). The geometries in the CAD dataset were derived from subject specific computed tomography (CT) and magnetic resonance imaging (MRI) medical images in the supine,
seated and standing postures. The strengths of each imaging modality were then leveraged to develop F05 anatomy that can be used to characterize relevant crash-induced injuries (CIIs).

The objectives of this study are two-fold. The first is to use the CAD data derived from the F05 dataset to develop a finite element model of the F05-O in the seated posture. The second is focused on full body model validation by simulating biomechanical studies conducted on post mortem human subjects (PMHS). Previously developed scaling techniques are used to compare the model to experimental corridors. However, due to limitations inherent to scaling data from a female body habitus to that of a male, we hypothesize that the most effective technique for validation will be geometrically morphing the F05-O model to the M50 body habitus, rather than applying post-hoc scaling methods to the output data. To date, the data set used for the development of this model is the first of its kind, acquired with the explicit purpose of developing a full-body finite element model of the F05-O for the enhancement of injury prediction.

VIII. METHODS

Medical Imaging Protocol and CAD Development

The subject recruitment and imaging protocol was approved by the Wake Forest School of Medicine Institutional Review Board (IRB #5705). An individual representing the 5th percentile female (F05) in terms of height (149.9cm), weight (48.0±0.63kg), and body mass index (21.4kg/m2) was selected. Additionally, 15 external anthropomorphic measurements were taken from the subject to meet criteria published by Gordon et al (Gordon et al. 1989). A description of the medical imaging protocol and development of
the occupant CAD dataset has previously been reported by Davis et al (Davis et al. 2014; Gayzik et al. 2011). Prior to finalizing the geometries, select thoracoabdominal organ volumes were compared to ranges of small female organ volumes for verification (Davis et al. 2015).

**Model Development**

Mesh criteria for the GHBMC F05-O was set in a fashion similar to that of the GHBMC M50 occupant model (M50-O) in terms of element number, size, quality, and type. The model was developed with a target element edge length of 2-3 mm to reach a total element count of approximately 2.5 million. With regards to element quality, stringent thresholds were placed on several criteria: Jacobian (>0.3 for all solid elements and >0.4 for all shells), tetra-collapse (>0.2 for all elements), zero intersections, and a minimum time step value of 0.1 μs. The goal for these hard targets was to have 100% adherence to the standards. Additional element quality criteria (such as minimum angle and aspect ratio) were also in place with a target of 99% compliance for all elements.

In an effort to increase model stability, three of the main focuses of F05-O development were: 1) limiting the number of required contacts in each body region and limiting parts in multiple contacts, 2) maintaining mesh uniformity throughout the model, and 3) limiting the number of intersections and penetrations between parts (Gayzik et al. 2012; Shin et al. 2012). To accomplish these goals, a variety of meshing techniques were employed using commercially available software. These techniques included structured hexahedral meshing, tetrahedral meshing, morphing, and element property assignment based on CAD data.
For development of the average male model, the GHBMC established body region model (BRM) centers of expertise (COE) for the development and validation of regional finite element models using CAD geometries developed by the full body model (FBM) COE. These regions were then integrated by the FBM COE for full body validation. For development of the F05-O, the CAD and mesh development for each body region was conducted by the FBM COE. Throughout the meshing process, BRM COEs were consulted regarding the approach for body regional development. The BRM COEs provided valuable insight derived from lessons learned during the development of the M50-O model that became the foundation for F05-O development. In addition, the FBM COE worked closely with each BRM COE throughout the development process to integrate model enhancements identified by BRM COEs during regional validation. A complete listing with the location and principle investigator for each BRM COE can be found in the acknowledgements. A review of the meshing approach applied to each body region can be found in Appendix A.

**Simulations**

Validation efforts in this study are limited to the full body level. Full body level validation was performed through simulation of the F05-O model in ten impact conditions ranging from regional loading to full body sled tests. All simulations were conducted on version 2.3 of the model. Overall, six regional loading regimes were simulated: a frontal chest impact (Kroell et al. 1971; Kroell et al. 1974; Lebarbé and Petit 2012), two lateral shoulder impacts (Kemper et al. 2008; Koh et al. 2005), an oblique thoracoabdominal impact (Viano 1989), a frontal abdomen impact (Hardy et al. 2001), and a lateral pelvis impact (Bouquet 1998). Four full body impact environments were
also simulated: a frontal sled test (Shaw et al. 2009), a lateral sled test (Cavanaugh et al. 1990; Cavanaugh et al. 1993), a drop test (Stalnaker 1979), and a rear seat frontal impact (Forman et al. 2009; Michaelson et al. 2008). A summary of the PMHS characteristics for each impact environment can be seen in Table G 1. Table G 1 includes information on the number of males and females in each experiment, as well as the average age, weight, and height where available. All simulations were performed with LS-Dyna (v7.1.2 rev95028, LSTC, Livermore, CA) on the Wake Forest University Distributed Environment for Academic Computing (DEAC) cluster. The DEAC Cluster is a high performance computing environment running Linux Red Hat 6 with a heterogeneous collection of 2,368 computational cores and 3.9 TB of total memory.

Regional Impacts

For all regional impacts, the scaling methodology as reported by the PMHS studies referenced were used for scaling the F05-O results. The chest impact used a 23.4 kg cylindrical hub impactor with a 15 cm diameter and a prescribed impact velocity of 6.7 m/s. According to the experimental description, the impactor was aligned against the sternum at the 4th intercostal space. Thoracic deflection was described as the change in chest depth throughout the simulation. Forces were reported as the contact force of the impactor. Model data were filtered using SAE CFC300 (SAE 2014). The model outputs were compared to a meta-analysis of thoracic impact data performed by Lebarbé and Petit (Lebarbé and Petit 2012) in which an average response and corridors were created from a meta-analysis of several chest impact studies (Bouquet et al. 1995; Kroell et al. 1971; Kroell et al. 1974; Trosseille et al. 2008). The sample was comprised of 19 high speed impacts with 14 males and 5 females. Overall, the average mass of subjects used
for corridor development was 67.0 ± 13.1 kg (Table G 1). In agreement with corridor development, the F05-O baseline response was normalized to the average male corridors using a normalization technique developed by Mertz et al. and adapted to two mass systems by Viano et al (Mertz 1984; Viano 1989). This technique represents the impact as a lumped mass model consisting of two masses, one representing the model and one representing the impactor, connected through a linear spring. Typically when using this technique, an impulse momentum analysis is used to calculate the effective mass of the impacted body region. In this case however, as the authors were using metadata to develop the corridors, limitations in the data required the effective mass to be computed as proposed by Horsch and Patrick (Horsch and Patrick 1976). With this approach, the equations of the dissipated energy and conservation of momentum are combined to calculate the effective mass. The effective mass ratio ($R_m$) was then calculated as the ratio of the reference effective mass obtained from the mean corridor curve ($M_{eff\text{-}target}$) to the model effective mass. By assuming geometric similitude between the model and experimental dataset, the stiffness ratio ($R_k$) was calculated as the ratio of the reference M50 chest depth ($L_{target} = 229.5$ mm) to the F05-O chest depth ($L_{model}$). These values were then used to obtain the scale factors of interest as reported by Lebarbe and Petit (Lebarbé and Petit 2012).

The thoracoabdominal impact used the same impactor as the chest impact with a nominal velocity of 6.7 m/s. Following the experimental description, the center of the impactor was aligned 7.5 cm below the xiphoid process at 60° from the anterior aspect of the body. Model data were filtered using SAE CFC600 (SAE 2014). In this impact environment, the normalized force time-history corridors presented by Viano et al. were
used for model comparison (Viano 1989). Corridors were developed using the response of 5 subjects, 4 male and 1 female, with an average mass of 64 ± 15.1 kg (Table G 1). The experimental responses were normalized using the methods described above as adapted by Viano et al in 1989. Unlike the chest impact, the effective mass used for the thoracoabdominal impact was calculated by relating the impulse of the impactor to the acceleration of the model. Also, for the development of the stiffness ratio, the abdominal breadth was used to develop the stiffness ratio with an M50 reference value of 314 mm. Otherwise, the normalization process was similar to the chest impact with $R_m$ and $R_k$ used for development of the scale factors.

The lateral shoulder impact, as per Koh et al., was simulated with an impact velocity of 4.5 m/s using the same impactor applied for the chest and thoracoabdominal impacts. This was a 90° lateral impact occurring through the head of the humerus. For this study, deflection was normalized by the half bi-acromial width of the PMHS for comparison to the model and is presented as a percentage. Force data were obtained as the contact force of the impactor. Model shoulder compression was modeled as the difference in length between a node on the acromion and a node on the first thoracic vertebrae. The impact force was modeled as the contact force between the impactor and the shoulder. Model data were filtered using SAE CFC180 as per Koh et al (Koh et al. 2005). Corridors were developed from unpadded impacts to 5 male and 1 female PMHS with an average mass of 68.8 ± 12.0 (Table G 1). Force data were equal stress/equal velocity normalized using the procedure outlined by Eppinger et al. (Eppinger 1976). For this scaling technique, the only input data needed is the full body mass of both the model and the target mass used for corridor development. These data are then used to develop a
full body mass ratio ($\lambda$) of the target ($M_{\text{target}}$) and the model ($M_{\text{model}}$). This ratio is then used to derive the scale factors of interest. For this impact, a full body mass of 75 kg was used as the reference (Koh et al. 2005).

A second lateral shoulder impact was performed to simulate the experimental study by Kemper et al (Kemper et al. 2008). In this case, the simulation was conducted to represent the destructive thoracic side impact tests with the arm at 45 degrees. This impact consisted of a 23.4 kg impacting plate (41.5 x 25.5 cm) with an initial impact velocity of 12 m/s. As per the experimental study, special care was taken during impactor alignment to avoid contact with the pelvis. The contact force between the impacting plate and the model was used to report the model response. Model data were filtered using SAE CFC180 (SAE 2014). For comparison to the experimental subjects, model force and time data were scaled using an extension of the impulse momentum analysis for two mass systems presented by Viano et al. (Viano 1989). The procedures for scaling were obtained from the International Standards Organization (ISO) technical report ISO/TR-9790:1999 (ISO 1999). For this study, the standard effective mass was reported as 29.2 kg and the standard chest breadth used for development of the stiffness ratio was 349 mm (Kemper et al. 2008). For the impact configuration simulated in this study, Kemper et al. impacted two male subjects with an average mass of 66.7 kg (Table G 1).

The frontal abdominal impact consisted of a 2.5 cm diameter, 48 kg bar at 6.0 m/s. This was a free-back impact and occurred at the level of the umbilicus, or approximately L3. To represent the effective limit to abdominal compression, deflection corridors were normalized by the compressible abdominal depth of the PMHS for comparison to the
model (Vavalle et al. 2015). Compressible abdominal depth was defined as the abdominal depth from the anterior surface of L3 to the anterior surface of the body normal to L3, i.e. the L3 depth. The force of the impact was measured as the contact force of the rigid bar. Model data were filtered using SAE CFC180. Corridors were developed from three PMHS, 2 male and 1 female, with an average mass of 74 ± 23.3 kg (Table G 1). For comparison to the experimental corridors, the F05-O force and time signals were equal stress/equal velocity scaled using a reference mass of 78 kg (Eppinger 1976; Hardy et al. 2001).

