Driver Injury Metric and Risk Variability as a Function of Occupant Position in Real World Motor Vehicle Crashes

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# Table of Contents

Acknowledgements ................................................................................................................. ii

Table of Contents .................................................................................................................. iii

List of Tables ................................................................................................................................. v

List of Figures .............................................................................................................................. vi

Abstract ....................................................................................................................................... viii

CHAPTER I: INTRODUCTION & BACKGROUND ................................................................. 1

Motor Vehicle Crash Injuries ........................................................................................................ 1

Experimental and Computational Injury Biomechanics ............................................................ 2
  The Total Human Model for Safety ............................................................................................ 3
  Head Injury Criterion .................................................................................................................. 3
  Combined Thoracic Index .......................................................................................................... 4
  Maximum Femur Force ............................................................................................................. 5
  Strain-Based Lung Injury Metrics ............................................................................................. 5

Real World Motor Vehicle Crash Reconstruction .................................................................. 7
  Crash Injury Research and Engineering Network ................................................................ 7
  Real World Crash Injury Prediction ......................................................................................... 7
  Development of Novel Injury Metrics and Risk Functions using FE Reconstructions .......... 8

Chapter Summaries ................................................................................................................... 13
  Chapter II: Optimization of a Simplified Automobile Finite Element Model Using Time Varying Injury Metrics ...................................................................................... 13
  Chapter IV: Summary of Research .......................................................................................... 13

References ................................................................................................................................. 14

CHAPTER II: OPTIMIZATION OF A SIMPLIFIED AUTOMOBILE FINITE ELEMENT MODEL USING TIME VARYING INJURY METRICS .................................... 17

Abstract ....................................................................................................................................... 18

1. Introduction ............................................................................................................................. 19

2. Methods .................................................................................................................................. 21
  2.1 Development of a Generic Reduced Vehicle FEM ................................................................. 21
  2.2 Anthropomorphic Test Device Positioning ........................................................................... 22
  2.3 Optimization Criteria Development ....................................................................................... 24
  2.4 Variation Study Setup ........................................................................................................... 26
LIST OF TABLES

Table 1. The lowest comprehensive error simulations for each iteration ...................... 29

Table 2: Vehicle parameter ranges for variation study to tune the SVM restraint systems in the Camry and Cobalt cases to NCAP crash test data. The vehicle parameters selected to tune the SVM for each case are bolded. ................................................................. 43

Table 3. Occupant positioning parameter ranges for variation study to simulate various pre-crash occupant positions and postures in the Camry and Cobalt cases ......................... 45

Table 4. Summary of calculated injury metrics and risks for all real world MVC reconstruction simulations with different occupant positions, including baseline simulations positioned according to the CIREN case descriptions. ................................. 48

Table 5. Publication plan for research outlined in this thesis. ...................................... 62
**LIST OF FIGURES**

Figure 1: An example strain threshold corresponding to a known volume of contused lung tissue. .......................................................... 6

Figure 2. (Left) For each real world injury risk function developed, several CIREN cases will be selected containing injury and non-injury crashes. (Center) These CIREN cases will be reconstructed using the methods presented in Chapters 2 and 3. (Right) The calculated injury metrics will be used to subsequently develop novel injury risk curves for metrics measured in THUMS based on real world crash data. ................................. 10

Figure 3: (a) An example of the PC maximum principal strain vs. lung volume relationship using the baseline simulation. (b) An example (baseline simulation) THUMS lung with the elements exceeding the threshold strain value highlighted. .................. 12

Figure 4. Overview of the simulation process: A.) development of a reduced vehicle configuration from target vehicle data and literature review, B.) response validation using a Hybrid III ATD and a matched NHTSA crash test, and C.) CIREN case simulation, with a variation study, using the THUMS as the occupant................................. 20

Figure 5. Diagram of the steering wheel parts that control shear bolt failure and steering column stroke. ...................................................... 22

Figure 6. (a) The abbreviations for the ATD positioning. (b) The experimental test and simulated model pre-collision configuration of the H3 ATD [5]. ................................. 23

Figure 7. (a) The simulation ATD position closely resembles (b) the test ATD position [5]................................................................. 23

Figure 8. Parameter distribution plot used to identify parameter ranges that resulted in more optimized iterations of the Camry Model. Areas of white (A) represent error termination simulations. Dark blue regions, such region B, indicate simulations ranges of values for each variable that are expected to be more optimal. .................................. 28

Figure 9. (a) Before the knee airbag characteristics were optimized, the force on the femur was a large magnitude, short duration spike. (b) Following the optimization procedure, the magnitude of the femur forces in the regulatory test and simulation matched, but the duration of the contact force was shorter in the simulation than the test. ........................................................................................................ 29

Figure 10. Each case reconstruction involved three phases: (I) Parameters related to the frontal airbag (green), seatbelt (blue), steering column (red) and knee airbag/knee bolster (purple) were varied to tune the H3 response of the head, chest, pelvis, femur and seatbelts. (II) Positioning THUMS by (A) shifting the SVM with respect to the seat in order to simulate a change in seat track position and (B) rotating the seat back angle around a pivot point defined on the seat. (III) Applying the crash pulse derived from the CIREN case EDR to the tuned SVM with THUMS positioned. ........................................ 41
Figure 11. HIC15 and CTI as a function of five positioning variables in the Camry and Cobalt cases. HIC15 and CTI values in the baseline simulation that most closely matched the occupant position documented in CIREN are indicated by the red star. .................................. 49

Figure 12. Head injury risk (AIS 1+, 2+, 3+) as a function of HIC15 (NHTSA, 1995) for all simulated positions in the Camry and Cobalt cases. Head injury risk in the baseline simulation that most closely matched the occupant position documented in CIREN is highlighted with darker shading. ............................................ 49
ABSTRACT

Motor vehicle crashes (MVCs) are a worldwide public health concern, resulting in annual totals of approximately 1.24 million deaths and 20-50 million injured occupants. Real world crash reconstructions using finite element (FE) vehicle and human body models (HBMs) have the potential to elucidate injury mechanisms, predict injury risk, and evaluate injury mitigation system effectiveness, ultimately leading to a reduced risk of fatality and severe injury in MVCs. The purpose of the work presented herein was to create a novel framework for FE frontal MVC reconstruction and injury analysis considering two primary constraints: (1) a shortage of specific FE vehicle models and (2) uncertainty in the case occupant’s position immediately before the crash event.

The novel reconstruction process was developed to address these two constraints using a pair of subsequent studies presented herein as individual chapters. First, a generic simplified vehicle model was developed and tuned to mimic the frontal crash environment of a specific vehicle model using crash test data. Subsequently, FE reconstructions of two CIREN frontal crash events were performed. Regional level injury metrics based on occupant kinematics were implemented into the Total HUman Model for Safety (THUMS) and analyzed to predict injury risks as a function of the occupant’s pre-crash position within the occupant compartment.

The results from these studies demonstrate the ability to reconstruct a wide array of real world frontal MVCs and predict regional level injury risks and variability due to occupant position. The novel MVC reconstruction paradigm will facilitate future injury metric and risk function development based on living human subjects in real world MVCs.
Chapter I: Introduction & Background

Motor Vehicle Crash Injuries

Motor vehicle crashes (MVCs) are a serious public health concern, both domestically and internationally. In 2013, over 32,000 people died in MVCs while an additional 2.3 million Americans sustained MVC injuries (NHTSA, 2014). According to the World Health Organization’s 2013 Global Status Report on Road Safety, MVCs accounted for approximately 1.24 million deaths annually, while an additional 20-50 million MVC occupants sustained non-fatal injuries (WHO, 2013). Understanding real MVC injury mechanisms leads to improved occupant restraint system performance and subsequently a reduced risk of fatality and severe injury (NHTSA, 2014). Analysis of real world MVC data allowed researchers to evaluate the effectiveness of occupant restraint systems (Gabauer, et al., 2010; Griffin, et al., 2012; Loftis, et al., 2011).

Motor vehicle collisions are often classified by crash mode. Each crash mode is uniquely characterized by vehicle crash characteristics and consequential injury mechanisms and patterns. Common crash characteristics used to classify crash modes include the primary direction of force (PDOF) and the location of impact with respect to the crash vehicle. When classified by PDOF and location of impact, the most common mode in injurious MVCs is the frontal crash mode. In 2013, over 55% of injurious MVCs in the United States were classified as frontal crashes (NHTSA, 2014). The purpose of the research described herein was to develop a protocol to allow researchers to computationally reconstruct real world frontal MVCs and analyze the corresponding sensitivity of predicted injuries as a function of the case occupant’s initial position within the occupant compartment.
EXPERIMENTAL AND COMPUTATIONAL INJURY BIOMECHANICS

A focus of experimental injury biomechanics has been to develop and evaluate metrics capable of predicting injury in impact loading conditions, such as the MVC environment. Many of these injury metrics were developed using post-mortem human subjects (PMHSs) instrumented with accelerometers, load cells, chest bands and pressure sensors in controlled laboratory experiments (Eppinger, 1989; White, et al., 2009; Yoganandan, et al., 2009). These metrics were used to develop biofidelic human-surrogate anthropomorphic test devices (ATDs) for impact testing conditions. Two common ATDs used in blunt trauma research within the automotive industry include the Hybrid III (H3) and THOR. Injury risk functions, describing the probability of injury, were calculated using statistical relationships between the injury metric measurements and damage to PMHSs during impact tests.

Although experimental PMHS sled tests and ATDs have traditionally performed well to predict kinetic and kinematic occupant injury metrics in sled and crash tests, modern human body finite element models (FEMs) have the ability to predict additional organ-level injury metrics and risks (Shigeta, et al., 2009). While FEMs of anatomical regions and individual organs were useful to develop novel organ-level injury metrics (Gayzik, et al., 2007; Stitzel, et al., 2002; Takhounts, et al., 2008), human body models (HBM) are useful to simulate whole body impact events such as MVCs (Vavalle, et al., 2013). HBM are advantageous to assess entire body impact events because they allow researchers to study occupant kinematics, bone strains, and internal soft tissue organ pressures simultaneously (Hayashi, et al., 2008).
The Total Human Model for Safety

One common HBM capable of predicting organ-level injury in whole body impact simulations is the Total HUman Model for Safety (THUMS) (Shigeta, et al., 2009). THUMS version 4.0 contains 1.80 million elements and 630,000 nodes capable of measuring forces and deformations distributed throughout the human body. THUMS was previously used to simulate controlled laboratory experiments and reconstruct real world crash events using the explicit, nonlinear, transient dynamic finite element LS-Dyna solver (LSTC, Livermore, CA). These simulated laboratory studies included evaluations of the injury potential for different astronaut suit configurations (Danelson, et al., 2011), the effects of seatbelt location on bilateral carotid artery injuries in far side impact (Danelson, et al., 2009), and a comparison of the biomechanical responses of THUMS and the H3 ATD under various belt and airbag configurations in a frontal crash environment (Mroz, et al., 2010).

