Biomechanical Response of the PMHS Thorax to High Speed Lateral and Oblique Impacts

Thesis

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By

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Abstract

Previous research in the kinematic biomechanical response of the thorax to lateral impact has been limited. Eppinger, in 1984, and Viano, in 1989, conducted lateral and oblique (30° anterior of lateral) pendulum impacts of the thorax that resulted in the development of lateral biofidelity corridors. These corridors are used to design anthropomorphic test devices (ATD) that respond in a human-like way to lateral impacts. Viano determined that oblique impact responses can be grouped with lateral impact responses because the peak forces and deflections were found to be similar. Shaw et al., in 2006, performed lateral and oblique impacts at non-injurious speeds and found the post mortem human surrogate (PMHS) thoracic response to be directionally dependent. A query of the National Accident Sampling System – Crashworthiness Data System (NASS/CDS) shows that the frequency of oblique loading of vehicle occupants occurs at a frequency three times that of lateral loading. Understanding when there is a directional dependence in these types of impacts is critical for ATD development.

In this study, nine PMHS were impacted laterally or obliquely at speeds meant to cause a 50% probability of AIS3+, or serious, injury to occur. Loads imparted to the subject were recorded using a ram mounted accelerometer and thoracic deflections were
measured using a chestband. Strain gauges mounted to the subject’s ribs allowed the timing of rib fracture to be estimated. In all of the 4.5 m/sec tests rib fractures were the major injuries that were documented. Additionally, the major vasculature, including the superior vena cava (SVC), inferior vena cava (IVC), aorta, and pulmonary arteries and veins, were filled with saline to mimic the vascular pressures of a living human. Internal pressure sensors were placed in the aortic arch to detect pressure changes during the impact in the event of an aortic rupture. None of these impacts resulted in such an injury.

The average peak normalized load for the lateral tests was 2557 N (σ=59.7 N) and for the oblique tests was 2632 N (σ=346.3 N). The average peak normalized deflection for the lateral tests was 70.0 mm (σ=0.15 mm) and for the oblique tests was 56.1 mm (σ=2.85 mm). The peak pressures recorded from each test are included in the results section. Estimated rib fracture timings are also included in the results section.

It appears from these results that the directionally dependent kinematic response of the thorax is not present when the subject is impacted at injurious speeds. This may be due to the viscoelastic response of the internal thoracic organs taking over at these higher speeds. The structural integrity of the boney thorax was compromised in many of the tests; flail chest was a common injury. When the ribs fracture, the internal organs carry more of the load from the ram plate. More of these types of pendulum impacts should be performed to determine when the directional dependence of the thoracic response changes.
Dedication

To my beautiful wife, Sabrina, and our wonderful daughter, Grace
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There are many people to thank for the support they provided in this research. None of the work that I have done would have been possible without each of them.

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Chapter 1: Introduction

Much of the history of automotive safety research has focused on researching frontal impact protection. This has led to the development of effective safety implements such as frontal airbags for driver and passenger and seatbelts with pretensioners. It is estimated, for instance, that the cumulative number of lives saved by seatbelt use from 1975 to 2008 is 255,115 [1]. The National Highway Traffic and Safety Administration (NHTSA) recently improved standards that require better protection for occupants in side impact accidents. One such standard is the Federal Motor Vehicle Safety Standard 214 (FMVSS 214) [2].

The goal of FMVSS 214 is to simulate the most common side impact crashes, those occurring at intersections. A moving deformable barrier (MDB) moving at 54 kmh impacts a stationary vehicle. The MDB travels along a line rotated 63° from the long axis of the stationary vehicle and impacts with its front end parallel to the side of the vehicle. Figure 1.1 shows the FMVSS 214 test setup for the vehicle and MDB.
Figure 1.1: FMVSS 214 test setup [2]

The resulting impact from this test is a primary direction of force (PDOF) that is not purely lateral to the occupant. Instead the PDOF is approximately 60°, which Samaha [3] described as being consistent with real world side impact accidents for cars model year 1995 and later. According to the National Accident Sampling System –
Crashworthiness Data System (NASS/CDS) in 2004 and 2005, accidents involving a PDOF of 60° outnumbered accidents involving a PDOF of 90° about three to one, depicted in Figure 1.2.

When vehicles are tested for FMVSS 214, occupants are seated on the near side of the impact. These occupants are anthropomorphic test devices (ATDs), or crash test dummies, that are meant to replicate the response of humans in side impacts. Through their use, vehicle safety research experts can determine injuries the occupant could have expected to incur as a result of the accident. This information is inferred from
measurements that sensors on the ATD record throughout the test. It is critically
important that ATDs accurately represent human response. In order to validate the ability
of an ATD to do replicate human responses, crash test dummy manufacturers must have
access to data that details human responses to impact loading. This human response data
is typically gathered from testing either post mortem human surrogates (PMHS), or
through the use of surrogate animal models. For the testing that will be discussed in this
study, PMHS were used to characterize the response of the thorax to impact at high
speeds.