The lateral pelvis impact used a square-faced impactor weighing 16 kg impacting with 800 J of energy. This required giving the impactor a 10 m/s velocity in the lateral direction. The impactor contacted the trochanter and iliac crest at 90 degrees lateral, according to the literature (Bouquet 1998). Pelvic deflection was measured as the relative displacement between the outer hip and a node on the center of the sacrum. Similar to the other rigid impact simulations, the contact force of the impacting plate was used to obtain force data. Model data were filtered using SAE CFC600 (SAE 2014). No normalization was applied for development of the experimental corridors. The experimental PMHS specimens consisted of 3 males and 1 female with an average mass of 55.4 ± 18.3 kg (Table G 1).

Full Body Impacts and Sled Simulations

The lateral sled test was modeled as a 6.7 m/s lateral impact using a Heidelberg-type sled into fixed steel plates (Cavanaugh et al. 1990; Cavanaugh et al. 1993). The impact environment included a flat rigid wall as a backrest, a Teflon seat, and five rigid impacting plates located at the shoulder, thorax, abdomen, pelvis, and knee. Torso forces
were obtained as the sum of the shoulder, thorax, and abdomen forces. The pelvis force was measured as the contacting force at the pelvis plate. Model data were filtered using SAE CFC300. Experimentally, the 6.7 m/s unpadded impacts were conducted on 3 subjects, 2 males and 1 female, with an average mass of 64.2 ± 0.6 kg (Table G 1). For comparison to the experimental data, the F05-O force and time signals were scale using the equal stress/equal velocity technique with a reference mass of 74.8 kg (Cavanaugh et al. 1990; Eppinger 1976).

The lateral drop test was simulated to specify a 1 meter free fall onto an instrumented rigid surface (Stalnaker 1979). For this study, contacts were developed to measure the force response of isolated body regions. In this case, the contact force of the torso region was used to assess model response. Model data were filtered using SAE CFC180. Corridors were developed from impacts to 2 male specimens with an average mass of 61 kg (Table G 1). For comparison to experimental corridors, the model force and time signals were scaled using the impulse momentum technique described by Mertz et al. as directed by ISO/TR-9790:1999. For this study, the standard effective mass was reported as 38 kg and the standard thoracic depth used for development of the stiffness ratio was 236 mm (ISO 1999).

The model was also simulated as a rear seat occupant in a frontal crash pulse as per Michaelson et al. (Michaelson et al. 2008) and Forman et al. (Forman et al. 2009). This simulation represented a 48 km/h frontal impact, which the experimentalists chose to represent the vehicle deceleration experienced by a mid-sized sedan during a full frontal barrier test. While the authors reported results from both 1) standard and 2) force-limiting with pretensioning retractor seat belt schemes, for this study all comparisons
were made to the standard belt setup. The boundary condition included a sled buck with a rear seat representing a 2004 mid-sized sedan. Also, as no front seat was included in the experimental impacts, no contact was defined between the model and the front seat during the simulation. Prior to simulation, the model was gravity settled in the seat for 100 ms to obtain realistic flesh contours within the seat. Seatbelt tension was measured over the duration of the pulse by modelling a cross-section plane output in the webbing. Head CG, shoulder, and pelvis kinematics were obtained using nodal displacements. Model responses were compared to the normalized responses of 2 male cadavers with an average mass of 57 kg (Table G 1).

Lastly, the frontal sled test was modeled as per data from Shaw et al. (Shaw et al. 2009). This simulation represented a frontal impact pulse with an overall change in velocity of 40 km/h. The boundary condition consisted of a simplified buck modeled as a rigid body with belt properties developed to match experimental conditions (26 kN of force at 7% strain). To recreate the experimental setup, no pretensioners or load-limiters were included. A foam knee bolster was included to limit forward excursion of the model lower extremities. Prior to simulation, the model was gravity settled for 200 ms to obtain realistic flesh contours on the seat plate. Both kinetic and kinematic responses were obtained for comparison to experimental values. All kinematic data were output in the SAE J1733 coordinate system. Reaction forces at the knee bolster and foot rests were also recorded in the global coordinate system and then transformed into a local coordinate system per the literature (Ash et al. 2012). Resultant force data were also obtained to represent the responses of the upper and lower shoulder belt and the outer lap belt. As the experimental data set was comprised of eight male PMHS with
approximately 50th percentile stature and mass, the experimental data were not scaled for PMHS size or age (Shaw et al. 2009). The average mass for the experimental subjects was 75.5 ± 2.8 kg (Table G 1).

**Morphing**

As a supplement to data scaling, a second approach for model validation was to geometrically morph the small female model to the body habitus of the average male. The goal of this analysis was to quantitatively compare the outcomes from both the post-hoc scaling technique described above and outputs from a model actually representing the target mass and anthropometry. Also, this allows a direct comparison to experimental data in full body sled tests, where confounding factors such as contact timing and vehicle interactions make it difficult to apply post-hoc scaling. To conduct this comparison, the F05-O model was morphed to the body habitus of the GHBMC M50-O model using radial basis function (RBF) interpolation with a thin-place spline as the basis function and a relaxation algorithm (Bookstein 1989; Donato and Belongie 2002; Stayton 2009). This approach has previously been applied for the development of the GHBMC large male (M95) and a GHBMC older occupant (Schoell et al. 2015; Vavalle et al. 2014). As part of the GHBMC project, external surface geometries representing the outer flesh of both the M50-O and F05-O have been developed from the same subjects used to construct these models (Davis et al. 2014; Gayzik et al. 2012). Therefore, the actual M50-O surface geometry was used as the target for the morph. The required inputs to perform thin plate spline interpolation are homologous landmarks on both the reference (F05-O) and target (M50-O) geometries, as well as the nodal coordinates for the baseline model (F05-O). To develop the homologous landmarks, the surface data of the F05-O geometry
were first converted to a point cloud using Geomagic Studio software (v12, Geomagic, Rock Hill, SC). In order to create homologous landmarks, the F05-O and M50-O surface data were registered using symmetric diffeomorphic registration in Advanced Normalization Tools (ANTS) (Avants and Gee 2004; Avants et al. 2008). The rigid, affine, and non-affine transformations obtained from this process were then applied to the F05-O landmarks to create the homologous target landmarks (M50-O). In total, each landmark set consisted of 1,871 points. Thin-plate spline interpolation and relaxation algorithms were performed using custom MATLAB scripts (R2013a, The MathWorks Inc., Natick, MA).

**Objective Evaluation**

Following simulation, the ISO/TS 18571 standard was used to quantitatively compare the model outputs to experimental time history curves. The standard uses a weighted average of four metrics, corridor, phase, magnitude and slope, to derive a total score describing the fit of data to a reference curve. Based on the algorithm, curve comparisons yield a score ranging from 0 to 1 for each component rating metric with 1 representing a good match within a set threshold, and 0 representing a poor match. When available, experimentally derived corridors were used to obtain the corridor score for a simulation. In cases where no inner and outer experimental corridors were produced, the ISO recommended fixed width corridors (±5% for the inner corridor and ±50% for the outer corridor) were used. The constant half widths of both inner and outer corridors are defined as the product of the respective half width percentage and the absolute maximum amplitude of the test signal ((ISO) 2013).
When reporting final quantitative evaluations for validation, researchers commonly report the average of each component rating metric to give a global view of the model response. However, it may be necessary to discriminate between signals of varying magnitude to provide a clearer picture of model behavior. This is particularly important for signals of low magnitude which can register artificially low scores. For instance if a given experimental signal peaks at 0.5 kN and the predicted signal is 0.25 kN it is off by 50%, yet in some circumstances this may not be significant compared to an orthogonal and biomechanically relevant signal that is 10 to 20 times that magnitude. Therefore, in cases where signals at a single anatomical location are being compared with a primary axis (ex. X-axis) or primary plane of motion (ex. X-Y plane), a magnitude weighting factor was applied to determine the overall average objective evaluation rating. With regard to kinematics, this weighting was only applied to the orthogonal components of the same signal (i.e. from the same sensor). Weight factors were derived by normalizing the peak value for each signal from the same sensor (e.g. the X, Y, and Z signals) by the sum of peaks for each orthogonal signal as per Equation (41). This factor will be referred to as the Test Magnitude Factor, or TMF.

$$\text{TMF} = \frac{R_i}{R_x + R_y + R_z}$$  \hspace{1cm} (41)

Where $R_i$ is the peak value of the test trace for a given plot and is divided by the sum of peaks for each orthogonal signal. The magnitude factor is then applied to the average ISO score for each respective orthogonal signal. The final ISO score for a signal is then considered to be the sum of the magnitude weighted orthogonal components.
**Hard Tissue Injury Prediction**

The GHBMC F05-O model predicts hard tissue injury in a selection of bony geometries by defining either stress or strain thresholds for fracture based on the literature (Li et al. 2010; Vavalle et al. 2015; Vavalle et al. 2013). Variations in the approach were dictated by the data available in the literature and methods required to accurately model failure in a variety of structures. For this study, the element deletion method was used to simulate fracture by removing elements from the simulation once the specified stress or strain threshold was exceeded. For the F05-O, cortical bone fracture of the skull and face is predicted when the maximum principle strain exceeds 0.0042 and diploe failure is predicted when the principle stress in an element exceeds 20 MPa (Hodgson et al. 1970). Failure strains for the ribs of the model were determined by computational optimizations reported by Li et al. (Li et al. 2010). From these optimizations, rib fracture is predicted using effective plastic strain with thresholds of 0.018 and 0.13 for cortical and trabecular bone respectively. Fracture of pelvic cortical bone is predicted using and effective plastic strain threshold of 0.02 and trabecular bone failure is modeled using a maximum principle strain threshold of 0.426 (Song et al. 2005). For the lower extremity, fracture is modeled using effective plastic strain with a value of 0.0088 and trabecular bone failure is modeled using a maximum principle strain threshold of 0.426 (Untaroiu et al. 2013). Following simulation, each impact condition was reviewed for fractures. Fractures were determined by visual inspection and verified by element failure reports output from LS-Dyna. The number of fractures was compared to the experimental results in each impact condition. Additionally, Abbreviated Injury
Scale (AIS 2005, Update 2008) scores were compared with experimental AIS codes by a trained AIS coder.

**VIII. RESULTS**

**Medical Imaging and CAD Development**

The selected volunteer 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. For all 15 external anthropomorphic measurements taken from Geraghty et al., the average subject percent deviation was 4.1% (threshold for inclusion was 5%) (Gordon et al. 1989). Data on each measurement has previously been reported in the literature (Davis et al. 2014).

In total, 66 scan series were collected across all modalities for a total of 14,170 images. Using this data, 3D CAD geometries were developed for all skeletal structures and each organ that was explicitly modeled. The skeleton consists of 182 individual bones. Explicit representations of 32 organs were developed, encompassing brain, thoracic, and abdominal organs relevant to biomechanical modeling. An in-depth review of structures included in the F05-O CAD dataset can be found in the literature (Davis et al. 2014). Figure 34 displays the assembled CAD model of the F05-O.  

![Figure 34. F05 occupant CAD model.](image)
Finite Element Model Development

The GHBMC F05-O finite element model was developed from the CAD dataset described above. Overall, the F05-O v2.3 met each of the pre-defined element quality criteria targets and all elements have an initial time step greater than 0.1 μs. Additionally, the percent added mass upon initialization is less than 1% of the full body mass. A review of the number of elements by body region and by element type can be found in Figure 35 and Figure 36 respectively. Additionally, Figure 37 displays the element quality distribution for the two element quality criteria that required 100% adherence (tetra collapse and Jacobian). Summary statistics of the F05-O model can be found in Table 16.