Several common injury metrics developed within the field of experimental injury biomechanics have been adapted and implemented to THUMS for analysis in computational simulations.

Head Injury Criterion

The Head Injury Criterion (HIC) was developed to assess head injury risk in frontal impacts (Eq. 1). HIC was derived from the linear resultant acceleration measured during embalmed PMHS head drop tests onto rigid and padded surfaces (Hodgson, et al., 1977; Lissner, et al., 1960).

\[
HIC = \max \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) \, dt \right\}^{2.5} (t_2 - t_1) \]

Eq. 1
In this equation, \( t_1 \) and \( t_2 \) are the initial and final time of the HIC calculation and \( a(t) \) is the resultant linear acceleration. HIC15 and HIC36 are calculated by limiting the difference between \( t_1 \) and \( t_2 \) to 15 ms and 36 ms, respectively (\( t_2 - t_1 \leq 15 \) or 36 ms). HIC15 is more commonly used in modern automotive injury biomechanics to assess injury risks than HIC36 and is used within this study. Tri-axial acceleration data required to calculate HIC was measured in THUMS using a nodal accelerometer located at the head’s center of gravity (CG) and constrained to the skull using previously defined accelerometer methods for THUMS (Golman, et al., 2015).

**Combined Thoracic Index**

The Combined Thoracic Index (CTI) is an injury metric that combines the spinal acceleration and chest deflection (Eq. 2) to account for injuries caused by excessive force or excessive deflection (Eppinger, et al., 1999). The CTI was shown to be a better predictor of chest injury than chest acceleration or chest deflection independently, as it weights the contributions of deformation and force equally in injury prediction (Eppinger, et al., 1999). The CTI metric promotes equal contributions of deflection and acceleration to the overall metric by putting bounds on the maximum allowable chest deflection and acceleration.

\[
CTI = \frac{A_{\text{max}}}{A_{\text{int}}} + \frac{D_{\text{max}}}{D_{\text{int}}} \tag{Eq. 2}
\]

In this equation, \( A_{\text{max}} \) is the 3 ms clip value of spinal acceleration, measured at the first thoracic vertebrae (T1), \( D_{\text{max}} \) is the maximum chest deflection, and \( A_{\text{int}} \) and \( D_{\text{int}} \) are critical chest acceleration and deflection intercept values scaled to subject occupant size (Eppinger, et al., 2000).
Sternal deflection was measured using simulated chest bands in the THUMS model. Three chest bands, identified as upper, middle, and lower, were implemented within THUMS at the levels of the 4th, 8th, and 10th ribs to mimic instrumented PMHSs, respectively (Eppinger, 1989; Golman, et al., 2015; Pintar, et al., 1997). Peak chest deflections from five chest regions were evaluated for CTI calculations: left, center, and right deflection at the level of the 4th rib, and left and right deflection at the level of the 8th rib (Eppinger, et al., 1999). The maximum of the five chest deflection measurements is defined as $D_{\text{max}}$ in Eq. 2 above.

**Maximum Femur Force**

Several experimental biomechanics studies indicate that axial femur load is a good predictor of knee, thigh and hip (KTH) injuries (Kuppa, et al., 2001; Morgan, et al., 1990). Cross-sections of elements orthogonal to the axial dimension of each femur were defined in THUMS to simulate load cells. The location of this cross section was chosen to match the location of load cells instrumented in PMHSs for sled tests (Kuppa, et al., 2001).

**Strain-Based Lung Injury Metrics**

While the aforementioned injury metrics can be evaluated in HBM$s$ and in ATD$s$, one unique advantage of using HBM$s$ such as THUMS is the ability to develop and evaluate specific organ-level strain based injury metrics. One such injury metric related to the work presented quantified pulmonary contusion (PC). Previous work has quantified PC in animal models and subsequently the injury metric model was applied to THUMS.

PC was quantified using the strain developed in individual elements of FE lung models. The methods used in human models were adapted from an FEM of a rat lung in
experimental impact tests (Gayzik, et al., 2007). The volume of PC in each rat lung was measured from CT scan evaluations. Several injury metric criteria were evaluated, including the maximum principal strains and the product of the maximum principal strain and maximum principal strain rate (Gayzik, et al., 2011). These injury metric criteria and element volume were calculated in each element of the lung model throughout the duration of each simulation. The peak injury metric value in each element during the simulation was extracted. Subsequently, the elements were sorted in order of decreasing peak injury metric value, and the cumulative lung volume for all elements with a greater peak injury metric value was calculated and plotted as a percent of the total lung volume (reconstructed in Figure 1). Using a known volume of contused tissue from CT scans of the lung, maximum principal strain thresholds required to injure the known lung tissue volume were estimated.

Figure 1: An example strain threshold corresponding to a known volume of contused lung tissue.
REAL WORLD MOTOR VEHICLE CRASH RECONSTRUCTION

Crash Injury Research and Engineering Network

The National Highway and Traffic Safety Administration (NHTSA) established the Crash Injury Research and Engineering Network (CIREN) to evaluate injury causation in detailed MVC investigations. The CIREN combines teams of medical and engineering professionals to improve the prevention, treatment, and rehabilitation of motor vehicle crash injuries to reduce deaths, disabilities, and human and economic costs. For each crash investigation case, a crash investigator evaluates the damage to the case vehicle and the case occupant consents to release their medical information and radiology collected at the CIREN trauma center. The severity of reported case occupant injuries are triaged using the Abbreviated Injury Scale (AIS), where each injury is assigned a severity index ranging from 1-6, with increasing level of severity (AAAM, 2008).

The CIREN team attributes each injury to a specific mechanism of the crash event based on crash investigation evidence and medical reports. Due to the varying levels of confidence in these predictions, these mechanisms are coded in the CIREN database as Certain, Probable, Possible or Unknown. These uncertainties stem from a paucity of direct evidence between the injury and the case vehicle. One factor capable of obscuring injury mechanism predictions is the uncertainty in the case occupant’s position within the vehicle immediately before the crash event. The resulting case vehicle, occupant medical information, and injury mechanism predictions are available within the CIREN database.

Real World Crash Injury Prediction

While computational reconstruction of controlled laboratory experiments provides a wealth of injury risk information for a known set of prescribed boundary and restraint
conditions, the reconstruction of real world MVCs allows researchers to study the biomechanics of living human subjects and can account for a variety of human and environmental factors. The CIREN database is one source of real world MVC data that has been used for computational MVC reconstructions. Aortic rupture modes (Belwadi, et al., 2012) and pulmonary contusion injuries in near-side crashes (Danelson, et al., 2015) have been studied data from the CIREN database. However, only one study has comprehensively analyzed the HBM response to predict injury risks across the entire body in a real world side impact CIREN case (Golman, et al., 2014). Although the study established a protocol to evaluate injury risks across body regions of the HBM, only one case featuring a 2001 Ford Taurus was studied due to a paucity of full vehicle FEMs openly available to the research community. The purpose of the work presented herein was to develop a protocol capable of predicting injuries in real world frontal MVCs despite the paucity of available FE vehicle models and the uncertainty associated with the pre-crash occupant position in CIREN cases.

**Development of Novel Injury Metrics and Risk Functions using FE Reconstructions**

The methods developed and validated within this study serve as the foundation for a larger overall study. The MVC occupants studied within the CIREN database are considered living human volunteers for the severe impact loading conditions within a motor vehicle crash environment; a scenario that would not be ethically viable in a traditional laboratory environment. Many injuries are identified primarily through physiologic responses to blunt impact mechanical insult. These injuries are best studied in living tissue and therefore are most frequently studied in animal models; however CIREN cases allow researchers to correlate injury mechanisms to biomechanical injury metrics in human subjects in a common trauma environment.
Recent studies within the field of sports biomechanics have used instrumented volunteer subjects for injury biomechanics analysis. The helmets of professional, collegiate, high school and youth football players have been instrumented with accelerometer systems in an attempt to correlate head acceleration to traumatic brain injury and concussion; an injury that only manifests in living subjects (Funk, et al., 2007; Funk, et al., 2012; Rowson, et al., 2012). Using accelerative loading conditions that have been shown to increase risk of concussion in these volunteer subjects, computational models of the head and brain have been simulated to study the mechanical loading and deformation of brain structures and tissue during a blunt trauma impact event (Takhounts, et al., 2008). Based on these simulations, new injury metrics and risk functions have been developed to improve injury prediction sensitivity (Takhounts, et al., 2013; Takhounts, et al., 2008). Ultimately, an improved understanding of the complex biomechanical insult and the development of new injury metrics and risk functions allowed engineers to target the appropriate injury metric thresholds in the research and development of protective equipment such as football helmets (Rowson, et al., 2014).

These research strategies are mirrored within the automotive biomechanics research community. Impact loading conditions of human volunteers in the CIREN study can be defined with a high level of certainty. The geometry of the crash environment for each CIREN case is defined by the case occupant’s age, height, weight, seatbelt usage, airbag deployment, and vehicle model. With technological advances, newer vehicle models are required to have event data recorder (EDR) devices that provide information about seatbelt usage, airbag deployment timing, and a report of the delta-V signal corresponding to each crash event. The delta-V crash pulse allows dynamic loading conditions to be applied to the geometry of the crash environment. Using the geometry
and loading conditions from the CIREN report, FE reconstructions of real world MVCs were simulated within the work presented. Subsequently injury metrics for the finite element reconstructions were evaluated.

The opportunity to implement novel injury metrics within the crash simulations will be explored in the future. For a specific injury or injury mechanism, cases classified by injury or non-injury will be selected for reconstruction. These injury and non-injury cases will be reconstructed using the process described in Chapters 2 and 3. Subsequently, data from these reconstructions will be aggregated to develop injury risk curves and thresholds. The overall process to develop and evaluate novel injury metrics and risk functions is summarized in the flowchart in Figure 2. This example highlights the process that would be used to evaluate pulmonary contusion risks in frontal crashes using THUMS in the FE reconstructions.

Figure 2. (Left) For each real world injury risk function developed, several CIREN cases will be selected containing injury and non-injury crashes. (Center) These CIREN cases will be reconstructed using the methods presented in Chapters 2 and 3. (Right) The calculated injury metrics will be used to subsequently develop novel injury risk curves for metrics measured in THUMS based on real world crash data.
The techniques to quantify PC thresholds developed by Gayzik et al and applied to nearside impact CIREN case reconstructions by Danelson et al, were applied to the frontal CIREN reconstructions described in Chapter 3 with the intent to quantify occupant PC and to develop injury risk curves based on real world data in the future. These techniques have been adopted to estimate PC within the left lung of one injury case and assess the corresponding variability of the identified PC threshold as a function of the pre-crash occupant configuration. Segmentation of contused pulmonary tissue in the CIREN volunteer’s chest CT scan indicated that 49% of the occupant’s lung volume was injured. For each different occupant position configuration, the PC strain-volume relationship was evaluated, and the threshold corresponding to 49% of the lung volume being contused was extracted. Subsequently, the elements exceeding the identified strain threshold were highlighted in gray for each simulation (Figure 3a). The distribution of strain threshold values across all occupant position configurations is shown in Figure 3b. The baseline simulation, using an estimated occupant position according to the CIREN case report, is highlighted in dark green (threshold value = 0.321). The distribution is bi-modal, however most simulations had threshold values fall between 0.30 and 0.35, consistent with the 0.34 threshold define by Gayzik et al, 2011.