<table>
<thead>
<tr>
<th>Table 16. F05-O v2.3 summary statistics.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of Parts</td>
</tr>
<tr>
<td>Number of Elements</td>
</tr>
<tr>
<td>Number of Nodes</td>
</tr>
<tr>
<td>Model Mass (kg)</td>
</tr>
<tr>
<td>Number of Penalty Contacts</td>
</tr>
<tr>
<td>Number of Tied Contacts</td>
</tr>
</tbody>
</table>

In total, 796 material cards were used to characterize parts within the model. This includes duplicated cards for some structures, such as bony materials of the left and right extremities. Individual material properties were developed to provide the user with the ability to modify parameters for specific structures if desired for a particular application. To expedite development, material models were primarily carried over from the average male model (GHBMC M50-O v. 4.4). A comprehensive list of materials can be found in the GHBMC M50-O manual (GHBMC 2011). This was also necessary as there is relatively little biomechanical data specifically characterizing female biological tissues and the current literature suggests limited sex-related differences (Schoell et al. 2015).
This approach not only provided a baseline for model development, but also facilitated comparisons to the GHBMC M50-O.

![Figure 35. Element breakdown by body region.](image1)

![Figure 36. Element breakdown by element type.](image2)

![Figure 37. Distribution of 3D element quality for hard targets. (Left) 3D tetra collapse. (Right) Jacobian](image3)
The use of node-to-node connections and element assignment techniques, where appropriate, significantly reduced the number of required contacts as compared to the GHBMC M50-O model. For example, the femur, quadriceps, hamstrings, and thigh flesh were all developed using a continuous mesh, removing the need for a contact between these parts. The F05-O model implements 32 contacts, which is a 92.8% reduction in the number of contacts compared to the M50-O. The assembled F05-O model can be seen in Figure 38.

**F05 Morphed Model**

As described above, the F05-O model was also morphed to the body habitus of the M50-O to compare the model to mass normalized corridors as a supplement to *post-hoc* scaling of the baseline model data. In total 1,871 points were used for both the reference (F05-O) and target (M50-O) landmark datasets. Following the application of the radial basis function-thin plate spline technique, the F05-O morphed model had a full body mass of 76.2 kg. Also, because the morphing process only affects the nodal coordinates of the model, all other model aspects remained identical to the baseline F05-O model. Therefore, the F05-O morphed model contained the same number of nodes, elements, and parts with the same material formulations and contact definitions as the F05-O. All element quality criteria were met using the same standards as the baseline F05-O model.
As such, changes in mesh quality and element size are not expected to affect the results. The landmarks and final morphed model can be seen in Figure 39.

**Simulation Results**

Both the baseline F05-O and F05-O Morphed models ran to normal termination in all 10 impact conditions for a total of 20 simulations. A simulation time lapse for each of the localized, rigid hub impacts can be seen in Figure 40. The model shown in each of these figures is the baseline F05-O model (i.e. no geometric morphing has been applied). Each of these cases represents a simulation where the impactor is considered to have a finite mass.

The model response for each of the localized, rigid impacts can be seen in Figure 41. For each simulation, these figures contain the baseline F05-O model response, the F05-O post-hoc data scaled response, and the F05-O Morphed response. As described above, the baseline F05-O model response was scaled for each of the rigid impact conditions using the technique applied by the experimentalists. Table G 2 provides a summary of the scale factors used for each impact condition. Within Figure 41, the curves are plotted against the experimental average male corridors in each case. For quantitative evaluation, the model responses were compared to the experimental mean time history traces. Overall, the average ISO score for these localized impacts was 0.68 for the scaled responses and 0.76 for the morphed model responses. The two techniques were most similar in the overall slope scores (0.63 for scaled and 0.61 for morphed). The largest disparity between the two approaches was found in the corridor score, with the scaled data scoring an average of 0.62 and the morphed model an average of 0.76. A full
summary of each component rating metric for the localized, rigid impacts can be seen in Table 17.

![Figure 39. F05-O Morphed finite element model (Left). F05-O (black) and M50-O (grey) landmarks used to control the model morph. (Right). Final geometry of the F05-O Morphed model compared to target landmarks.](image)

**Table 17. ISO scores for rigid impacts. Results included both scaled and morphed model responses.**

<table>
<thead>
<tr>
<th>Simulation</th>
<th>Signal</th>
<th>Response For Comparison</th>
<th>Corridor</th>
<th>Slope</th>
<th>Size</th>
<th>Phase</th>
<th>ISO Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thoraco-Abdominal</td>
<td>Force vs. Time</td>
<td>Scaled</td>
<td>0.61</td>
<td>0.55</td>
<td>0.94</td>
<td>0.56</td>
<td>0.65</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Morphed</td>
<td>0.81</td>
<td>0.48</td>
<td>0.97</td>
<td>0.79</td>
<td>0.77</td>
</tr>
<tr>
<td></td>
<td>Force vs. Time</td>
<td>Scaled</td>
<td>0.69</td>
<td>0.65</td>
<td>0.88</td>
<td>0.88</td>
<td>0.76</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Morphed</td>
<td>0.84</td>
<td>0.66</td>
<td>0.91</td>
<td>0.90</td>
<td>0.83</td>
</tr>
<tr>
<td></td>
<td>Force vs. Time</td>
<td>Scaled</td>
<td>0.57</td>
<td>0.82</td>
<td>0.68</td>
<td>0.78</td>
<td>0.68</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Morphed</td>
<td>0.86</td>
<td>0.83</td>
<td>0.94</td>
<td>0.91</td>
<td>0.88</td>
</tr>
<tr>
<td>Lateral Hub</td>
<td>Force vs. Time</td>
<td>Scaled</td>
<td>0.75</td>
<td>0.49</td>
<td>0.77</td>
<td>0.75</td>
<td>0.70</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Morphed</td>
<td>0.67</td>
<td>0.49</td>
<td>0.66</td>
<td>0.84</td>
<td>0.66</td>
</tr>
<tr>
<td></td>
<td>Force vs. Time</td>
<td>Scaled</td>
<td>0.49</td>
<td>0.62</td>
<td>0.50</td>
<td>0.84</td>
<td>0.59</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Morphed</td>
<td>0.62</td>
<td>0.58</td>
<td>0.83</td>
<td>0.67</td>
<td>0.66</td>
</tr>
<tr>
<td>Abdominal Bar</td>
<td>Force vs. Time</td>
<td>Scaled</td>
<td>0.62</td>
<td>0.63</td>
<td>0.75</td>
<td>0.76</td>
<td>0.68</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Morphed</td>
<td>0.76</td>
<td>0.61</td>
<td>0.86</td>
<td>0.82</td>
<td>0.76</td>
</tr>
<tr>
<td>Average</td>
<td>Scaled</td>
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<td>0.62</td>
<td>0.63</td>
<td>0.75</td>
<td>0.76</td>
<td>0.68</td>
</tr>
<tr>
<td></td>
<td>Morphed</td>
<td></td>
<td>0.76</td>
<td>0.61</td>
<td>0.86</td>
<td>0.82</td>
<td>0.76</td>
</tr>
<tr>
<td></td>
<td>T = 0</td>
<td>T = 1/3</td>
<td>T = 2/3</td>
<td>T = T_{final}</td>
<td></td>
<td></td>
<td></td>
</tr>
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<td>---------</td>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Frontal Chest</td>
<td><img src="image1" alt="Image" /></td>
<td><img src="image2" alt="Image" /></td>
<td><img src="image3" alt="Image" /></td>
<td><img src="image4" alt="Image" /></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Oblique Thoraco-Abdominal</td>
<td><img src="image5" alt="Image" /></td>
<td><img src="image6" alt="Image" /></td>
<td><img src="image7" alt="Image" /></td>
<td><img src="image8" alt="Image" /></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Lateral Hub (6.7m/s)</td>
<td><img src="image9" alt="Image" /></td>
<td><img src="image10" alt="Image" /></td>
<td><img src="image11" alt="Image" /></td>
<td><img src="image12" alt="Image" /></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lateral Plate (12m/s)</td>
<td><img src="image13" alt="Image" /></td>
<td><img src="image14" alt="Image" /></td>
<td><img src="image15" alt="Image" /></td>
<td><img src="image16" alt="Image" /></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Abdominal Bar</td>
<td><img src="image17" alt="Image" /></td>
<td><img src="image18" alt="Image" /></td>
<td><img src="image19" alt="Image" /></td>
<td><img src="image20" alt="Image" /></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lateral Pelvis</td>
<td><img src="image21" alt="Image" /></td>
<td><img src="image22" alt="Image" /></td>
<td><img src="image23" alt="Image" /></td>
<td><img src="image24" alt="Image" /></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Figure 40. Simulation time-lapse for localized, rigid impact validation cases.
Figure 41. Local, rigid hub impact model results plotted against experimental corridors, where available. In each case, the baseline F05-O response, post-hoc scaled response, and morphed model response are shown.
The scaled and morphed model responses for the frontal chest impact can be seen in the upper left corner of Figure 41. For this simulation, the F05-O model response was scaled by a factor of 1.28 for force signals and 1.21 for deflection. Within the figure, the model outputs are compared to the corridors reported by Lebarbé and Petit (Lebarbé and Petit 2012). For this simulation, both the scaled data and morphed model responses fell within the corridors for the duration of the event. Experimentally, Kroell et al. reported a contrast between specimens with regards to a “leading edge” or force spike at low levels of deflection (Kroell et al. 1971). Some specimens were found to have short duration, high amplitude spikes while others did not display these spikes. While Kroell et al. originally smoothed these spikes for the development of force-deflection corridors, Lebarbe and Petit included these early spikes on some of the force-time responses, leading to the introduction of these spikes in the force-deflection corridors presented in Figure 41. The experimental mean curve from the corridors experiences a peak force of 4.27 kN at 12.1 mm of deflection. After a decline in force, the force then rises to a second peak of 4.0 kN that occurred at 74.4 mm of deflection. The peak of the mean experimental deflection was 84.3 mm. The F05-O model did not exhibit the early spike in force. The baseline F05-O output had lower peaks for both force and deflection (3.41 kN and 72.5 mm) compared to the secondary force peak and deflection peak from the experimental mean. The scaled response was closer to the experimental values with a peak force of 4.4 kN and a maximum deflection of 87.4 mm. The morphed model had peak values for force and deflection of 3.62 kN and 73.7 mm respectively.

For the frontal abdominal bar impact, the model responses can be seen in the upper right corner of Figure 41. In this case, the response force was scaled by a factor of 1.31.
However, it should be noted that no scaling was applied to the deflection response as compression of the bar was normalized by the anterior abdominal depth, thus normalizing the response as a deflection percentage. Therefore, the deflection for the baseline F05-O and data scaled response are the same. In this case, the force response for these 6m/s mid-abdomen impacts exhibits the single mode response reported experimentally (Hardy et al. 2001). The peak force of the baseline F05-O model was 4.5 kN compared to 5.9 kN and 5.0 kN for the scaled and morphed model outputs respectively. Experimentally, the normalized PMHS test data ranged from 3.8 to 5.0 kN. In general, the model exhibited a similar trend to the experimental data with a constant stiffness up to peak penetration followed by a sharp drop in force. However, the baseline model experienced a sharper drop in force compared to the morphed model.