Using injury metrics, and threshold ranges, injury risk curves will be developed by aggregating data from the injury and non-injury CIREN case reconstructions. Strain based injury metrics, such as the PC injury model would result in unique injury risk functions because they could be developed from data that falls on a spectrum. Most injury risk functions evaluate injuries on a binary scale, whereas the strain based injury metrics allow organ level injury to be estimated on a continuous spectrum by. Several
additional soft tissue organs can be investigated as part of the overall study including brain, lung, spleen and liver. Additionally, specific fracture patterns within the vertebrae and pelvis will be studied at an elemental level during the frontal MVC event. The overall study will allow new injury metrics and corresponding risk functions to be developed and continuously updated as new real world crashes are added to the datasets when CIREN evaluates new cases.

![Graph showing lung volume relationship](image)

Figure 3: (a) An example of the PC maximum principal strain vs. lung volume relationship using the baseline simulation. (b) An example (baseline simulation) THUMS lung with the elements exceeding the threshold strain value highlighted.

The work presented in the subsequent chapters established a protocol to reconstruct and analyze FE simulations of real world frontal MVC crashes under the consideration of two major constraints: (1) a paucity of specific FE vehicle models and (2) uncertainty in the case occupant’s position immediately before the crash event. The sensitivity of regional level injury metrics and risks as a function of pre-crash occupant position was evaluated with the intent to be used in a larger overall study.
CHAPTER SUMMARIES

Chapter II: Optimization of a Simplified Automobile Finite Element Model Using Time Varying Injury Metrics

A method to optimize a generic vehicle interior model as a frontal crash environment for a specific vehicle model was developed.

Chapter III: Driver Injury Risk Variability in Finite Element Reconstructions of Crash Injury Research and Engineering Network (CIREN) Frontal Motor Vehicle Crashes

Two full frontal CIREN MVC cases were reconstructed using finite element vehicle model specific and human body models. An automated process to re-position the THUMS within the vehicle environment and quantify the injury response was developed.

Chapter IV: Summary of Research

A brief overview of work presented in this thesis.
REFERENCES


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Chapter II: Optimization of a Simplified Automobile Finite Element Model Using Time Varying Injury Metrics

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The following manuscript has been published in *Biomedical Sciences Instrumentation*. Stylistic variations are due to the requirements of the journal.
**Abstract**

In 2011, frontal crashes resulted in 55% of passenger car injuries with 10,277 fatalities and 866,000 injuries in the United States. To better understand frontal crash injury mechanisms, human body finite element models (FEMs) can be used to reconstruct Crash Injury Research and Engineering Network (CIREN) cases. A limitation of this method is the paucity of vehicle FEMs; therefore, we developed a functionally equivalent simplified vehicle model. The New Car Assessment Program (NCAP) data for our selected vehicle was from a frontal collision with Hybrid III (H3) Anthropomorphic Test Device (ATD) occupant. From NCAP test reports, the vehicle geometry was created and the H3 ATD was positioned. The material and component properties optimized using a variation study process were: steering column shear bolt fracture force and stroke resistance, seatbelt pretensioner force, frontal and knee bolster airbag stiffness, and belt friction through the D-ring. These parameters were varied using three successive Latin Hypercube Designs of Experiments with 130-200 simulations each. The H3 injury response was compared to the reported NCAP frontal test results for the head, chest and pelvis accelerations, and seat belt and femur forces. The phase, magnitude, and comprehensive error factors, from a Sprague and Geers analysis were calculated for each injury metric and then combined to determine the simulations with the best match to the crash test. The Sprague and Geers analyses typically yield error factors ranging from 0 to 1 with lower scores being more optimized. The total body injury response error factor for the most optimized simulation from each round of the variation study decreased from 0.466 to 0.395 to 0.360. This procedure to optimize vehicle FEMs is a valuable tool to conduct future CIREN case reconstructions in a variety of vehicles.
1. **INTRODUCTION**

The use of human body and vehicle finite element models (FEMs) to study real world collisions will allow engineers to account for sensitivity to differences in occupants, postures, and impact location to further mitigate crash injuries. Previously, a side impact CIREN case was reconstructed using the Total Human Model for Safety (THUMS) and two full vehicle finite element models. This was set up as a two part optimization problem. First, the side impact study optimized the computational simulation for damage to vehicle. Second, injury risks for the computational and NCAP test occupant were compared across multiple body regions [1].

Ultimately, the goal of this study was to reconstruct multiple real world frontal collisions described by the Crash Injury Research and Engineering Network (CIREN) database and model the documented injuries. However, a series of tools and processes needed development to study occupant injury mechanisms in frontal collision finite element (FE) simulations of real world crashes.

The biggest obstacle of computationally reconstructing many CIREN cases is the lack of FE vehicle models for the different vehicle makes and models in the CIREN crash database. Research groups such as the National Crash Analysis Center (NCAC) have independently developed a small library of FE vehicle models [2]. Although vehicle models from NCAC were sufficient for one side-impact CIREN case reconstruction, they will not be sufficient for the hundreds of possible vehicle makes and models selected for frontal analysis.

In order to address this problem, a strategy to create simplified vehicle models that closely mimic the characteristics of the real vehicles for each selected CIREN case
was developed. This strategy consisted of a two part process shown in Figure 4A and B. The first part (Figure 4A) included development of the simplified vehicle geometry and active vehicle components (i.e. seat belt, airbag and steering column) as defined by regulatory testing.

The second phase of the simplified vehicle model development (Figure 4B) was to fine tune the properties of key vehicle components using an optimization process. This optimization process simulated anthropomorphic test device (ATD) FEMs in hundreds of iterations of the vehicle model experiencing the acceleration field recorded during a regulatory test. The objective function for the optimization was formulated to represent the difference between injury metric curves derived from the computational simulations versus those from corresponding regulatory tests. Once these simplified vehicles have been optimized they can be used in conjunction with full human body FEMs, such as THUMS, to predict specific injuries to a variety of body regions as shown in Figure 4C. It will be important to streamline the approach to creating a library of simplified vehicle makes and models to study injury in CIREN cases.

Figure 4. Overview of the simulation process: A.) development of a reduced vehicle configuration from target vehicle data and literature review, B.) response validation using a Hybrid III ATD and a matched NHTSA crash test, and C.) CIREN case simulation, with a variation study, using the THUMS as the occupant.
This paper describes the development and optimization method of a single simplified vehicle model that mimics a 2011 Toyota Camry. The optimization method varied 7 to 10 vehicle parameters in three successive Latin Hypercube Designs of Experiments to optimize the overall injury response of the occupant. An algorithm based on several local injury metrics was developed to create a total body injury biofidelity ranking for each iteration of the regulatory test.

2. Methods

2.1 Development of a Generic Reduced Vehicle FEM

The first CIREN case selected for frontal collision analysis involved a 2010 Toyota Camry because of the crash configuration, the minimal vehicle intrusion, the occupant size, and the occupant injuries. The reduced Camry LS-DYNA FEM was developed using portions of the interior of NCAC’s Ford Taurus geometry and material properties [2]. The Taurus geometry was selected because it was NCAC’s most complete vehicle model that was similar to the Camry [3][4]. When parameters and material models were unknown, generic vehicle parameters were used for the baseline values in the variation studies.

Two active components were added to the model in order to better replicate the crash mechanics in a modern Camry. In order to appropriately model the steering column behavior in a frontal collision, the Taurus steering column was replaced with a column with simpler geometry. In a frontal collision, the impact force of the occupant on the steering column causes failure of shear bolts that connect the steering column to the frame of the vehicle. The spotweld connections between the steering column and the simplified vehicle frame were defined to fail when the force between the steering wheel
and steering column exceeded a threshold value. The new steering column allowed for 70 mm of displacement (stroke distance) along the axis of the steering column after the shear bolt failure.

![Diagram of the steering wheel parts that control shear bolt failure and steering column stroke.]

In addition, a simplified knee airbag (KAB) was modeled using a foam block with a rigid back that replaced the contoured Ford Taurus knee bolster. The material properties were based on industry conducted tests for similarity between a deploying knee airbag and a simplified knee airbag.

### 2.2 Anthropomorphic Test Device Positioning

In order to validate and optimize the simplified vehicle model, we compared the injury response of the 50th percentile male Hybrid III (H3) anthropomorphic test device in a real-world regulatory test to a finite element reconstruction of the same collision. This comparison ensured that the occupant in the reduced vehicle FEM was exposed to loading conditions similar to those in the full vehicle test. The regulatory test for this simulation came from the frontal New Car Assessment Program (FNACAP) Test #6750. The FNACAP report provided accelerometer data detailing vehicle accelerations throughout the test [5]. This acceleration pulse was applied to the FEM using prescribed acceleration boundary conditions.
The H3 has been modeled and validated by Humanetics in LS-DYNA. The H3 and dashboard components were positioned within the simplified model using pre-collision data for the 2010 Toyota Camry [5]. Eleven measurements reported in the crash report were used to modify the angles and locations of dashboard components, the seat, and ATD joint angles (Figure 6). The final positioning (shown in Figure 7) of the ATD was used in all simulation iterations because the dimensions were well defined in the FNCAP report.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Test</th>
<th>Sim.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Steering Wheel Angle (º)</td>
<td>SWA</td>
<td>65.0</td>
</tr>
<tr>
<td>Steering Column Angle (º)</td>
<td>SCA</td>
<td>25.0</td>
</tr>
<tr>
<td>Seat Back Angle (º)</td>
<td>SA</td>
<td>4.6</td>
</tr>
<tr>
<td>Nose to Rim Length (mm)</td>
<td>NR</td>
<td>422.0</td>
</tr>
<tr>
<td>Nose to Rim Angle (º)</td>
<td>NA</td>
<td>9.4</td>
</tr>
<tr>
<td>Chest to Dash (mm)</td>
<td>CD</td>
<td>535.0</td>
</tr>
<tr>
<td>Chest to Steering Hub (mm)</td>
<td>CS</td>
<td>322.0</td>
</tr>
<tr>
<td>Knee to Dash Length (mm)</td>
<td>KD</td>
<td>176.0</td>
</tr>
<tr>
<td>Knee to Dash Angle (º)</td>
<td>KDA</td>
<td>22.6</td>
</tr>
<tr>
<td>Pelvic Angle (º)</td>
<td>PA</td>
<td>24.0</td>
</tr>
<tr>
<td>Tibia Angle (º)</td>
<td>TA</td>
<td>45.1</td>
</tr>
</tbody>
</table>

Figure 6. (a) The abbreviations for the ATD positioning. (b) The experimental test and simulated model pre-collision configuration of the H3 ATD [5].