The response to the oblique thoracoabdominal hub impact can be seen on the second row of Figure 41. The model is compared to the force time history on the left and the structural force-deflection response on the right. The force and deflection of the baseline F05-O model were scaled by factors of 1.13 and 0.99 respectively. Overall, the shape of the force time history plot matches the experimental data for the morphed model, which falls within the corridors for the duration of the signal. The peak force of the scaled response (4.5 kN) was at the upper range of the experimental corridors (2.5 kN to 4.5 kN). With regards to deflection, both the scaled (10.2 cm) and F05-O Morphed (12.2 cm) model responses fell within the experimental range of 8 to 15 cm.

The lateral hub impact response can be seen on the third row of Figure 41. In this case, the left plot represents the force-time history of the model and the right plot represents the thoracic deflection normalized by the distance from the acromion to
sternum. The force response of the baseline F05-O model was scaled by a factor of 1.28. As the deflection response was normalized by half of the bi-acromial breadth, no scaling was applied to this response. For force, both the scaled (2.6 kN) and F05-O Morphed (2.4 kN) model responses are near the lower limit of the experimental corridors (2.7 kN). With regards to deflection percentage however, there is a larger difference between the scaled (27.2%) and F05-O Morphed (21.8%) model response with the morphed model displaying less thoracic compression compared to half bi-acromial breadth. In this case, the morphed model falls within the deflection percentage corridors for the duration of the signal.

The model’s response to the lateral plate impact can be referenced in the bottom left corner of Figure 41. In this case, the model response is compared to the two destructive experimental PMHS tests reported in the literature (Kemper et al. 2008). For this impact condition, the baseline F05-O force response was scaled by a factor of 1.08. The peak forces for both the scaled F05-O (9.6 kN) and morphed model (10.0 kN) more closely matched the lower of the two cadaveric impacts, which had a peak force of 10.1 kN.

The applied force-deflection response of the lateral pelvis impact can be seen in the bottom right corner of Figure 41. For this impact configuration, no scaling was applied as the experimental corridors were not normalized to represent the average male. However, the baseline F05-O and morphed models were compared to the overall experimental range. The peak force for the baseline F05-O, 11.8 kN, was within the experimentally reported range of 6 kN to 14 kN. The peak force of the morphed model was near the upper limit of the corridors with a peak force of 14.3 kN.
deflection for the morphed model was 20.0% higher than the baseline response, but both curves were within the experimental ranges displayed by the force-deflection corridors.

The simulation time-lapse for the full body, rigid wall impacts and sled test validation cases can be seen in Figure 42. As described above, the model shown in each of these figures is the baseline F05-O model (i.e. no geometric morphing has been applied).

The temporal thoracic force response for the 1 meter drop test can be seen in Figure 43. To directly compare to the experimental corridors, in this case, the model response curves were aligned to the experimental average using the methods described by Maltese et al. (Maltese et al. 2002). This alignment was performed for each model curve (baseline, scaled, and morphed) individually. The force vs. time corridors were obtained from ISO TR 9790 (ISO 1999). The baseline F05-O force response was scaled to the average male corridors using a factor of 1.30. For this case, the scale response was within the corridors for the duration of the signal and the F05-O Morphed response was within for all but 1 ms. In terms of peaks, both the scaled (5.0 kN) and F05-O Morphed (5.1 kN) responses were within the corridors, but underestimated the mean peak of 6.5 kN.
Figure 42. Simulation time-lapse for full body, rigid wall and sled validation cases.

Figure 43. Model response for lateral drop test.

The F05-O model response in the lateral sled test can be seen in Figure 44. For this case, both the temporal torso and pelvis forces are presented. As the timing of pelvis response is confounded by contact timing of the torso, the pelvis forces were aligned to the peak of the mean experimental curve. For this impact, the baseline F05-O forces were scaled to the average male corridors using a factor of 1.28 for both the torso and pelvis. For torso forces, the scaled response had a peak of 9.3 kN whereas the F05-O
Morphed model had a peak of 7.6 kN. For this case, the upper bounds of the corridors were 8.5 kN. With regards to pelvis forces, the scaled and F05-O Morphed responses predicted peak forces of 12.6 kN and 11.4 kN respectively compared to 9.8 kN as the upper bounds of the experimental corridor.

Overall, the average ISO score for the lateral sled test was 0.67 for the scaled response and 0.75 for the F05-O Morphed. A breakdown of each component rating metric can be seen in Table 18. Based on the combined objective results from this rigid wall impact and the localized hub impacts described above, no scaling was applied to the results from the two sled impact cases. Rather, the F05-O Morphed response was used to compare to experimental data. This approach allowed a direct comparison to the experimental data without the potential sources of error that arise when attempting to scale response data in a complex impact environment, such as interaction with multiple vehicle components like seat belts and knee bolsters.

Figure 44. Model response for lateral sled test.
Table 18. ISO scores for the lateral sled test. Results included both scaled and morphed model responses.

<table>
<thead>
<tr>
<th>Simulation</th>
<th>Signal</th>
<th>Response For Comparison</th>
<th>Corridor</th>
<th>Slope</th>
<th>Size</th>
<th>Phase</th>
<th>ISO Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lateral Sled Torso</td>
<td>Scaled</td>
<td>0.72</td>
<td>0.53</td>
<td>0.92</td>
<td>0.81</td>
<td>0.74</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Morphed</td>
<td>0.76</td>
<td>0.44</td>
<td>0.90</td>
<td>0.72</td>
<td>0.71</td>
<td></td>
</tr>
<tr>
<td>Pelvis</td>
<td>Scaled</td>
<td>0.66</td>
<td>0.31</td>
<td>0.46</td>
<td>0.93</td>
<td>0.60</td>
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<tr>
<td></td>
<td>Morphed</td>
<td>0.87</td>
<td>0.39</td>
<td>0.81</td>
<td>0.97</td>
<td>0.78</td>
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</tr>
<tr>
<td>Average</td>
<td>Scaled</td>
<td>0.69</td>
<td>0.42</td>
<td>0.69</td>
<td>0.87</td>
<td>0.67</td>
<td></td>
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<tr>
<td></td>
<td>Morphed</td>
<td>0.82</td>
<td>0.42</td>
<td>0.86</td>
<td>0.85</td>
<td>0.75</td>
<td></td>
</tr>
</tbody>
</table>

The upper shoulder belt tension and kinematics of the head, shoulder, and pelvis for the rear seat frontal impact can be seen in Figure 45. As described above, no scaling was applied to the baseline F05-O response. Rather, the F05-O Morphed response was used to make direct comparison to the experimental data. In this case, the F05-O Morphed model closely matched peak upper belt tension with a peak force of 7.4 kN compared to 7.1 kN experimentally. Kinematic traces of the head, shoulder, and pelvis compared to an average of the two experimental traces can be seen in Figure 45. The F05-O Morphed model exhibited similar head trajectories in both the X and Z directions the PMHS tests, but overestimated forward shoulder excursion. Conversely, the model underestimated peak forward excursion of the pelvis.

Overall, the average magnitude weighted ISO score for the F05-O Morphed model in the rear seat sled test validation cases was 0.65. In this case, the belt forces and the sagittal plane motion for each location of kinematic output were used to develop the weighting factors. A summary of scores for the rear seat impact condition can be seen in Table 19. The detailed objective ratings for each signal as well as the weighting factors used for analysis can be found in Table G 3.
Figure 45. Model response for rear seat frontal sled pulse. (Left) Upper shoulder belt forces. (Right) Model kinematics. Shoulder data is from the left side of the model.

Table 19. ISO scores for the rear seat frontal sled

<table>
<thead>
<tr>
<th></th>
<th>Corridor</th>
<th>Shape</th>
<th>Size</th>
<th>Phase</th>
<th>ISO Score</th>
<th>ISO Mag Weighted Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Seat Belt Forces</td>
<td>0.73</td>
<td>0.53</td>
<td>0.84</td>
<td>0.84</td>
<td>0.74</td>
<td>0.74</td>
</tr>
<tr>
<td>Head Kinematics</td>
<td>0.78</td>
<td>0.81</td>
<td>0.75</td>
<td>0.65</td>
<td>0.75</td>
<td>0.80</td>
</tr>
<tr>
<td>Shoulder Kinematics</td>
<td>0.61</td>
<td>0.61</td>
<td>0.13</td>
<td>0.40</td>
<td>0.47</td>
<td>0.59</td>
</tr>
<tr>
<td>Pelvis Kinematics</td>
<td>0.52</td>
<td>0.53</td>
<td>0.20</td>
<td>0.07</td>
<td>0.37</td>
<td>0.49</td>
</tr>
<tr>
<td>Average</td>
<td>0.66</td>
<td>0.62</td>
<td>0.48</td>
<td>0.49</td>
<td>0.58</td>
<td>0.65</td>
</tr>
</tbody>
</table>

The seat belt tensions for the frontal sled impact for both the baseline F05-O and F05-O Morphed models can be seen in Figure 46. Knee bolster forces and foot pan forces can be seen in Figure 47. Kinematic traces of the head and spine can be found in the appendix in Figure G 1 - Figure G 4. With regards to seatbelt forces, the F05-O Morphed model response was a close match to the mean experimental peak shoulder belt forces (upper shoulder peak = 6.8 kN, lower shoulder peak = 5.2 kN) with a peak upper shoulder belt force of 7.0 kN and a lower shoulder belt peak force of 5.2 kN. The model predicted a peak lap belt force of 1.3 kN compared to the upper corridor limit of 1.0 kN. When looking at the primary direction of loading for the knee bolster and the foot pan, the F05-O Morphed model’s peak forces were found to be within the corridors for each. With regards to the knee bolster forces, the model predicted right knee forces of 2.8 kN.
and left knee forces of 3.0 kN. These compare to experimental means of 2.3 kN and 2.5 kN for the right and left knees respectively. For foot forces, the model predicted a peak load of 4.9 kN compared to the mean experimental peak force of 5.0 kN.

With regards to the frontal sled case, each kinematic trace was comprised of three orthogonal signals representing X, Y, and Z displacements. Therefore, individual signals in this case were magnitude weighted in order to develop the ISO score for each measurement location. The overall average magnitude weighted score for the frontal sled impact was 0.72. Both the magnitude weighted scores and average ISO scores for this impact configuration can be seen in Table 20. A detailed summary of each component rating metric prior to application of weighting factors can be found in the appendix in Table G 4.
Figure 46. Model seat belt response for frontal sled pulse.

Figure 47. Model kinetic responses for frontal sled pulse.
Table 20. ISO scores for frontal sled impact. Overall magnitude weighted scores are reported for each signal group.