Figure 7. (a) The simulation ATD position closely resembles (b) the test ATD position [5].
2.3 Optimization Criteria Development

In order to validate the generic buck for Camry frontal crash simulations, four successive variation studies were conducted to determine the optimal geometric and material properties of the reduced vehicle FEM. The optimization process aimed to minimize the difference between the total body response in the real world test and the total body response in the simulated crash.

The total body response is quantified by seven time varying metrics: resultant head acceleration, resultant 6th thoracic vertebra (T6) acceleration, resultant pelvis acceleration, left and right femur forces, and shoulder and lap belt forces. With the exception of the belt forces, these metrics are instrumented in the H3 for all tests. The belt forces are measured by force transducers on the belt. The finite element output definitions for these metrics are defined in the Humanetics H3 user manual [6].

To compare the kinetics and kinematics of the ATD from the simulation to test data, the time history responses of each injury metric must be compared quantitatively. The model and experimental data were compared using a Sprague and Geers comparison [7]. This method calculates the magnitude ($M$) and phase ($P$) error factors between simulation and crash test signals. These error factors are calculated using Equations 1-5. The magnitude error factor accounts for differences in the shape and peaks of the injury metric data; whereas the phase error factor accounts for the timing of the injury metrics.

\[
M_{\text{metric}} = \sqrt{\frac{\theta_{mm}}{\theta_{ee}}} - 1 \quad \text{Eq. 1}
\]

\[
P_{\text{metric}} = \frac{1}{\pi} \arccos \left( \frac{\theta_{em}}{\sqrt{\theta_{mm} \theta_{ee}}} \right) \quad \text{Eq. 2}
\]
Where

\[ \theta_{mm} = \frac{1}{t_2-t_1} \int_{t_1}^{t_2} m^2(t) \, dt \]  
   Eq. 3

\[ \theta_{ee} = \frac{1}{t_2-t_1} \int_{t_1}^{t_2} e^2(t) \, dt \]  
   Eq. 4

\[ \theta_{em} = \frac{1}{t_2-t_1} \int_{t_1}^{t_2} e(t)m(t) \, dt \]  
   Eq. 5

Where \( m(t) \) corresponds to the metric data for the simulation model and \( e(t) \) corresponds to the metric data from the experimental test. The magnitude and phase error factors were calculated over a time period from \( t_1 \leq t \leq t_2 \). In the variation studies, the time range for these calculations was from 0 to 100 ms. After 100 ms, the peak loads and accelerations leading to injury in each simulation passed. The comprehensive error factor between corresponding signals combines the magnitude and phase errors as shown in Equation 6 [7].

\[ C_{\text{metric}} = \sqrt{M_{\text{metric}}^2 + P_{\text{metric}}^2} \]  
   Eq. 6

Based on these definitions, \( M, P \), and \( C \) are standardized measures of the difference between the signals of the experimental tests and simulation models. Smaller values represent more similar signals. The comprehensive error factors for each injury metric are combined into a total human response comprehensive error factor \((C_{Tot. \, Resp.})\) as seen in Equation 7. The average of the femur and belt forces were used in order to avoid excessive weighting of these metrics.

\[ C_{Tot. \, Resp.} = \sqrt{\left(\frac{C_{R \, Femur} + C_{L \, Femur}}{2}\right)^2 + \left(\frac{C_{\text{Lap \, Belt}} + C_{\text{Shoulder \, Belt}}}{2}\right)^2 + C_{\text{Head}}^2 + C_{T6}^2 + C_{\text{Pelvis}}^2} \]  
   Eq. 7
2.4 Variation Study Setup

To vary the injury response of the ATD in the FEM, seven to ten vehicle parameters were changed in the variation studies. The steering column parameters that were varied were the force at which the steering column shear bolts fracture and the force resisting steering column compression after shear bolt fracture. In addition, the flow rate of air to inflate the frontal airbag (FAB inflation) was varied. The four knee airbag parameters that were modified were the stress and strain scale factors for the stress-strain curve of the foam material, the damping factor for the foam material, and the thickness of the foam. Three seatbelt properties were varied: scale factors for the force on the pretensioner and retractor curves and friction of the shoulder seatbelt moving through the D-ring.

To study the effects of the vehicle parameters an initial one factor at a time (OFAT) study was conducted to setup parameter ranges for three subsequent Latin Hypercube Designs of Experiments studies. The Latin Hypercube Design (LHD) method varied each parameter linearly and independently of the values for the other variables within a predefined range. The LHD sampling method used for these variation studies was the optimumLHS script in R [8].

The first LHD (LHD1) consisted of 200 simulations that varied seven different vehicle parameters. Based on the observations from LHD1, a second variation study (LHD2) was conducted to narrow the range of the seven variables. Additionally, LHD2 added two new variables and consisted of 135 simulations. The third LHD (LHD3) narrowed the parameter ranges further and added a tenth parameter, for a total of 130 simulations.
Upon completion of each LHD, the simulation data was processed and the $C_{Tot. Resp}$ values for each simulation were ranked. To visualize the effect of each variable on the biofidelity of the simulation, a graphical representation (heat map) was developed to help visualize and narrow the subsequent LHD parameter ranges (Figure 8). For each parameter listed on the left hand side, the values of this variable were sorted in ascending order and the corresponding $C_{Tot. Resp}$ value for each simulation was stored. The range of $C_{Tot. Resp}$ scores was fit to a linear scale and mapped to a shade of blue in the graphic. The darker blue lines represent simulations that had lower $C_{Tot. Resp}$ scores which meant the simulation more closely matched the experiment. White lines correspond to simulations that did not run to completion due to errors. Based on the third row of Figure 8 (Pretensioner Force), simulations with the 35 highest pretensioner force curves caused errors (see location A.) The fourth row of Figure 8 (FAB Inflation S.F.) shows a region of darker blue located at location B. This darker region indicated that the subsequent LHD should tighten the range of the FAB inflation curve scale factor to yield a more optimized Camry model. The light blue seen in region C (KAB Stress S.F. row) in Figure 8 indicates that scaling the stress response to strain in the KAB yields a vehicle model that does not simulate the injuries seen in the regulatory test as closely as vehicle models with lower stress responses.

The simulation with the lowest total body response comprehensive error factor in the final variation study was selected as the best simplified Toyota Camry finite element model in frontal collision scenarios.
Figure 8. Parameter distribution plot used to identify parameter ranges that resulted in more optimized iterations of the Camry Model. Areas of white (A) represent error termination simulations. Dark blue regions, such region B, indicate simulations ranges of values for each variable that are expected to be more optimal.

3. RESULTS

For each subsequent variation study, the most optimized simulations continued to have improved overall comparison values between the experimental and simulated injury responses. Table 1 summarizes the comprehensive error factors for the top five simulations of each LHD iteration (a value of 0 would result if two identical curves were compared). Because the range of input parameters were narrowed for each LHD based on cases with poor results, fewer simulations failed in each successive iteration. The first iteration had 82/200 simulations fail, while the second iteration had 34/130 simulations fail and the final iteration had only 12/135 simulations fail.
Table 1. The lowest comprehensive error simulations for each iteration

<table>
<thead>
<tr>
<th>LHD 1 (200 Sim)</th>
<th>LHD 2 (130 Sim)</th>
<th>LHD 3 (135 Sim)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sim #</td>
<td>C</td>
<td>Sim #</td>
</tr>
<tr>
<td>-------</td>
<td>---</td>
<td>-------</td>
</tr>
<tr>
<td>122</td>
<td>0.466</td>
<td>117</td>
</tr>
<tr>
<td>22</td>
<td>0.477</td>
<td>81</td>
</tr>
<tr>
<td>186</td>
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<td>22</td>
</tr>
<tr>
<td>165</td>
<td>0.513</td>
<td>28</td>
</tr>
<tr>
<td>47</td>
<td>0.522</td>
<td>91</td>
</tr>
</tbody>
</table>

Figure 9 shows the transformation of the FEM left femur force prior to the first LHD to the most optimized simulation. As shown in Figure 9, the optimized simulation closely matched the magnitude; however, the duration of the simulation femur impact force was shorter than the duration of the impact force from the test data. This was characteristic of all the injury metrics except head acceleration. The simulated head acceleration pulse was longer than the test data pulse.

Figure 9. (a) Before the knee airbag characteristics were optimized, the force on the femur was a large magnitude, short duration spike. (b) Following the optimization procedure, the magnitude of the femur forces in the regulatory test and simulation matched, but the duration of the contact force was shorter in the simulation than the test.
4. **DISCUSSION**

The total body response comprehensive error factor derived from the Sprague and Geers comparison functions is a valuable metric to determine the validity of a simplified vehicle model in frontal collisions. The series of Latin Hypercube Designs of experiments allowed quick identification of a simplified model that closely matched the regulatory testing of a Camry.

This initial variation study to optimize occupant and vehicle interactions in a frontal Toyota Camry collision was valuable to identify the sensitivity of several individual vehicle components. By identifying key ranges of specific vehicle parameters and the parameters that have larger impacts on injury response, future simplified vehicle models can be optimized more efficiently. The KAB material properties were the most influential parameters studied and appeared to affect the injury metrics analyzed in each part of the body. The peak force on the femur was a result of contact with the knee airbag. The force in the femurs and the lap belt peaked nearly simultaneously just prior to 50 ms. However, the simulation loads were often incorrectly distributed between the belt and the knee airbag. Initially, it was predicted that increasing the scale factor on the retractor curve would help transmit more load to the belt than the femur. In addition, the increased femur force appeared to have an effect that was transmitted through the upper body; in the first two LHDs the peak chest and head accelerations were larger and later than the test data indicated. However, when the retractor properties did not improve the force distribution as expected, the KAB damping factor was studied closer. The damping factor for the KAB most significantly influenced the optimization of these injury metrics.
Upon visual inspection of the NCAC occupant and vehicle FEM validation, the magnitude and phase error for the full vehicle models is very similar to the error seen in the optimized reduced vehicle model. These similarities are a sign that the reduced vehicle model will be a valuable tool in frontal collision reconstruction. In the future we expect to quantify the Sprague and Geers coefficients for the frontal collision validations of the NCAC vehicle FEMs. This will provide a quantitative comparison of biofidelity between the optimized reduced model versus the complete finite element vehicle model.

The development of the reduced 2010 Camry Model acted as a proof of concept for a process to quickly develop a library of generic vehicle models that can be used in CIREN reconstruction cases. Based on this first case, the entire computational optimization of a new vehicle model can be completed within a 24 hour period following the initial positioning of the H3 ATD.