<table>
<thead>
<tr>
<th>Signal Group</th>
<th>Corridor</th>
<th>Shape</th>
<th>Size</th>
<th>Phase</th>
<th>ISO Score</th>
<th>ISO Mag Weighted Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Seat Belt Forces</td>
<td>0.75</td>
<td>0.69</td>
<td>0.71</td>
<td>0.84</td>
<td>0.75</td>
<td>0.87</td>
</tr>
<tr>
<td>R Knee Forces</td>
<td>0.52</td>
<td>0.26</td>
<td>0.32</td>
<td>0.82</td>
<td>0.48</td>
<td>0.57</td>
</tr>
<tr>
<td>L Knee Forces</td>
<td>0.62</td>
<td>0.38</td>
<td>0.54</td>
<td>0.80</td>
<td>0.59</td>
<td>0.68</td>
</tr>
<tr>
<td>Foot Forces</td>
<td>0.48</td>
<td>0.49</td>
<td>0.50</td>
<td>0.66</td>
<td>0.52</td>
<td>0.67</td>
</tr>
<tr>
<td>Head Kinematics</td>
<td>0.79</td>
<td>0.79</td>
<td>0.47</td>
<td>0.87</td>
<td>0.74</td>
<td>0.79</td>
</tr>
<tr>
<td>T1 Kinematics</td>
<td>0.62</td>
<td>0.70</td>
<td>0.32</td>
<td>0.49</td>
<td>0.55</td>
<td>0.74</td>
</tr>
<tr>
<td>T8 Kinematics</td>
<td>0.60</td>
<td>0.68</td>
<td>0.35</td>
<td>0.64</td>
<td>0.57</td>
<td>0.75</td>
</tr>
<tr>
<td>L2 Kinematics</td>
<td>0.59</td>
<td>0.67</td>
<td>0.48</td>
<td>0.75</td>
<td>0.62</td>
<td>0.72</td>
</tr>
<tr>
<td>Average</td>
<td>0.62</td>
<td>0.58</td>
<td>0.46</td>
<td>0.73</td>
<td>0.60</td>
<td>0.72</td>
</tr>
</tbody>
</table>

*Hard Tissue Injury Prediction*

A comprehensive report of hard tissue injuries for the F05-O model can be seen in Table G 5. Within this table, all F05-O fracture predictions were obtained from the baseline model with no geometric morphing applied. In addition, the F05-O model is also compared to experimental injury reports, as well as the outputs of the M50-O v4.4 model. For the F05-O and M50-O results, the rib level and side of the ribcage relative to the mid-sagittal plane are reported by labeling right or left as R and L respectively followed by the rib level. In cases where there are multiple fractures at the same rib level and side, the number of local fractures are labeled in parentheses (e.g. R3(2)). For rigid impacts with engagement of the thorax (seven cases), the F05-O model predicted an average of 2.4 thoracic fractures compared to 4.3 for the M50-O. For the three impact conditions with pelvic engagement, the F05-O model predicted an average of 6 fractures compared to 1 for the M50-O. For the sled cases, the F05-O model predicted an average of 8.5 fractures compared to 12 for the M50-O.
VIII. DISCUSSION

This study presents the development of a detailed human body finite element model of a representative 5th percentile female. The model is unique in that its geometry and assembly were based on an extensive external anthropometry and multi-modality image dataset of a specific subject matching small female anthropometry, specifically collected for the purpose of FEM development. This approach provided detailed data of both bony tissue and internal structures and their relative orientation in a driving posture. The full dataset was leveraged, where appropriate, to facilitate model reconstruction. Following mesh development, simulation validation was presented in 10 loading conditions ranging from localized, rigid impacts to full body sled cases. The model was also exercised in a variety of impact energies and loading directions. To compare the model directly to experimental data, two methods for evaluation were performed. The first was to mathematically scale the baseline F05-O results to the target experimental corridors. To make an accurate comparison, the outputs were scaled using the techniques reported by the experimentalists. Therefore, the F05-O model was simulated using the same impactor and input energy as if it were one of the PMHS included in the experimental dataset, and then scaled as per the experiment. The second approach to model validation was to geometrically morph the F05-O model to the body habitus of the M50 to create a direct comparison and a supplement for post-hoc data scaling.

In general, the F05-O model showed fair to good agreement with the experimental data based on the ISO objective rating metric. Overall, the model attained an average ISO score of 0.67 for the scaled responses and 0.76 for the F05-O Morphed responses in the rigid impacts. When looking at individual component rating metrics for the rigid
impacts, the scaled responses attained the highest scores in phase and size, with scores of 0.79 and 0.74 respectively. For F05-O Morphed, size was the highest scoring component rating metric with an average score of 0.86. Qualitatively, this agrees with the comparison of peaks presented in the results. In addition, the F05-O morphed model attained higher corridor scores (0.78) compared to the scaled responses (0.64).

In general, the overall shape or structural response of the scaled data was similar to the F05-O Morphed outputs. This is also demonstrated in the similar slope scores between the scaled response and the F05-O Morphed model response (0.57 and 0.55 respectively). One exception to this was the abdominal bar impact, where the F05-O scaled results had a sharper drop in force compared to the morphed model. This is likely due to the difference in engagement with the lower rib cage and solid organs of the abdomen that caused a more pronounced rebound of the bar in the baseline case.

Based on the results of this study, full body morphing was found to be an effective normalization approach when assessing the biofidelity of FEMs. This finding was further substantiated through qualitative analysis of the resulting curves. Therefore, the F05-O Morphed model was used for comparison to the experimental average male corridors for the simulations within a vehicle buck. This approach was taken to avoid the potential limitation of data scaling the model in a setting where contact timing and interactions with multiple components can influence the scaling factors and results. Overall, the model was found to capture the primary axis loads and kinematics well in terms of peaks and attained an average weighted ISO score of 0.69 for the two sled impact conditions.

While the F05-O Morphed model was found to produce response data representative of the average male, it should not be considered a replacement for the
The F05-O Morphed model does have the same body habitus of the M50-O, but the morph was based solely on the outer surface of the M50-O. Therefore, the internal geometries are not necessarily representative of the average male. For example, the ribcage of the F05-O Morphed model does not possess the same shape as the M50-O based solely on the morph controlled by external landmarks. Therefore, this approach allows the evaluation of global kinetics and kinematics, like the responses evaluated in this study, but prevents the model’s use for localized tissue injury prediction.

As localized tissue injury prediction is a desired use for these detailed HBMs as a research tool, this also provides reasoning for the ground up development approach used for the F05-O model. While components of the M50-O model could have been morphed to the F05-O, significant geometrical differences between the subjects in areas like the skull, ribcage, pelvis, and the majority of thoracoabdominal organs limited the feasibility of this approach as an option for full model development. In order to have more control over key model parameters, such as fidelity to the source F05-O geometries and maintaining element quality, the mesh was recreated directly from the source data. This also allowed the implementation of the model enhancements described in the methods.

With regards to hard tissue injury prediction, the F05-O model was within the reported number of fractures for the majority of the experimental impact conditions. While the number of rib fractures tended to be on the lower end of the range, the geometry of the rib cage was developed from a healthy 24 year old subject, which could lead to some discrepancies when comparing to cadaveric datasets that are often composed of older subjects. Additionally, the biovariability of the subjects used in the experimental datasets can yield large differences in injury results. For example, in the frontal sled
impact condition, the number of rib fractures ranged from 2 in a specific subject to 27. However, as the geometries used to develop the model were taken from a young, healthy individual and the material properties were primarily derived from experimental studies involving older specimens, it makes it difficult to characterize the general age of the model. As age is frequently associated with overall injury risk, future work could apply the techniques presented by researchers such as Klein et al. and Schoell et al. to parametrically adjust the age of the model by morphing geometries and scaling material properties to meet specific age targets (Klein 2015; Schoell et al. 2015).

While the injuries evaluated in this study were limited to the prediction of hard tissue injury, the anatomical components identified for inclusion within the model were selected specifically for the evaluation of crash induced injuries (CIIs). For example, components were included to provide the ability to model injuries beyond fracture such as cerebral contusion or cervical spine disc avulsion. However, other structures designed to facilitate passive load transfer and promote accurate kinematics were also included. For example, secondary branches of the aorta, vena cava, and hepatic portal vein were included due to their natural role as a tether for abdominal organs. Passive musculature has also been included as a means to allow a biofidelic transfer of forces following blunt impact. Because this model will ultimately be used for the evaluation of tissue response to blunt impact, much of the microvasculature of the human body was not included. This approach was taken since final model validation was compared to empirical data obtained from experiments conducted at the organ or full-body levels.

One limitation of this work was the sample size used for model development. However, this was a pragmatic decision made to enhance the practicality of the study.
The careful recruitment of one anthropometrically representative female allowed for the collection of an extensive external anthropometry and medical imaging dataset that would not have been feasible with additional volunteers. By taking this approach, medical imaging data was available to verify results of the developed geometries. Where possible, medical images of representative small females were consulted for additional verification (Davis et al. 2015).

There is also limited data in the literature detailing the biomechanical response of the small female. In the impact simulations conducted in this study, the experimental data was developed using predominately male subjects with a range of anthropometries. Therefore, the data was typically normalized to the average male for corridor development. This variation in body habitus also poses a challenge for normalization as one of the main normalization assumptions is geometric similitude. To account for this, some techniques, such as the impulse momentum analysis developed by Mertz et al., seek to take body region specificity into account by deriving an effective mass and implementing anthropometric ratios (Mertz 1984). Therefore, direct comparisons of the baseline model to published corridors require transformation of the data to the target mass of the experimental work. However, this work has shown that geometrically morphing the model to the target body habitus can be a valuable tool for making these direct comparisons.

Another limitation for this work is that detailed external geometries may not always be available as a target for morphing. The methods for geometrically morphing the small female to the average male were facilitated in this study by using detailed surface geometries of a representative average male. However, recent work from the University
of Michigan Transportation Research Institute (UMTRI) has developed detailed parametric shape models of the outer surface of adult males and females that can increase the feasibility of this type of validation (Reed et al. 2014). These surface data could allow this approach to be used not only for modeling the average male, but also a wider range of anthropometries (Hu et al. 2016; Klein 2015). Additionally, Jolivet et al. have shown that it is feasible to create morphed models using a reduced dataset consisting of anthropometric measurements (Jolivet et al. 2015).

While this work has focused on full body validation of the small female finite element model, regional validation has been conducted concurrently by GHMBC-sponsored academic research centers (see Acknowledgements). All local and single body region-based validation cases were performed by partnering BRM COEs over the course of the project. Future work will focus on further validation of the GHBMC F05-O finite element model at both the regional and full body levels, with an emphasis on soft tissue injury. The validation cases reported in this study were comprised of rigid, fixed energy impacts in a variety of direction and two frontal sled cases. In the future, additional sled tests will be simulated to evaluate the model response in oblique and lateral conditions.

VIII. CONCLUSIONS

This study has detailed the development of the GHBMC 5th percentile female finite element model. The model was developed from an extensive dataset based on a single, living subject closely matching anthropometric criteria for a small female. The model includes detailed bony and soft tissues relevant to assessing crash induced injuries in omnidirectional impact conditions. The F05-O model has 2.6 million elements, 1.4 million nodes, and weighs 51.1 kg. The use of increased node-to-node connections and
element assignment techniques were implemented during development to substantially limit the number of required contacts, employing 92.8% fewer contacts than the GHBMC M50-O model.

Overall, the model was found to be robust and showed fair to good agreement with experimental biomechanical response data as per the results of ISO/TS 18571 objective rating metrics. Based on the findings of this study, it was quantitatively demonstrated that full body morphing can be a more effective means of assessing biofidelity than post-hoc data scaling. The GHBMC F05-O model is poised to provide needed insight to the biomechanics of small female occupants as human body modeling is extended to address ever more challenging aspects of injury biomechanics.

VIII. ACKNOWLEDGEMENTS

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VIII. APPENDIX F

Head

All geometries of the cranium were developed using a structured hexahedral mesh in TrueGrid (XYZ Scientific Applications, Livermore, CA). Examples of the head mesh of the F05-O can be seen in Figure F 1.