5. CONCLUSIONS

Optimizing a simplified finite element vehicle model to the injury metric responses of an ATD in a regulatory test appears to be a valid approach to create appropriate occupant loading in frontal collision reconstruction. After the model was optimized in three iterations of a Latin Hypercube Design of experiments, simulation injury metric curves more closely matched the test data in magnitude and phase. The error between simulation and test data was similar in the optimized reduced vehicle to the error in the data for the full vehicle models developed by NCAC [3]. Development of the optimization process allows for a streamlined approach to creating reduced models for a variety of vehicle makes and models. Using this optimized reduced 2010 Camry, a
human finite element model such as the Total Human Model for Safety can be implemented to model and reconstruct occupant loading for a CIREN case.

6. ACKNOWLEDGMENTS

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REFERENCES


Chapter III: Driver Injury Risk Variability in Finite Element Reconstructions of Crash Injury Research and Engineering Network (CIREN) Frontal Motor Vehicle Crashes

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ABSTRACT

Objective: A three phase real world motor vehicle crash (MVC) reconstruction method was developed to analyze injury variability as a function of pre-crash occupant position for two full-frontal Crash Injury Research and Engineering Network (CIREN) cases.

Method: Phase I: A finite element (FE) simplified vehicle model (SVM) was developed and tuned to mimic the frontal crash characteristics of the CIREN case vehicle (Camry or Cobalt) using frontal NCAP crash test data. Phase II: The Toyota HUman Model for Safety (THUMS) v4.01 was positioned in 120 pre-crash configurations per case within the SVM. Five occupant positioning variables were varied using a Latin Hypercube design of experiments: seat track position, seat back angle, D-ring height, steering column angle, and steering column telescoping position. An additional baseline simulation was performed that aimed to match the pre-crash occupant position documented in CIREN for each case. Phase III: FE simulations were then performed using kinematic boundary conditions from each vehicle’s event data recorder (EDR). HIC15, combined thoracic index (CTI), and femur forces were evaluated to predict regional-level injury risks.

Results: Tuning the SVM to specific vehicle models resulted in close matches between simulated and test injury metric data, allowing the tuned SVM to be used in each case reconstruction with EDR-derived boundary conditions. Simulations with the most rearward seats and reclined seat backs had the greatest HIC15, head injury risk, CTI and chest injury risk. Calculated injury risks for the head, chest, and femur closely correlated to the CIREN occupant injury patterns. CTI in the Camry case yielded a 54% probability of AIS 2+ chest injury in the baseline case simulation and ranged from 34 to 88% (mean
= 61%) risk in the least and most dangerous occupant positions. The greater than 50% probability was consistent with the case occupant’s AIS 2 hemomediastinum. The injury metrics evaluated for the Cobalt case occupant indicated a low risk of injury. The highest risk of injury was in the chest region. The baseline simulation estimated 33% risk of chest injury, while the Cobalt case occupant had an AIS 2 pulmonary contusion.

**Conclusions:** A method to compute injury metrics and risks as functions of pre-crash occupant position was developed and applied to two CIREN MVC FE reconstructions. The reconstruction process allows for quantification of the sensitivity and uncertainty of the injury risk predictions based on occupant position to further understand important factors that lead to more severe MVC injuries.

**Keywords:** Finite Element Modeling, Human Body Model, Latin Hypercube, Occupant Position, THUMS, Variation
1. INTRODUCTION

Approximately 1.24 million motor vehicle crash (MVC) deaths occur annually worldwide, while an additional 20-50 million MVC occupants sustain non-fatal injuries (WHO, 2013). The frontal crash mode is most common within the United States, accounting for 60% of fatal MVCs and 54% of injurious MVCs in 2012 (NHTSA, 2014). As the understanding of injury mechanisms in MVCs and the performance of occupant restraint systems improves it is expected that the risk of fatality and severe injury in MVCs will continue to decrease. Analysis of real world MVCs allows researchers to evaluate the effectiveness of occupant restraint systems under an array of variables not tested in a laboratory crash environment, potentially providing information that could better protect the occupant.

Although anthropomorphic test devices (ATDs) are capable of predicting occupant kinetics and kinematics in physical tests which have been correlated to injury risks, modern human body finite element models (FEMs) have the ability to predict more specific organ-level injury metrics and risks (Shigeta, et al., 2009). While FEMs of anatomical regions and individual organs have been useful to develop novel organ-level injury metrics (Gayzik, et al., 2007; Stitzel, et al., 2002; Takhounts, et al., 2008), human body models (HBMs) have been useful to simulate whole body impact events such as MVCs (Vavalle, et al., 2013). HBMs are advantageous to assess entire body impact events by allowing researchers to study occupant kinematics, bone strains, and internal soft tissue organ pressures/strains simultaneously (Hayashi, et al., 2008). One commonly used HBM capable of predicting organ injuries for whole body impact simulations is the Total HUman Model for Safety (THUMS) (Shigeta, et al., 2009). THUMS has been used
to simulate controlled laboratory experiments as well as reconstructions of real world

crash events (Danelson, et al., 2011; Danelson, et al., 2009; Danelson, et al., 2015;

Forman, et al., 2012; Golman, et al., 2015; Golman, et al., 2014; Iraeus, et al., 2014;

Iraeus, et al., 2013; Mroz, et al., 2010; Poulard, et al., 2015).

While controlled laboratory experiments provide injury risk information related to

a well-defined set of prescribed boundary and restraint conditions, simulations of real

world MVCs can account for a variety of human and environmental factors. The Crash

Injury Research and Engineering Network (CIREN) database is one source of real world

MVC data that has been used for computational MVC reconstructions (Belwadi, et al.,

2012; Danelson, et al., 2015). One previous study comprehensively analyzed the HBM

response to predict injury risks across the entire body in a side impact CIREN case, but

the reconstruction protocol used an open source 2001 Ford Taurus National Crash

Analysis Center (NCAC) full vehicle FEM and was thus limited due to the paucity of

open source full vehicle FEMs (Golman, et al., 2014).

One challenge of reconstructing CIREN cases is the uncertainty in the occupant’s

position and posture at the time of the crash (Danelson, et al., 2015). Although CIREN

collects information related to the occupant restraint mechanisms and positioning, this

data is collected post-crash and is subject to an inherent amount of uncertainty. The

purpose of this study was to establish a protocol to reconstruct a broader range of CIREN

frontal MVCs using FEMs and quantify the variability of injury risks associated with
different positioning and posture of the driver.
2. METHODS

2.1 Case Selection

Two full frontal CIREN MVCs with crash and occupants characteristics similar to regulatory crash tests were selected for reconstruction.

**Camry Case Details:** The first CIREN frontal crash reconstructed involved a 160 cm, 64 kg, 21 year old belted female driver in a 2010 Toyota Camry. The case vehicle struck a 1999 Jeep Cherokee at a 10° Principal Direction of Force (PDOF) with an EDR longitudinal delta-V of 64 km/h, resulting in a maximum crush of 62 cm (Collision Deformation Code (CDC) 12FDEW3). The driver frontal and knee airbags deployed. The occupant was documented to be seated between the forward-most and mid-track position with the seat back “slightly reclined” before the crash. The D-Ring anchorage was at the lowest position. The occupant sustained a left bimalleolar fracture (AIS 854464.3), hemomediastinum (AIS 442208.2), and AIS 1 neck, upper arm, shoulder, breast, chest, and abdominal contusions coded with AIS version 2005 update 2008.

**Cobalt Case Details:** The second CIREN frontal crash reconstructed involved a 183 cm, 77 kg, 80 year old belted male driver in a 2006 Chevrolet Cobalt. The case vehicle struck a 2002 Ford Expedition at 350° PDOF with an EDR longitudinal delta-V of 43 km/h, resulting in maximum crush of 58 cm (CDC code 12FDEW3). The driver frontal airbag deployed. The occupant was documented to be seated between the mid-track and rearward-most position with the seat back in an upright position. The D-Ring anchorage was at the lowest position. The occupant sustained pulmonary contusion with pneumothorax (AIS 441406.3), L1 and L3 burst fractures (AIS 650632.2), loss of
consciousness less than one hour (160202.2), and AIS 1 shoulder and skin contusions/abrasions coded with AIS version 1990 update 98.

2.2 Case Reconstruction Process

The reconstruction process of each CIREN case involved three distinct phases. Phase I involved establishing a vehicle FEM that was suitable for simulating the CIREN frontal crash by tuning the occupant restraint systems of a generic simplified vehicle model (SVM) using regulatory crash test data. In Phase II, a variation study was conducted to automatically position THUMS v4.01 within the tuned SVM in a range of occupant positions and postures. Phase III applied the crash pulse derived from the CIREN case vehicle’s event data recorder (EDR) and assessed injury risks for each potential occupant position. The three phases of the reconstruction process are illustrated in Figure 10. The finite element solver used for each phase of this study was LS-DYNA (MPP, Version 971, R6.1.1, LSTC, Livermore, CA) run on a computer cluster.

2.3 Phase I - simplified vehicle model (SVM) development and tuning

A FEM of a simplified vehicle was developed and tuned to accurately simulate frontal New Car Assessment Program (NCAP) crash tests of each CIREN case vehicle. The dashboard and seat of the SVM were based on the 2001 Ford Taurus NCAC model (Marzougui, et al., 2012) and the steering and frontal airbag were based upon the open source NCAC inflating airbag model (Bedewi, et al., 1996). A custom steering column, capable of compressing under large axial loads, was developed for the SVM. A calibrated foam material model was implemented to model the effects of a knee airbag in the Camry case only.
Figure 10. Each case reconstruction involved three phases: (I) Parameters related to the frontal airbag (green), seatbelt (blue), steering column (red) and knee airbag/knee bolster (purple) were varied to tune the H3 response of the head, chest, pelvis, femur and seatbelts. (II) Positioning THUMS by (A) shifting the SVM with respect to the seat in order to simulate a change in seat track position and (B) rotating the seat back angle around a pivot point defined on the seat. (III) Applying the crash pulse derived from the CIREN case EDR to the tuned SVM with THUMS positioned.
For each case, an NCAP frontal crash test of the case vehicle model or a sister or clone vehicle model was reconstructed (Anderson, 2013). The 50th Percentile male Humanetics H3 ATD FEM was positioned within the SVM according to steering column positioning, seat back angle, pelvis and tibia angles, and nose to rim, chest to steering hub, knee to dash, knee to knee, and ankle to ankle measurements reported in the NCAP report (NHTSA, 2005; NHTSA, 2011). The occupant restraint systems were parameterized by seven to ten variables corresponding to properties of the frontal airbag (inflation rate; vent area), seatbelt (pretensioner force; load limiter force; buckle friction coefficient), steering system (shear bolt fracture force; stroke resistance; rim stiffness), knee airbag (foam modulus, maximum strain, damping factor, and thickness), and the knee bolster stiffness (Table 2). Latin Hypercube design (LHD) of experiments sampling methods were used to assign restraint system parameter values to 200 simulations of the Camry NCAP test and 150 simulations of the Cobalt NCAP test (Stocki, 2005). The LHD is an effective space filling design used in design of experiment parameter studies. In a LHD, each parameter has as many levels as there are experiments in the design. The levels are spaced evenly from the lower bound to the upper bound of the parameter.

For each NCAP reconstruction, kinematic boundary conditions were derived from video tracking the rear and middle floor sill photo-targets from the crash test video to capture the longitudinal translation and pitching of the vehicle for 150 ms. Boundary conditions were applied to the SVM at a rigidly constrained photo-target shown in Figure 10-I, as pitch angle (about the Y-axis) displacement, and vertical (Z-axis) and longitudinal (X-axis) displacements.
Table 2: Vehicle parameter ranges for variation study to tune the SVM restraint systems in the Camry and Cobalt cases to NCAP crash test data. The vehicle parameters selected to tune the SVM for each case are bolded.

<table>
<thead>
<tr>
<th>Restraint System</th>
<th>Parameter</th>
<th>Units</th>
<th>2010 Toyota Camry</th>
<th>2006 Chevrolet Cobalt</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Min</td>
<td>Selected</td>
</tr>
<tr>
<td>Frontal Airbag</td>
<td>Peak Inflation Rate</td>
<td>kg / s</td>
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<td>2.55</td>
</tr>
<tr>
<td></td>
<td>Vent Area</td>
<td>mm</td>
<td>800</td>
<td>1750</td>
</tr>
<tr>
<td>Seatbelt</td>
<td>Peak Pretensioner Force</td>
<td>N</td>
<td>2750</td>
<td>2950</td>
</tr>
<tr>
<td></td>
<td>Retractor Load Limiter Force</td>
<td>N</td>
<td>2700</td>
<td>3100</td>
</tr>
<tr>
<td></td>
<td>Belt Buckle Friction Coefficient</td>
<td>Unitless</td>
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<td></td>
</tr>
<tr>
<td>Steering Column/Wheel</td>
<td>Shear Bolt Fracture Force</td>
<td>N</td>
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<td>4750</td>
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<td></td>
<td>Stroke Resistance</td>
<td>N</td>
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<tr>
<td></td>
<td>Rim Modulus</td>
<td>MPa</td>
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<td></td>
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<tr>
<td>Knee Airbag (KAB) / Knee Bolster</td>
<td>KAB Modulus</td>
<td>kPa</td>
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<tr>
<td></td>
<td>KAB Maximum Strain</td>
<td>Unitless</td>
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<td>0.945</td>
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<td></td>
<td>KAB Damping Factor</td>
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<td>0.55</td>
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<td></td>
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<td>104</td>
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<tr>
<td></td>
<td>Bolster Modulus</td>
<td>MPa</td>
<td>N/A</td>
<td></td>
</tr>
</tbody>
</table>

Seven ATD and restraint system signals (head, chest, and pelvis accelerations; left and right femur and shoulder and lap belt forces) were compared between each simulated iteration of the crash test and the physical crash test as shown in Figure 10-I. The Sprague and Geers magnitude (M), phase (P) and comprehensive (C) error factors were calculated for the first 100 ms of each signal, which represented the portion of the crash event prior to the rebound phase (Sprague, et al., 2004). The comprehensive error factors for each signal were combined using Eq. (1) to evaluate the total body response ($C_{Body}$) error between each simulated crash test iteration and the physical test. Simulations with the lowest $C_{Body}$ error were identified as the most accurate sets of occupant restraint system parameters to reconstruct frontal crashes for a given vehicle model using the SVM.

$$C_{Body} = \sqrt{\left(\frac{C_{Femur} + C_{CL.Femur}}{2}\right)^2 + \left(\frac{C_{Lap.Belt} + C_{Shoulder.Belt}}{2}\right)^2 + C_{Head}^2 + C_{Chest}^2 + C_{Pelvis}^2}$$  \hspace{1cm} (1)
2.4 Phase II - human body model (HBM) scaling and positioning:

CIREN reconstructions were performed with the seated 50th percentile male THUMS v4.01 HBM (73.7 kg, 178.6 cm) (Shigeta, et al., 2009), updated from v4.0 to consider the influence of the occupant’s weight in a seated posture and improve overall model stability (Toyota Central R&D Labs, Inc. (TCRDL), Nagakute, Japan). THUMS v4.01 was length-scaled by the ratio of the case occupant’s height to the unscaled THUMS’s height (178.6 cm) for each case. Length scaling was selected rather than mass scaling to focus on the effects of occupant position and posture with respect to vehicle restraint systems. Scaling was achieved by length scaling occupant size isometrically along the X, Y, and Z-axes with no adjustments in occupant mass other than the mass changes due to size changes.

THUMS was translated and rotated using the LS-PrePost (LSTC, Livermore, CA) interface and the input deck until it was as close to the SVM seat bottom and back as possible without any initial penetration between the surfaces. Prior to settling THUMS, the airbag and seatbelt models were deactivated. THUMS was settled into the seat using a 400 ms accelerative loading simulation in the vertical and longitudinal directions. Subsequently, boundary prescribed motions were applied to position the limbs of THUMS within the SVM.

For each CIREN reconstruction, five variables related to the occupant’s position within the vehicle were varied: 1) seat track position, 2) seat back angle, 3) D-ring height, 4) steering column angle, and 5) telescoping position of the steering column. Based on descriptions and photographs from the CIREN database, a baseline set of positioning variables was estimated for each CIREN case. The baseline estimation for the positioning
variables were bounded by 120 sets of positioning variables assigned using a LHD (Table 3). These positioning variables were used to automatically generate simulations to re-position THUMS and vehicle components from the “settled” state to a “pre-collision” state representing a specific occupant posture. With THUMS seated in the tuned SVM, the seat track position and seat back angle were simultaneously modified during a 300 ms re-positioning simulation so that the occupant’s joint angles would change to fit the resulting SVM geometry. Before re-positioning THUMS, the bones in the hands and feet were constrained to the steering wheel and floor, respectively. To set the longitudinal seat track position, a linear boundary prescribed displacement was applied to the entirety of the SVM except the seat (Figure 10-IIA). The seat track position range for each case’s LHD was first defined by the seat track length for the driver’s seat in each case vehicle and then was narrowed to exclude positions where the legs of the scaled THUMS did not fit within the tuned SVM or the constraints between the feet of THUMS and the floor caused the occupant to move forward in the seat. The mid-track seat position was referred to as the zero position, while positive seat track positions indicated that the seat was moved away from the dashboard.

Table 3. Occupant positioning parameter ranges for variation study to simulate various pre-crash occupant positions and postures in the Camry and Cobalt cases.

<table>
<thead>
<tr>
<th>Positioning Variable</th>
<th>Units</th>
<th>2010 Toyota Camry</th>
<th>2006 Chevrolet Cobalt</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Min</td>
<td>Baseline</td>
</tr>
<tr>
<td>Steering Column Angle</td>
<td>Deg</td>
<td>23.2</td>
<td>25</td>
</tr>
<tr>
<td>Steering Column Telescoping Position</td>
<td>Mm</td>
<td>-21</td>
<td>0</td>
</tr>
<tr>
<td>Seat Back Angle</td>
<td>Deg</td>
<td>-1.5</td>
<td>+10</td>
</tr>
<tr>
<td>Seat Track Position</td>
<td>Mm</td>
<td>-142</td>
<td>-71</td>
</tr>
<tr>
<td>D-Ring Height</td>
<td>Mm</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>
During the re-positioning simulation, the seat back and THUMS’s back were concurrently rotated about an axis, maintaining contact between the occupant and seat back cushion (Figure 10-IIIB). Seat back angle ranges were estimated from reports of preferred seating postures (Reed, et al., 1999) and measured between the back surface of the seat and the vertical axis. The D-ring adjustment and steering column were programmatically adjusted between the re-positioning and crash reconstruction simulations using ranges defined for each vehicle in the NCAP report. The zero position for D-ring height was the lowest anchor point. Steering column angle was measured between the axis of the steering column and horizontal axis, while the steering column position was positive when moved closer to the occupant from the mid-position.

2.5 Phase III – CIREN MVC simulations and injury risk estimations

For each re-positioning simulation, a subsequent CIREN crash reconstruction simulation was automatically generated. The case occupant was re-belted and the longitudinal delta-V crash pulse from the CIREN case vehicle’s EDR was applied as the boundary conditions (Figure 10-III). The Camry case had a maximum longitudinal delta-V of 64 km/h over a period of approximately 90 ms, while the Cobalt case had a maximum longitudinal delta-V of 43 km/h over a 110 ms period. Simulated accelerometer, load cell, and chest band instrumentation was adapted from THUMS v4.0 for THUMS v4.01 (Golman, et al., 2015) to measure accelerations at the head center of gravity (CG), T1, T6, T9, T12, and pelvis, forces in the femur and pelvis, and chest deformations over time. Head injury risk was evaluated with Head Injury Criterion (HIC15), while chest injury risk was evaluated with Combined Thoracic Index (CTI), an injury metric that evaluates the effects of both chest compression and acceleration.
Injury metrics and risks were post processed using custom MATLAB (Mathworks, Natick, MA) scripts and batched LS-PrePost (LSTC, Livermore, CA) command files. Injury metric risks for head, chest and femur injuries were used to estimate the likelihood of regional-level injury risks. Logistic injury risk functions were used to calculate risk from HIC15 (NHTSA, 1995), CTI (Eppinger, et al., 1999), and maximum femur force (Kuppa, et al., 2001) using equations in Appendix B.

3. RESULTS

3.1 Simplified Vehicle Model (SVM) Tuning

Vehicle restraint parameters were selected for the two cases described above using the combined Sprague and Geers error factor, C\text{body}. The parameter values selected for the tuned SVM used in each case reconstruction are summarized in Table 2. Injury metrics and seatbelt force comparisons between the simulated and physical crash test data are shown in Appendix A and the Sprague and Geers error factors are summarized in Table A1.

3.2 Injury Risk Estimations

Injury metrics and risks were calculated for each simulation of the two case reconstruction variation studies, including one simulation (the baseline simulation) that best represented the positioning information reported in the CIREN case (Table 4). HIC15 and CTI for both case reconstructions are plotted against each independent occupant positioning variable in Figure 11 for the two case reconstructions. The risks of AIS level 1+, 2+, and 3+ head injury associated with the occupant positioning range for all simulations are plotted in Figure 12.
Table 4. Summary of calculated injury metrics and risks for all real world MVC reconstruction simulations with different occupant positions, including baseline simulations positioned according to the CIREN case descriptions.