![Figure F 1. (a) Skull and facial bones of the F05-O. (b) Medial representation of the model skull, facial bones, and brain structures. Elements were blanked on the model centerline. (c) Depiction of outer brain structures including the CSF, tentorium, superior sagittal sinus, cerebrospinal fluid, arachnoid. (d) Deep brain tissues including the corpus callosum, lateral ventricle, fornix, basal ganglia, thalamus, midbrain, and third ventricle.](image)

The included geometries and mesh strategy were similar to those reported by Mao et al (Mao et al. 2013). For the F05-O, the cranial vault and cerebrum were developed as a continuous mesh. This was done to ensure the quality of the brain-to-skull interface. The posterior surface of the brainstem is connected node-to-node with the cerebellum and the superior surface is connected to the cerebrum via the mid-brain. Remaining geometric features within the brain were developed using element assignment techniques based off of the CAD data. These techniques were used to maintain the node-to-node connection throughout the brain. The solid geometries of the face, with the exception of the mandible, were hexahedrally meshed in TrueGrid. Because of the similarity in shape...
between the F05-O and M50-O mandible, this geometry was morphed from the GHBMC M50-O using ANSA software (BETA CAE, Thessaloniki, Greece).

**Neck**

The modeling of the neck region proceeded in a similar fashion to the M50-O (Cronin et al. 2012; DeWit and Cronin 2012). The neck region of the model was predominately meshed using hexahedral elements. The exceptions were tetrahedral transition regions used to allow node-to-node connections at the origin and insertion sites of the 52 explicitly represented neck muscles. Images of the F05-O neck region can be seen in Figure F 2.

![Figure F 2, F05-O neck mesh. Top) Cervical spine with intervertebral discs. Middle) 52 explicit neck muscles. Bottom) Example of node-to-node attachment of neck muscle at insertion](image)

1D Hill-type muscle elements were also embedded in the passive 3D musculature to allow for closed-loop control of muscle activation parameters as developed for the average male model by Dewit and Cronin (DeWit and Cronin 2012; Hedenstierna and
Halldin 2008; Iwamoto et al. 2009). The solid hexahedral elements of the cervical spine were used to represent trabecular bone, with quadrilateral shell elements used to characterize cortical bone. The facets of each cervical vertebra were developed via extrusion from the surface of transverse process. The intervertebral discs (IVDs) at each vertebral level of the model were separated into two parts representing the nucleus and ground substance of the disc. In addition, the ground substance of each IVD was modeled with four layers of elements in the anterior-posterior direction to allow the inclusion of five annulus fibrosis layers as shell elements (DeWit and Cronin 2012). The IVDs were connected to the superior and inferior vertebral bodies via tiebreak contact to allow the modeling of disc avulsion (DeWit and Cronin 2012).

**Thoracoabdominal and Pelvic Region**

The thorax of the F05-O model was predominately meshed with solid hexahedral elements (Li et al. 2010; Li et al. 2010; Poulard et al. 2015). The sternum, ribs, and costal cartilage were developed using a structured “butterfly” mesh topology in TrueGrid. Two shell layers were implemented between each rib to represent the internal and external intercostal muscles. The heart, lungs, aorta, vena cava, and secondary large vessels were also hexahedrally meshed. Long bones of the upper extremity (humerus, radius, and ulna) were morphed from the M50-O using control point morphing in ANSA software. Because they are modeled as rigid parts, the thoracic vertebrae were also morphed on a part by part basis from the M50-O to the CAD of the F05-O. Similar to the cervical spine, cortical bone in the thoracic region was applied as shell elements with solid hexahedral elements representing trabecular bone. Cortical thickness maps were applied to the shell elements of the ribs by collaborating researchers at the University of
Virginia using techniques that were applied for the development of the M50-O (Li et al. 2010). As there is currently no published literature specific to rib cortical thickness differences between males and females at each rib level, the same thickness maps developed for the M50-O were applied to the F05-O. Joints of the torso region, such as the costovertebral, acromioclavicular, and sternoclavicular joints, were modeled with 0-length, 6 degree-of-freedom beams with a prescribed stiffness about each axis of rotation and along each translational axis.

The abdominopelvic region of the F05-O was primarily developed using tetrahedral elements (Beillas and Berthet 2012). The liver, spleen, kidneys, gallbladder, and pancreas were modeled with solid elements and a layer of shell elements on the outer surface representing parenchyma. Due to severe compression of the abdomen in blunt impact scenarios, the remaining abdominal structures were modeled as airbags to increase model robustness. Also, the small intestine was modeled as a control volume due to its geometric complexity. During CAD development, an abdominal compartment was created from the seated uMRI data by surrounding all of the organs and abdominal fat deep to abdominal muscles. This compartment was utilized for the development of abdominal fat by employing Boolean relationships to characterize space within the compartment not occupied by organs. Where physiologically relevant, the abdominal fat was connected node-to-node with the underlying soft tissues to reduce contact requirements.

The flesh and muscle of the thoracoabdominal region were modeled as a continuous mesh with muscle CAD being used to assign element material properties rather than explicitly meshing CAD geometry. Using this approach, the flesh of the upper
extremities was meshed continuously from the outer surface down to the bone. The CAD data was then used to assign elements within the flesh to a material representing muscle. This approach was used in an effort to represent the heterogeneity of the region while reducing mesh discontinuities or the use of contacts. However, in order to maintain biofidelity of the structures across joints, at the shoulder where the muscles originate, CAD data was used to explicitly mesh muscle connections to the bone. With regards to the flesh of the thorax, the pectoralis major, latissimus dorsi, and rhomboid major were developed via element assignment. Because the more superficial abdominal muscles serve mainly for passive load transfer, specific abdominal muscles were grouped to simplify the region. For example, the rectus abdominis and rectus oblique muscles were combined into one part with the same material type. The psoas major is the one exception for muscle development in the abdomen. The psoas muscles were explicitly meshed using tetrahedral elements and connected node-to-node with the lumbar spine and the pelvis. The remaining pelvic musculature was explicitly meshed using tetrahedral elements.

The pelvis of the F05-O was meshed using solid tetrahedral elements to represent trabecular bone and triangular elements to represent cortical bone. Cortical thickness maps were applied to the shell elements of the ribs by collaborating researchers at the University of Virginia. Images of the F05-O thorax, abdomen, and pelvis, as well as techniques employed for musculature, can be seen in Figure F 3.
Figure F 3. Upper Left) Internal thoracic components of the F05-O model. Anterior muscles on the right side have been removed for visualization. Upper Middle) Cross-section of a rib showing mesh topology. Upper Right) Cross-section of humerus showing continuous mesh with bone, muscle, and flesh. Lower Left) Thoracabdominal Cavity. Lower Right) Bones and muscle of the pelvic region.

Lower Extremity

The lower extremity of the F05-O model was primarily meshed using hexahedral elements (Shin et al. 2012; Untaroiu and Shin 2013; Untaroiu et al. 2013). The diaphyses of the femur and tibia were modeled with solid cortical bone elements and a void representing the intramedullary canals. Due to restrictions based on minimum element size, the fibula was modeled with solid trabecular elements throughout the structure with shells representing cortical bone. The cruciate and collateral ligaments (ACL, PCL, MCL, LCL) were each modeled using solid hexahedral elements and connections to bones were made node-to-node. Cartilage of the knee was modeled from the corresponding bone surfaces using shell elements. The menisci were hexahedrally meshed from the CAD and connected node-to-node to the tibia.
The flesh and muscle of the lower extremity were continuously meshed to limit the number of required contacts. In the upper leg, muscles were grouped to have parts representing the quadriceps and the hamstrings. The muscles of the lower leg consisted of the grouped calf muscles (soleus and gastrocnemius) and the tibialis anterior (Figure F4). The approach to these muscles was similar to that of the upper extremity, where the main body of the muscle was developed by element assignment and the origin/insertion connections were developed explicitly from the CAD.

Figure F4. (a) Outer surface of the F05-O lower extremity. (b) Lateral view of lower extremity depicting element assignment approach for muscle development. (c) Cross-sectional view of the thigh illustrating the element assignment approach for muscle development. (d) Detail included in the F05-O knee joint.
VIII. APPENDIX G

Table G 1. Experimental PMHS specimen characteristics

<table>
<thead>
<tr>
<th>Experiment</th>
<th>N</th>
<th>M:F</th>
<th>Average Subject Age (years)</th>
<th>Average Subject Mass (kg)</th>
<th>Average Subject Height (cm)</th>
<th>Mass Scaled to Average Male?</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frontal Chest</td>
<td>19</td>
<td>14:5</td>
<td>59.4 ± 17.4</td>
<td>67.0 ± 13.1</td>
<td>174 ± 10</td>
<td>Yes</td>
<td>Lebarbe et al. 2012</td>
</tr>
<tr>
<td>Thoracoabdominal</td>
<td>5</td>
<td>4:1</td>
<td>54.2 ± 15.1</td>
<td>64.0 ± 9.4</td>
<td>169.1 ± 6.9</td>
<td>Yes</td>
<td>Viano et al. 1989</td>
</tr>
<tr>
<td>Lateral Hub</td>
<td>6</td>
<td>5:1</td>
<td>71 ± 17.5</td>
<td>68.8 ± 12.0</td>
<td>172.9 ± 7.0</td>
<td>Yes</td>
<td>Koh et al. 2005</td>
</tr>
<tr>
<td>Lateral Plate</td>
<td>2</td>
<td>2:0</td>
<td>54</td>
<td>66.7</td>
<td>168.5</td>
<td>Yes</td>
<td>Kemper et al. 2008</td>
</tr>
<tr>
<td>Abdominal Bar</td>
<td>3</td>
<td>2:1</td>
<td>88.3 ± 5.6</td>
<td>74 ± 23.3</td>
<td>167.7 ± 4.6</td>
<td>Yes</td>
<td>Hardy et al. 2001</td>
</tr>
<tr>
<td>Lateral Pelvis</td>
<td>4</td>
<td>3:1</td>
<td>70.3 ± 7.1</td>
<td>55.4 ± 18.3</td>
<td>166.8 ± 7.9</td>
<td>No</td>
<td>Bouquet et al. 1998</td>
</tr>
<tr>
<td>Drop Test</td>
<td>2</td>
<td>2:0</td>
<td>47</td>
<td>61</td>
<td>N/A</td>
<td>Yes</td>
<td>Stalnaker et al. 1979</td>
</tr>
<tr>
<td>Lateral Sled</td>
<td>3</td>
<td>2:1</td>
<td>65.6 ± 1.4</td>
<td>64.2 ± 0.6</td>
<td>168 ± 5.7</td>
<td>Yes</td>
<td>Cavanaugh et al. 1990</td>
</tr>
<tr>
<td>Rear Seat Sled</td>
<td>2</td>
<td>2:0</td>
<td>54</td>
<td>57</td>
<td>177</td>
<td>Yes</td>
<td>Forman et al. 2009</td>
</tr>
<tr>
<td>Frontal Sled</td>
<td>8</td>
<td>8:0</td>
<td>54.0 ± 4.9</td>
<td>75.52 ± 2.8</td>
<td>179.3 ± 1.16</td>
<td>No</td>
<td>Shaw et al. 2009</td>
</tr>
</tbody>
</table>

*Male to Female ratio

Table G 2. Scale factors applied to the F05-O model for each simulation

<table>
<thead>
<tr>
<th>Simulation</th>
<th>Signal</th>
<th>Scale Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frontal Chest</td>
<td>Force</td>
<td>1.28</td>
</tr>
<tr>
<td></td>
<td>Deflection</td>
<td>1.21</td>
</tr>
<tr>
<td></td>
<td>Force</td>
<td>1.13</td>
</tr>
<tr>
<td>Thoracoabdominal</td>
<td>Deflection</td>
<td>0.99</td>
</tr>
<tr>
<td>Lateral Hub</td>
<td>Force</td>
<td>1.28</td>
</tr>
<tr>
<td>Lateral Plate</td>
<td>Force</td>
<td>1.08</td>
</tr>
<tr>
<td>Abdominal Bar</td>
<td>Force</td>
<td>1.31</td>
</tr>
<tr>
<td>Drop Test</td>
<td>Force</td>
<td>1.30</td>
</tr>
<tr>
<td>Lateral Sled</td>
<td>Torso Force</td>
<td>1.28</td>
</tr>
<tr>
<td></td>
<td>Pelvis Force</td>
<td>1.28</td>
</tr>
</tbody>
</table>
Table G 3. ISO results and magnitude weight factors for establish total weighted score for the rear seat frontal sled impact.