<table>
<thead>
<tr>
<th>Case</th>
<th>Metric/Risk</th>
<th>Min</th>
<th>Max</th>
<th>Average</th>
<th>Std. Dev.</th>
<th>Baseline Simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Camry</td>
<td>HIC</td>
<td>220</td>
<td>1090</td>
<td>491</td>
<td>224</td>
<td>311</td>
</tr>
<tr>
<td></td>
<td>AIS 1+ (%)</td>
<td>27</td>
<td>99.5</td>
<td>68</td>
<td>23</td>
<td>46</td>
</tr>
<tr>
<td></td>
<td>AIS 2+ (%)</td>
<td>8.8</td>
<td>93</td>
<td>37</td>
<td>25</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td>AIS 3+ (%)</td>
<td>3.0</td>
<td>62</td>
<td>15</td>
<td>15</td>
<td>5.3</td>
</tr>
<tr>
<td></td>
<td>AIS 4+ (%)</td>
<td>0.6</td>
<td>22</td>
<td>4.0</td>
<td>4.9</td>
<td>1.2</td>
</tr>
<tr>
<td></td>
<td>CTI</td>
<td>0.694</td>
<td>1.113</td>
<td>0.885</td>
<td>0.115</td>
<td>0.832</td>
</tr>
<tr>
<td></td>
<td>AIS 2+ (%)</td>
<td>34</td>
<td>88</td>
<td>61</td>
<td>15</td>
<td>54</td>
</tr>
<tr>
<td></td>
<td>AIS 3+ (%)</td>
<td>3.6</td>
<td>46</td>
<td>15</td>
<td>11</td>
<td>9.1</td>
</tr>
<tr>
<td></td>
<td>AIS 4+ (%)</td>
<td>0.7</td>
<td>13</td>
<td>3.6</td>
<td>3.1</td>
<td>1.9</td>
</tr>
<tr>
<td></td>
<td>Max Femur Force (N)</td>
<td>767</td>
<td>5297</td>
<td>3170</td>
<td>1617</td>
<td>4358</td>
</tr>
<tr>
<td></td>
<td>AIS 2+ (%)</td>
<td>0.45</td>
<td>4.6</td>
<td>2.06</td>
<td>1.37</td>
<td>2.84</td>
</tr>
<tr>
<td></td>
<td>AIS 3+ (%)</td>
<td>0.88</td>
<td>3.7</td>
<td>2.13</td>
<td>0.98</td>
<td>2.81</td>
</tr>
<tr>
<td>Cobalt</td>
<td>HIC</td>
<td>49</td>
<td>467</td>
<td>179</td>
<td>78</td>
<td>161</td>
</tr>
<tr>
<td></td>
<td>AIS 1+ (%)</td>
<td>0.48</td>
<td>74</td>
<td>19</td>
<td>15</td>
<td>15</td>
</tr>
<tr>
<td></td>
<td>AIS 2+ (%)</td>
<td>0.17</td>
<td>34</td>
<td>6.7</td>
<td>6.0</td>
<td>5.0</td>
</tr>
<tr>
<td></td>
<td>AIS 3+ (%)</td>
<td>0.06</td>
<td>11</td>
<td>2.3</td>
<td>1.9</td>
<td>1.7</td>
</tr>
<tr>
<td></td>
<td>AIS 4+ (%)</td>
<td>0.01</td>
<td>2.29</td>
<td>0.49</td>
<td>0.42</td>
<td>0.38</td>
</tr>
<tr>
<td></td>
<td>CTI</td>
<td>0.398</td>
<td>0.963</td>
<td>0.678</td>
<td>0.121</td>
<td>0.690</td>
</tr>
<tr>
<td></td>
<td>AIS 2+ (%)</td>
<td>8.0</td>
<td>72</td>
<td>34</td>
<td>14</td>
<td>33.5</td>
</tr>
<tr>
<td></td>
<td>AIS 3+ (%)</td>
<td>0.45</td>
<td>20</td>
<td>4.3</td>
<td>3.3</td>
<td>3.6</td>
</tr>
<tr>
<td></td>
<td>AIS 4+ (%)</td>
<td>0.09</td>
<td>4.7</td>
<td>0.87</td>
<td>0.72</td>
<td>0.7</td>
</tr>
<tr>
<td></td>
<td>Max Femur Force (N)</td>
<td>1312</td>
<td>2737</td>
<td>2389</td>
<td>268</td>
<td>2287</td>
</tr>
<tr>
<td></td>
<td>AIS 2+ (%)</td>
<td>0.60</td>
<td>1.25</td>
<td>1.05</td>
<td>0.13</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td>AIS 3+ (%)</td>
<td>1.04</td>
<td>1.65</td>
<td>1.48</td>
<td>0.12</td>
<td>1.43</td>
</tr>
</tbody>
</table>
Figure 11. HIC15 and CTI as a function of five positioning variables in the Camry and Cobalt cases. HIC15 and CTI values in the baseline simulation that most closely matched the occupant position documented in CIREN are indicated by the red star.

Figure 12. Head injury risk (AIS 1+, 2+, 3+) as a function of HIC15 (NHTSA, 1995) for all simulated positions in the Camry and Cobalt cases. Head injury risk in the baseline simulation that most closely matched the occupant position documented in CIREN is highlighted with darker shading.
The Camry case occupant’s most severe injuries included an AIS 3 bimalleolar fracture and AIS 2 hemomediastinum. While none of the injury metrics evaluated in this study correlate to bimalleolar fracture, the hemomediastinum can be evaluated with the CTI injury risk function. Across all the potential occupant positions simulated, the average risk of AIS 2+ chest injury was estimated to be 61% (range: 34% to 88%), while the estimated risk according to the best estimate of the occupant’s position in CIREN (baseline position) was 54%. The reconstruction simulations and injury risk functions provide a good indicator of the likelihood of chest injury for this case. The range of occupant position simulations predicted a 27-99% risk of any AIS 1+ head injury, while the baseline CIREN occupant position simulation predicted a 46% risk of AIS 1+ head injury. Consistent with this prediction, the case occupant did not sustain any head injuries. Additionally, the case occupant had no knee, thigh, hip or femur injuries and all simulations predicted less than 5% risk of these injuries.

The three most severe injuries of the Cobalt case occupant were pulmonary contusion (AIS 3), L1 and L3 burst fractures (AIS 2) and loss of consciousness less than one hour (AIS 2). Unfortunately none of the injury metrics evaluated in the study directly correlated to pulmonary contusion, lumbar vertebra burst fracture, or unconsciousness. The injury metric models predicted low levels of injury risk in the head, chest and femur regions. Using the CIREN pre-crash occupant position documentation, there was a 15% probability of AIS 1+ head injury, 33% chance of AIS 2+ chest injury and less than 2% chance of AIS 2+ knee, thigh, or hip injury. A higher calculated probability of head and chest injury was expected, based on the pulmonary contusion and unconsciousness injuries, which may warrant further investigation.
4. DISCUSSION

The most ideal set of parameters for the tuned Camry model yielded close matches between the head acceleration, femur forces and seatbelt forces between the test and simulation data (Figure A1). The simulated chest and pelvis accelerations were overestimated by approximately 15 G’s but matched the test data closely for the first 60 ms of the crash event. The overestimated chest acceleration in the Camry crash test reconstruction may have yielded increased chest accelerations during the subsequent real world crash reconstruction. During the tuning of the Cobalt model, the simulated chest kinematics matched the test data very closely, particularly the shoulder belt force and chest acceleration (Figure A2). The simulated head acceleration was overestimated while the simulated pelvis acceleration and lap belt forces were underestimated. The simulated femurs did not experience both tension and compression modes during the crash event like the H3 in the physical test. Although some regions of the simulated ATD matched the test data better than others, the overall response of the H3 was a close match in both cases.

A few distinct relationships are present between the occupant positioning variables and the head and chest injury metrics. In both cases, the simulations with the most rearward seats had the greatest head acceleration and HIC. This is a function of the vehicle decelerating over an extended period of time before the occupant strikes the airbag. A similar, but weaker, relationship existed between CTI and seat track position. In both cases, the most reclined occupants had increased CTI and in the lower velocity Cobalt reconstruction case, the most reclined occupants also had increased HIC. However, in the higher velocity Camry case, it should be noted that some of the cases
with the highest HIC values were the most upright occupants. This inverted relationship could be attributed to the airbag deploying in close proximity to the occupant. The other three occupant positioning variables in this study did not have notable effects on HIC or CTI compared to the effect of the seat back angle and seat track position.

Calculated injury risk ranges for the three anatomical regions closely correlated to the injury patterns observed in the CIREN occupants. THUMS injury metrics correctly predicted the AIS 2 hemomediastinum (54% baseline risk) and lack of head and knee, thigh, and hip injuries (46% and 3% baseline risk, respectively) in the Camry CIREN occupant. For the Cobalt CIREN occupant, THUMS injury metrics correctly predicted the lack of knee, thigh, and hip injuries (1% baseline risk), but underestimated the AIS 2 pulmonary contusion (33% baseline risk) and AIS 2 head injury (5% baseline risk). The injury risk curves may underestimate head and chest injury risks in elderly occupants such as the 80 year old Cobalt driver. For instance, CTI evaluates chest acceleration and chest compression, while chest compression is increased in elderly crash occupants, yielding an overall increased risk of injury according to CTI (Ruan, et al., 2003). Future implementation of organ-level injury metrics, such as strain-based metrics for lung or brain injury, may yield more accurate injury predictions for the pulmonary contusion and head injury in the Cobalt occupant (Danelson, et al., 2015; Golman, et al., 2014).

4.1 Limitations

Due to the limited availability of occupant specific FE HBM's and detailed vehicle models, several assumptions and simplifications used in this study may have influenced the results. Many of the most significant assumptions involved the tuning of the SVM. Frontal NCAP crash tests are only performed at one speed and are not performed on
every vehicle model each year. Because each vehicle was not tested each year, sisters and clones were used to match the crash test vehicle calibration to the CIREN case vehicle. For a given model year, the occupant restraint systems could vary from a sister or clone vehicle despite having matching vehicle stiffness characteristics. Additionally, if the simulated H3 occupant response closely matched the frontal NCAP crash test, it was assumed that the THUMS occupant response within the same vehicle model would match a human’s response in crash events occurring with similar velocities. It was assumed that the properties varied within the SVM tuning LHD were not dependent upon the speed of the crash. Despite the inability to tune the SVM at varying crash speeds, performing vehicle-specific FE reconstructions is an improvement on past studies (Danelson, et al., 2015; Golman, et al., 2014).

Generating boundary conditions for the frontal crash reconstruction simulations from EDR reports resulted in a few limitations as well. While the vehicle tuning simulations incorporated pitching of the vehicle, EDR reports did not present pitching of the vehicle throughout the crash event and therefore the crash pulses applied to reconstruct the CIREN crashes only incorporated the linear acceleration pulse. Additionally, low magnitude lateral acceleration pulses can occur during frontal crashes that may not be recorded by the EDR in frontal crash events and therefore not implemented into the crash simulation.

Using the THUMS 50th percentile male model for all case reconstructions was another limitation in this study. The chosen scaling method was selected to maintain the same regional and organ-level mesh geometries. There was no variation in the occupant girth or weight to create occupant-specific models. To account for significant variation of
occupant girth from the scaled THUMS model, morphing techniques could be used to modify the shape of the case occupant or scaling could be performed on the 5\textsuperscript{th} percentile female or 95\textsuperscript{th} percentile male THUMS. Similarly, anatomical material properties remained the same for each case reconstruction. Material properties of individual organs and bones could be modified to account for age and sex differences.