<table>
<thead>
<tr>
<th>Signal</th>
<th>Corridor</th>
<th>Slope</th>
<th>Size</th>
<th>Phase</th>
<th>ISO score</th>
<th>Mag Weight Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper Shoulder Belt</td>
<td>0.74</td>
<td>0.53</td>
<td>0.79</td>
<td>0.94</td>
<td>0.74</td>
<td>0.56</td>
</tr>
<tr>
<td>Force</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lap Belt Force</td>
<td>0.73</td>
<td>0.54</td>
<td>0.90</td>
<td>0.74</td>
<td>0.73</td>
<td>0.44</td>
</tr>
<tr>
<td>Head X Displacement</td>
<td>0.92</td>
<td>0.90</td>
<td>0.97</td>
<td>0.70</td>
<td>0.88</td>
<td>0.70</td>
</tr>
<tr>
<td>Head Z Displacement</td>
<td>0.64</td>
<td>0.73</td>
<td>0.52</td>
<td>0.61</td>
<td>0.63</td>
<td>0.30</td>
</tr>
<tr>
<td>Shoulder X Displacement</td>
<td>0.66</td>
<td>0.73</td>
<td>0.26</td>
<td>0.72</td>
<td>0.61</td>
<td>0.94</td>
</tr>
<tr>
<td>Shoulder Z Displacement</td>
<td>0.57</td>
<td>0.49</td>
<td>0.00</td>
<td>0.08</td>
<td>0.34</td>
<td>0.06</td>
</tr>
<tr>
<td>Pelvis X Displacement</td>
<td>0.66</td>
<td>0.77</td>
<td>0.39</td>
<td>0.14</td>
<td>0.52</td>
<td>0.89</td>
</tr>
<tr>
<td>Pelvis Z Displacement</td>
<td>0.38</td>
<td>0.29</td>
<td>0.00</td>
<td>0.00</td>
<td>0.21</td>
<td>0.11</td>
</tr>
</tbody>
</table>

Figure G 1. Head CG kinematic response for the F05-O model in the frontal sled impact condition

*Note: Head displacements in the Y direction deviate from experimental results due to model head contact with the arm*
Figure G 2. T1 kinematic response for the F05-O model in the frontal sled impact condition

Figure G 3. T8 kinematic response for the F05-O model in the frontal sled impact condition
Figure G 4. L2 kinematic response for the F05-O model in the frontal sled impact condition
Table G. ISO results and magnitude weight factors used to establish total weighted score for frontal sled impact.

<table>
<thead>
<tr>
<th>Signal</th>
<th>Corridor</th>
<th>Shape</th>
<th>Size</th>
<th>Phase</th>
<th>ISO Total</th>
<th>Mag Weight Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Frontal Sled (Kinetics)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Upper Shoulder Belt</td>
<td>0.91</td>
<td>0.80</td>
<td>0.96</td>
<td>0.95</td>
<td>0.90</td>
<td>0.53</td>
</tr>
<tr>
<td>Lower Shoulder Belt</td>
<td>0.88</td>
<td>0.78</td>
<td>0.97</td>
<td>0.88</td>
<td>0.88</td>
<td>0.41</td>
</tr>
<tr>
<td>Lap Belt</td>
<td>0.46</td>
<td>0.49</td>
<td>0.20</td>
<td>0.67</td>
<td>0.46</td>
<td>0.06</td>
</tr>
<tr>
<td>R Knee X</td>
<td>0.62</td>
<td>0.36</td>
<td>0.62</td>
<td>0.87</td>
<td>0.62</td>
<td>0.76</td>
</tr>
<tr>
<td>R Knee Y</td>
<td>0.41</td>
<td>0.00</td>
<td>0.00</td>
<td>0.93</td>
<td>0.35</td>
<td>0.13</td>
</tr>
<tr>
<td>R Knee Z</td>
<td>0.52</td>
<td>0.41</td>
<td>0.33</td>
<td>0.65</td>
<td>0.49</td>
<td>0.11</td>
</tr>
<tr>
<td>L Knee X</td>
<td>0.76</td>
<td>0.55</td>
<td>0.84</td>
<td>0.83</td>
<td>0.75</td>
<td>0.73</td>
</tr>
<tr>
<td>L Knee Y</td>
<td>0.50</td>
<td>0.14</td>
<td>0.24</td>
<td>0.93</td>
<td>0.46</td>
<td>0.17</td>
</tr>
<tr>
<td>L Knee Z</td>
<td>0.59</td>
<td>0.44</td>
<td>0.54</td>
<td>0.65</td>
<td>0.56</td>
<td>0.10</td>
</tr>
<tr>
<td>Foot X</td>
<td>0.51</td>
<td>0.54</td>
<td>0.62</td>
<td>0.45</td>
<td>0.53</td>
<td>0.17</td>
</tr>
<tr>
<td>Foot Y</td>
<td>0.26</td>
<td>0.34</td>
<td>0.00</td>
<td>0.81</td>
<td>0.33</td>
<td>0.02</td>
</tr>
<tr>
<td>Foot Z</td>
<td>0.68</td>
<td>0.58</td>
<td>0.87</td>
<td>0.72</td>
<td>0.71</td>
<td>0.81</td>
</tr>
<tr>
<td>Head X</td>
<td>0.98</td>
<td>0.91</td>
<td>0.99</td>
<td>0.87</td>
<td>0.94</td>
<td>0.49</td>
</tr>
<tr>
<td>Head Y</td>
<td>0.69</td>
<td>0.66</td>
<td>0.00</td>
<td>0.78</td>
<td>0.56</td>
<td>0.27</td>
</tr>
<tr>
<td>Head Z</td>
<td>0.70</td>
<td>0.81</td>
<td>0.41</td>
<td>0.98</td>
<td>0.72</td>
<td>0.25</td>
</tr>
<tr>
<td>T1 X</td>
<td>0.95</td>
<td>0.84</td>
<td>0.97</td>
<td>0.88</td>
<td>0.92</td>
<td>0.67</td>
</tr>
<tr>
<td>T1 Y</td>
<td>0.60</td>
<td>0.55</td>
<td>0.00</td>
<td>0.00</td>
<td>0.35</td>
<td>0.19</td>
</tr>
<tr>
<td>T1 Z</td>
<td>0.31</td>
<td>0.71</td>
<td>0.00</td>
<td>0.60</td>
<td>0.38</td>
<td>0.14</td>
</tr>
<tr>
<td>T8 X</td>
<td>0.94</td>
<td>0.88</td>
<td>0.98</td>
<td>0.79</td>
<td>0.90</td>
<td>0.67</td>
</tr>
<tr>
<td>T8 Y</td>
<td>0.55</td>
<td>0.43</td>
<td>0.00</td>
<td>0.15</td>
<td>0.34</td>
<td>0.09</td>
</tr>
<tr>
<td>T8 Z</td>
<td>0.32</td>
<td>0.74</td>
<td>0.07</td>
<td>0.97</td>
<td>0.48</td>
<td>0.24</td>
</tr>
<tr>
<td>L2 X</td>
<td>0.89</td>
<td>0.75</td>
<td>0.96</td>
<td>0.82</td>
<td>0.86</td>
<td>0.62</td>
</tr>
<tr>
<td>L2 Y</td>
<td>0.61</td>
<td>0.63</td>
<td>0.47</td>
<td>0.43</td>
<td>0.55</td>
<td>0.13</td>
</tr>
<tr>
<td>L2 Z</td>
<td>0.27</td>
<td>0.63</td>
<td>0.00</td>
<td>1.00</td>
<td>0.43</td>
<td>0.25</td>
</tr>
</tbody>
</table>
### Table G.5. Hard tissue injury prediction for the F05-O v2.3 compared to experimental injury reports and M50-O v4.4 simulations

<table>
<thead>
<tr>
<th>Experiment</th>
<th>F05-O</th>
<th>M50-O</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Frontal Chest</strong></td>
<td>Specimens ranged from 0 – 14 rib fractures</td>
<td>See Note</td>
</tr>
<tr>
<td><strong>Thoraco-Abdominal</strong></td>
<td>947 (38 YO) — ribs 7, 8, 9(2)</td>
<td>R6, R7(2), R8</td>
</tr>
<tr>
<td></td>
<td>954 (66 YO) — ribs 6, 7(2), 8(2), 9, 10</td>
<td></td>
</tr>
<tr>
<td></td>
<td>956b (40 YO) — ribs 7, 8</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>RNY2 (64 YO) — no fracture</td>
<td></td>
</tr>
<tr>
<td><strong>Lateral Hub</strong></td>
<td>No fracture</td>
<td>No Fracture</td>
</tr>
<tr>
<td><strong>Lateral Plate</strong></td>
<td>9 posterior rib, 12 anterior/lateral rib, 1 on non-impacted side</td>
<td>Clavicle, R2, R3, R4, R5, R6, L6</td>
</tr>
<tr>
<td></td>
<td>Clavicle, 9 posterior rib, 12 anterior/lateral rib, 2 on non-impacted side</td>
<td></td>
</tr>
<tr>
<td><strong>Abdominal Bar</strong></td>
<td>GI3 (87 YO) — ribs 7–9, bilateral</td>
<td>R8, L8</td>
</tr>
<tr>
<td></td>
<td>GI4b (93 YO) — ribs 6–10, bilateral</td>
<td></td>
</tr>
<tr>
<td></td>
<td>GI6 (85 YO) — ribs 7 and 8, bilateral; ribs 6 and 7, left</td>
<td></td>
</tr>
<tr>
<td><strong>Lateral Pelvis</strong></td>
<td>LCB 01 (65 YO) — iliac-pubic branch</td>
<td>Superior and Inferior pubic rami (Left and Right), Acetabulum, Sacrum, Femoral Neck</td>
</tr>
<tr>
<td></td>
<td>LCB 02a (53 YO) — iliac-ischio-pubic ramus fractures</td>
<td>MAIS 2–3</td>
</tr>
<tr>
<td></td>
<td>LCB 03 (80 YO) — iliac &amp; ischio-pubic rami: iliac wing; femur</td>
<td></td>
</tr>
<tr>
<td></td>
<td>LCB 09 (65 YO) — iliac-pubic branch</td>
<td></td>
</tr>
<tr>
<td><strong>Drop Test</strong></td>
<td>Subject 106 (68 YO) — 15 rib fractures</td>
<td>Temporal fracture, Superior and Inferior pubic rami (Left and Right), Acetabulum</td>
</tr>
<tr>
<td></td>
<td>Subject 111 (52 YO) — 5 rib fractures</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Subject 155 (42 YO) — no rib fracture</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Subject 156 (42 YO) — 1 rib fracture</td>
<td></td>
</tr>
<tr>
<td><strong>Lateral Sled</strong></td>
<td>Subject 05 (67 YO) — chest w/ &gt; 5 flail ribs</td>
<td>Clavicle, L3 and Superior and Inferior pubic rami (Left and Right), Acetabulum, Femoral Neck</td>
</tr>
<tr>
<td></td>
<td>Subject 07 (66 YO) — flail chest w/ &gt; 5 flail ribs</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Subject 08 (64 YO) — bilateral flail chest</td>
<td></td>
</tr>
<tr>
<td><strong>Rear Seat</strong></td>
<td>Test 1262a: 13 rib fractures with flail chest, sternum, T12, L1</td>
<td>Maudible, Clavicle, R1, R2(2), R3, L1, L2, L4, L5, L6, L7</td>
</tr>
<tr>
<td></td>
<td>Test 1264b: 13 rib fractures with flail chest, sternum</td>
<td></td>
</tr>
<tr>
<td><strong>Frontal Sled</strong></td>
<td>Specimens ranged from 0-2 sternum fractures, 0 to 1 clavicle fractures, and 2 to 27 rib fractures</td>
<td>R1, R2(2), L2, L4, L5, L6, Clavicle</td>
</tr>
</tbody>
</table>

---

1: AIS not reported because flail injuries are not described in the literature
2: Denotes female subject
3: Only hard tissue injuries used for comparison
VIII. REFERENCES


Summers, L., Hollowell, W.T., and Prasad, A. (2001) Analysis of occupant protection provided to 50th percentile male dummies sitting mid-track and 5th percentile female dummies sitting full-forward in crash tests of paired vehicles with

Toyota (2010) Documentation of Total Human Model for Safety (THUMS) AM50 Pedestrian/Occupant Model


Watanabe, R., Katsuhara, T., Miyazaki, H., Kitagawa, Y., and Yasuki, T. (2012) Research of the relationship of pedestrian injury to collision speed, car-type,


Chapter IX: Conclusion

The presented work details the development and full body validation of the GHBMC 5\textsuperscript{th} percentile female finite element model. While this project was primarily focused on developing a tool for evaluating blunt injury risk, the steps to develop and validate this model led to several additions to the injury biomechanics literature. A summary of the publications that have been produced as a result of this work can be seen in Table 1. In order to develop a model truly representative of a seated 5th percentile female, the finite element model was based on CAD from a multi-modality image dataset of a prospectively recruited, representative subject. The development and assembly of geometries was enhanced by the collection of scans in supine, seated, and standing postures. To the author’s knowledge, no other existing small female models have been developed using posture-specific medical images. This is an important aspect of the work as research has shown significant differences between internal geometries in different postures. To enhance development, an external anthropometry dataset was also gathered to aid in model assembly. This dataset is the first of its kind, acquired for the explicit purpose of developing a full-body finite element model of the 5th percentile female. As part of the CAD development efforts, this work also created references for verifying the internal geometries of the small female. This includes a technique to develop CAD of long bones from clinical CT and a method to evaluate cortical thickness along the length of the bone as opposed to discrete cross-sections. Additionally, a dataset of thoracoabdominal organ volumes in small women was developed as a means to verify internal soft tissues.
The F05-O model consists of 981 parts, 2.6 million elements, 1.4 million nodes, and has a mass of 51.1 kg. The model was predominately meshed using structured hexahedral elements. In an effort to enhance the stability of the model, care was taken to meet stringent element quality criteria and to minimize the number of intersections and contacts. The number of contacts was minimized via node to node connections and element property assignment based on the underlying CAD data.

Another focus of this work was on validation techniques, specifically for models beyond the average male. This study has investigated the ability of various scaling techniques to scale biomechanical response data from a reference to a target anthropometry. As there is limited data in the literature specific to the small female, this was an important step to determine the best approach for validation. Also, the evaluation of scaling has applications that extend beyond human body modeling to experimental work with PMHS and ATD design. The results from this work indicated that no single method was appropriate for all situations. However, scaling based on a ratio of effective masses was found, in general, to be the most proficient at scaling a reference response to the target. Additionally, a study was conducted utilizing a human body model to evaluate the CORA and ISO/TS 18571 objective evaluation techniques. As these ostensibly objective techniques seek to quantitatively interpret the biofidelity of a model, the study was designed to determine if differences in the algorithms of the techniques would lead to significantly different interpretations of model performance. Using a biomechanical dataset produced by the human body model, significant differences were identified for each component of the two methods. Therefore, a survey of subject matter experts was conducted and suggests a mixed approach to reporting objective ratings, using the
magnitude method in CORA and the ISO shape and phase methods may be the most intuitive approach.

Ultimately, the model was validated against experimental data using two approaches: 1) post-hoc scaling of the model outputs using the approach applied in the respective experiment and 2) geometrically morphing the model to the target mass of the experimental corridors. Based on the findings of this study, it was quantitatively demonstrated that full body morphing can be a more effective means of assessing biofidelity than post-hoc data scaling.

Future work will focus on model applications for the purpose of assessing tissue specific injury and developing injury risk curves for the small female. While the work presented in this dissertation has focused on the development and validation of a small female occupant model, the multi-modality medical image set used to create the occupant model has also been applied for the development of a small female pedestrian. In the future, the approaches outlined in this project will be applied to the development of the GHBMC small female pedestrian model. Also, as part of the GHBMC project, simplified human body models have been developed to reduce run time while still maintaining the ability to accurately predict kinetics and kinematics. As part of that work, a simplified small female model was developed based on the geometries from the detailed model described in this study. In the future, internal geometries and techniques from this work may be used to supplement the simplified model development.
Curriculum Vitae

Matthew L. Davis
2171 Burke Meadows Road Apt 206 Winston Salem, NC, 27103
Cell: 919-920-4081 mattendav@wakewhealth.edu

EDUCATION

Virginia Tech – Wake Forest University, Winston-Salem, NC
Ph.D., Biomedical Engineering Exp. Spring 2016
Dissertation: Development and Full Body Validation of a 5th Percentile Female Finite Element Model
Advisor: Dr. F. Scott Gayzik

Virginia Tech – Wake Forest University, Winston-Salem, NC
M.S., Biomedical Engineering May 2014
Thesis: Development of a CAD Dataset of the 5th Percentile Female
Advisor: Dr. F. Scott Gayzik

University of North Carolina at Chapel Hill, Chapel Hill, NC
B.S., Biomedical Engineering May 2012

CURRENT ACADEMIC TITLE
PhD Candidate
Virginia Tech-Wake Forest Center for Injury Biomechanics (CIB)
Department of Biomedical Engineering, Wake Forest School of Medicine

RESEARCH OBJECTIVES
• To use computational modeling for injury biomechanics research aimed at reducing the social and economic costs of unintentional injury

HIGHLIGHTED SKILLS
• Computational biomechanics research: Dynamic nonlinear finite element analysis, analytical programming, and human body model development and validation
• Experimental biomechanics research: Biological tissue mechanical testing and data analysis

FELLOWSHIPS
Altair Graduate Student Fellow August 2015 – Present

JOURNAL PUBLICATIONS


7. Davis ML, Vavalle NA, Stitzel JD, Gayzik, FS. “A Technique for Developing CAD Geometry of Long Bones Using Clinical CT Data.” Medical Engineering and Physics, 37.11 (2015); 1116-1123.


**PEER-REVIEWED CONFERENCE PAPERS**


SELECTED CONFERENCE ABSTRACTS AND SCIENTIFIC EXHIBITS


5. Davis ML, Koya B, Schap, JM, Gayzik FS. “Comparison of Objective Rating Techniques for the Validation of Computational Human Surrogate Models.” Podium presentation at the Association for the Advancement of Automotive Medicine’s Student Symposium, Waikoloa, HI, September 2016.


15. Davis ML, Koya B, Gayzik FS. “Development of the GHBMC 5th Percentile Female Finite Element Model” Poster Presentation at the Annual BMES Symposium, Tampa, FL, October 2015


18. Davis ML, Stitzel JD, Gayzik FS. “A multi-modality imaging approach to generate a CAD dataset of the 5th percentile female for modeling applications.” Poster presentation at the Biomedical Engineering Society’s Annual Meeting, San Antonio, TX, October 2014.


**PROFESSIONAL MEETINGS AND TRAININGS**

1. LS-DYNA Fracture and Failure Training, Winston Salem, NC, March 2015
2. LS-DYNA Contacts Training, Winston-Salem, NC, January 2014
3. LS-DYNA Training, Troy, MI, August 2012
5. HyperWorks Training, Winston-Salem, NC, September 2014
6. HyperCrash Training, Winston-Salem, NC, December 2013

**RESEARCH AND WORK EXPERIENCE**

**Graduate Research Engineer**
School of Biomedical Engineering and Sciences Aug. 2012 – Present
Virginia Tech – Wake Forest Center for Injury Biomechanics, Winston-Salem, NC

**Computational Modeling**

1. Development and validation of the Global Human Body Model Consortium’s 5th percentile female finite element model
2. Validation of the Global Human Body Model Consortium’s mid-sized male FE model
3. Development of simplified versions of the Global Human Body Model Consortium’s finite element models
4. Validation and sensitivity analysis of the WIAMan ATD head-neck finite element model
**Orthopaedic Biomechanics**
1. Compared the biomechanical effects of fibular allograft to calcium phosphate fixation for split-depression tibial plateau fractures

**Undergraduate Research Assistant**
University of North Carolina at Chapel Hill May. 2010 – May 2012

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

**HONORS AND AWARDS**


*Stapp Car Crash Conference* 
John W. Melvin Student Award: 3rd Place Paper November 2016

*Johns Hopkins Applied Physics Lab REDD Award* November 2015

*Ohio State University Injury Biomechanics Symposium* 
Student Travel Award May 2015

*VT-WFU SBES Symposium* 
1st Place Master’s Presentation April 2014

*Rocky Mountain Bioengineering Symposium* 
Conference Chair’s Award April 2013

*University of North Carolina at Chapel Hill* 
Dean’s List 2010 – 2012
1st Place Presentation: UNC Human Movement Symposium February 2012

**COMPUTER SKILLS**

Programming Languages: Python, Matlab
Dynamics / Simulation Software: LS-Dyna
FEA Pre/Post Processing Software: HyperWorks, True-Grid, LS-PrePost, Oasys, ANSA
Image Analysis Software: Mimics, Amira
CAD: Rhinoceros, Geomagic Studio, SolidWorks
PROFESSIONAL SERVICE

GUEST EDITOR OF PEER REVIEWED JOURNAL
Special Issue of Biomedical Engineering and Computational Biology: Image Acquisition and Processing for Clinical Applications

SESSION CHAIR
ASME Verification and Validation, Las Vegas, NV May 2015
Uncertainty Quantification, Sensitivity Analysis, and Prediction

PROFESSIONAL MEMBERSHIPS

Biomedical Engineering Society
President, VT – Wake Forest Chapter 2014 – 2015
Treasurer, VT – Wake Forest Chapter 2013 – 2014
Association for the Advancement of Automotive Medicine 2016 – Present
American Society of Mechanical Engineers 2015 – Present