4.2 Future Work and Applications

The breadth of this study involved detailed reconstruction of two CIREN cases and evaluation of select regional-level injury metrics. The reconstruction methodology can be directly applied to reconstruct a larger number of frontal MVCs of varying vehicle types from CIREN and the National Automotive Sampling System – Crashworthiness Data System (NASS-CDS). The methodology could potentially be extended to reconstruct a variety of crash modes through the addition of vehicle structures such as the door and A-pillar that engage the occupant in offset frontal and near-side impacts. Additional injury metrics can be implemented in THUMS to evaluate risk for specific injuries rather than the regional injury risks described by this study. Future studies may evaluate fracture risks of complex anatomical structures such as the vertebral bodies, pelvis, and malleoli. By evaluating additional cases in further detail, new injury metrics and risk functions could be developed from the real world crash data to assess the effectiveness of restraint systems to prevent and mitigate injuries that are not easily studied using post-mortem human subjects (PMHS) or ATDS. Additionally, the ability to place bounds on injury risk as a function of pre-crash occupant position is valuable for assigning confidence levels to injury mechanism predictions in real world MVC analysis.
5. CONCLUSION

A three phase process was developed to reconstruct CIREN MVCs of varying vehicle types using a tuned simplified vehicle FEM. CIREN MVCs were reconstructed with various THUMS occupant positions by varying the seat track position, seat back angle, steering column angle and telescoping position, and the seatbelt D-Ring anchor height. The reconstruction process allows for quantification of the sensitivity and uncertainty of the injury risk predictions based on occupant position, which is often uncertain in real world MVCs. This study provides perspective on the sensitivity of pre-crash occupant positioning within the vehicle compartment. By studying a variety of potential occupant positions, we can understand important factors that lead to more severe injuries and potentially mitigate these injuries with advanced safety systems to protect occupants in more dangerous positions.

6. ACKNOWLEDGMENTS

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REFERENCES

Appendix A – Simplified Vehicle Model (SVM) Tuning Injury Metric Comparison

Table A1: Sprague and Geers error factors for the tuned SVM for the Camry and Cobalt cases. Comprehensive, magnitude, and phase errors are reported comparing the crash test signals to the simulation that resulted in the most similar response. The vehicle parameters from the simulation that produced the most similar response (comprehensive error nearer to zero) were used to tune the SVM.

<table>
<thead>
<tr>
<th>Injury Metric</th>
<th>2010 Toyota Camry</th>
<th>2006 Chevrolet Cobalt</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Comprehensive (C)</td>
<td>Magnitude (M)</td>
</tr>
<tr>
<td>Head Acceleration</td>
<td>0.17</td>
<td>-0.15</td>
</tr>
<tr>
<td>Chest Acceleration</td>
<td>0.14</td>
<td>-0.08</td>
</tr>
<tr>
<td>Pelvis Acceleration</td>
<td>0.15</td>
<td>-0.06</td>
</tr>
<tr>
<td>Left Femur Force</td>
<td>0.12</td>
<td>-0.04</td>
</tr>
<tr>
<td>Right Femur Force</td>
<td>0.19</td>
<td>0.15</td>
</tr>
<tr>
<td>Lap Belt Force</td>
<td>0.04</td>
<td>-0.02</td>
</tr>
<tr>
<td>Shoulder Belt Force</td>
<td>0.13</td>
<td>0.10</td>
</tr>
<tr>
<td>Total Body</td>
<td><strong>0.32</strong></td>
<td><strong>0.38</strong></td>
</tr>
</tbody>
</table>
Figure A1: Injury metric comparisons between the simulated and physical 2010 Camry crash tests.
Figure A2: Injury metric comparisons between the simulated and physical 2005 Cobalt crash tests.
## Appendix B – Injury Risk Functions

Table B1: Injury risk functions used for this study.

<table>
<thead>
<tr>
<th>Head Injury Risk</th>
<th>(NHTSA, 1995)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AIS 1:</td>
<td>( \frac{1}{1 + e^{(1.54 + 200/HIC_{15} - 0.0065 \cdot HIC_{15})}} )</td>
</tr>
<tr>
<td>AIS 2:</td>
<td>( \frac{1}{1 + e^{(2.49 + 200/HIC_{15} - 0.00483 \cdot HIC_{15})}} )</td>
</tr>
<tr>
<td>AIS 3:</td>
<td>( \frac{1}{1 + e^{(3.39 + 200/HIC_{15} - 0.00372 \cdot HIC_{15})}} )</td>
</tr>
<tr>
<td>AIS 4:</td>
<td>( \frac{1}{1 + e^{(4.9 + 200/HIC_{15} - 0.00351 \cdot HIC_{15})}} )</td>
</tr>
<tr>
<td>AIS 5:</td>
<td>( \frac{1}{1 + e^{(7.82 + 200/HIC_{15} - 0.00429 \cdot HIC_{15})}} )</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Chest Injury Risk</th>
<th>(Eppinger, et al., 1999)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AIS 2:</td>
<td>( \frac{1}{1 + e^{(4.847 - 6.036 \cdot CTI)}} )</td>
</tr>
<tr>
<td>AIS 3:</td>
<td>( \frac{1}{1 + e^{(8.224 - 7.125 \cdot CTI)}} )</td>
</tr>
<tr>
<td>AIS 4:</td>
<td>( \frac{1}{1 + e^{(9.872 - 7.125 \cdot CTI)}} )</td>
</tr>
<tr>
<td>AIS 5:</td>
<td>( \frac{1}{1 + e^{(14.242 - 6.589 \cdot CTI)}} )</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Knee, Thigh, Hip Injury Risk</th>
<th>(Kuppa, et al., 2001)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AIS 2:</td>
<td>( \frac{1}{1 + e^{(5.7949 - 0.5196 \cdot \text{Max Femur Force})}} )</td>
</tr>
<tr>
<td>AIS 3:</td>
<td>( \frac{1}{1 + e^{(4.9795 - 0.3260 \cdot \text{Max Femur Force})}} )</td>
</tr>
</tbody>
</table>
Chapter IV: Summary of Research

The work presented in this thesis has built an important framework for finite element reconstruction and analysis of real world motor vehicle crashes. The research presented here, indicated that an automated process can be used to perform the following tasks necessary to reconstruct real world MVCs from the CIREN database:

1) Tuning of a generic simplified vehicle model to mimic the frontal crash environment of a specific vehicle model.

2) Position a FE HBM within a vehicle model according to the seat track position, seatback angle, seatbelt D-ring height, and steering column position and angle.

3) Assess the sensitivity of pre-crash occupant position on regional level injury metrics and risks.

This work will make meaningful contributions to many future studies. It will ultimately allow researchers to address specific injury biomechanics research topics by reconstructing injury events of living human subjects. By reconstructing real world injuries, injury risk functions may be developed for injuries that cannot be evaluated with ATD, PMHS or animal experiments. Research presented in Chapter II and Chapter III is expected to be published in scientific journals listed in Table 5.

<table>
<thead>
<tr>
<th>Chapter</th>
<th>Topic</th>
<th>Journal</th>
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<tr>
<td>II</td>
<td>Optimization of a Simplified Automobile Finite Element Model Using Time Varying Injury Metrics</td>
<td>Biomedical Sciences Instrumentation †</td>
</tr>
</tbody>
</table>

†Published  
*Submitted
Scholastic Vita

James P. Gaewsky
Graduate Student and Research Engineer
Virginia Tech-Wake Forest University
Center for Injury Biomechanics
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EDUCATION
Virginia Tech – Wake Forest University, Winston-Salem, NC
Master of Science, Biomedical Engineering 2013 – Present
GPA: 3.60
Research Lab: Virginia Tech – Wake Forest University Center for Injury Biomechanics
Advisor: Dr. Joel Stitzel

University of Rochester, Rochester, NY
Bachelor of Science, Biomedical Engineering 2009 – 2013
Overall GPA: 3.48/4.0; Engineering GPA: 3.75/4.0

PROFESSIONAL EXPERIENCE
Graduate Research Engineer 2013 – Present
Virginia Tech- Wake Forest University Center for Injury Biomechanics
Wake Forest School of Medicine
Winston-Salem, NC
- Toyota’s Collaborative Safety Research Center
  - Crash Simulation using the Total Human Model for Safety

Undergraduate Research Intern Summer 2012
Virginia Tech- Wake Forest University Center for Injury Biomechanics
Winston-Salem, NC

HONORS AND AWARDS
Ohio State University Injury Biomechanics Symposium Travel Award May 2015
Rocky Mountain Bioengineering Symposium 2nd Place Paper April 2014

LEADERSHIP
Solar Splash -- Solar-Electric Boat Club Fall 2010 – Spring 2013
- President (Fall 2012-Present), Vice President (Spring 2011-Spring 2012)
- Solar Splash is an international collegiate competition in which schools design, build, analyze and drive a solar-powered boat. As a club we competed in Iowa in June 2011 and June 2012.
- Led design and fabrication teams for 17-foot hull used at 2011 and 2012 competitions.

COMPUTING SKILLS
- LS-DYNA - Matlab - Solidworks - R - Hyperworks
- LS-PrePost - Microsoft Office - Abaqus - JMP - Nastran

63
PROFESSIONAL AND STUDENT MEMBERSHIPS

<table>
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<tr>
<th>Organization</th>
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<tr>
<td>Biomedical Engineering Society, VT-WFU Chapter</td>
<td>2013 - Present</td>
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<tr>
<td>Biomedical Engineering Society, University of Rochester Chapter</td>
<td>2011 - 2013</td>
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PROFESSIONAL CERTIFICATION AND WORKSHOPS

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<th>Workshop &amp; Training</th>
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<tr>
<td>LS-DYNA Material Models Training</td>
<td>Winston-Salem, NC</td>
<td>February 2014</td>
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<td>LS-DYNA Contacts Training</td>
<td>Winston-Salem, NC</td>
<td>January 2014</td>
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<td>HyperCrash Training</td>
<td>Winston-Salem, NC</td>
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<td>Winston-Salem, NC</td>
<td>October 2013</td>
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<tr>
<td>LS-PrePost and LS-DYNA Training</td>
<td>Troy, MI</td>
<td>August 2013</td>
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BIBLIOGRAPHY

Papers in Refereed Publications

Papers in Refereed Conference Publications

Other Conference Papers (Abstract Style or Non-Refereed)

Technical Reports


5. Stitzel JD, Weaver AA, Danelson KA, Gaewsky JG. “Crash Simulation using the Total Human Model for Safety.” Year 3, Quarter 2, Report to Toyota’s Collaborative Safety Research Center, Ann Arbor, MI, March, 2014.


7. Stitzel JD, Danelson KA, Weaver CM, Gaewsky JG. “Crash Simulation using the Total Human Model for Safety.” Year 2, Quarter 4, Report to Toyota’s Collaborative Safety Research Center, Ann Arbor, MI, March, 2014.