One of the measurements that ATDs are able to record is rib deflection. This
typically is found to be one of the best indicators of thoracic trauma. However, many of
the side-impact dummies that have been developed are only able to measure thoracic
deflections in a purely lateral direction. When considering the PDOF frequency that is
shown in Figure 1.2, it is likely that accurate rib deflections are not detected by these
dummies. As a result, the ability of an ATD to accurately predict injury may be
compromised. ATDs that are able to measure rib deflections along a wider range of side
impact angles are being developed. One of these dummies, WorldSID [4], is able to
measure rib deflections along a range of impact directions from 30° posterior of lateral to
30° anterior of lateral. Not only is it important for the ATD to be able to measure
deflections that occur at angles other than 90°, but also the ATD needs to accurately
represent the response of a human when impacted from angles other than 90°. In order to
do that, this study has been developed to load the PMHS thorax from lateral (90°) and
oblique (60°; 30° anterior of lateral) directions.
Curved beam bending theory can be applied to the simplified human thorax to estimate how deflection may change for similar loads with different loading directions. Figure 1.3 shows a curved beam under load. According to the moment-area theory [5], the deflection that occurs at point B due to the load, F, can be calculated from the following equation

$$
\Delta_B = \frac{FR^3}{2EI} \sqrt{1 + \frac{\pi^2}{4}}
$$

(1.1)

Where: 
F is the load carried by the beam 
R is the radius of curvature at the location of interest 
E is the Young’s Modulus for the beam material 
I is the area moment of inertia for the beam cross section

Figure 1.3: Curved beam under load, F
For the complete derivation 1.1, please see Appendix A. If it is assumed that the load \( F \), elastic modulus \( E \), and the moment of inertia \( I \) are the same for some loading at any location along the rib, then the radius of curvature of the rib would be the only variable affecting the observed deflection. Thusly, the response of the thorax to impact would be very sensitive to the direction of loading. In Figure 1.4, the human thorax is modeled as an ellipse. The radius of curvature of the oblique thorax, represented by \( R \), is clearly large compared to the radius of curvature of the lateral thorax, represented by \( r \).

![Figure 1.4: Radius of curvature for lateral and oblique aspects of the human thorax](image)

Previously, Shaw et al. [6] showed that the response of the PMHS thorax to loading was sensitive to loading direction at low speeds (2.5 m/s). Specifically it was
found that when the thorax is loaded laterally, the thoracic response is more stiff than when the thorax is loaded obliquely. It is the purpose of this study to determine if this sensitivity exists when the thorax is loaded at higher speeds meant to cause injury (4.5 m/s and 5.5 m/s).
Chapter 2: Background

Comprised of two sections, this chapter will first provide an overview of the anatomy of the human thorax that is pertinent to this study. In the second section there is a discussion of past research that has been done involving the response of the thorax to blunt impact and its implications for safety research.

2.1 Thorax Anatomy

The thorax is the upper region of the human torso that is bounded superiorly by the inferior end of the neck and inferiorly by the diaphragm [7]. Twelve pairs of ribs, twelve thoracic vertebrae, the sternum, and the costal cartilages make up the solid structure of the thorax, commonly referred to as the rib cage. The rib cage performs several important functions including: muscle attachment for movement and protection of vital organs. Muscle attachments are for the muscles of the arm, abdomen, neck, and back. Organs that make up the contents of the thorax include the heart and lungs.

The boney attachment points for the thorax to the rest of the body are located along the spine. The 1st thoracic vertebra (T1) connects the thorax with the cervical spine and head, while T12 connects the thorax with the abdominal region and pelvis. Figure 2.1
depicts the bony aspects of the thorax, including the bones of the shoulder girdle that attach the upper extremity to the thorax.

Figure 2.1: Anterior view of bony thorax [7]
The ribs attach anteriorly to costal cartilage, which is then connected to the sternum. The ribs articulate posteriorly with the vertebra at articular facets, depicted in Figure 2.2. The ribs contact the vertebra of the same number designation (i.e. rib 7 with T7) on the superior facet and also with the inferior facet of the next vertebra up.

![Figure 2.2: Rib articulation with vertebrae](image)

At the ribs’ anterior articulation, rib one articulates with the manubrium, the most superior aspect of the sternum. Rib two articulates with the manubrium and body of the sternum. Ribs three through six articulate with body of the sternum while rib seven articulates with the body and xyphoid. These first seven ribs comprise the “true ribs”
group of the rib cage because they are the only ribs to have unique attachments to the spine and sternum. Ribs eight through ten do not articulate with the sternum, instead they attach to costal cartilage that merges with the cartilage of rib seven. Because of this, these ribs are considered “false ribs”. Ribs eleven and twelve do not have an anterior articulation, and are considered “floating ribs”. As stated previously, all of these anterior rib articulations are with costal cartilage which then articulates with the boney sternum. Figure 2.3 shows the articulation points for the ribs with the sternum.
Figure 2.4 shows the rib cage and the internal thoracic organs it protects. The lungs attach superiorly and medially to the bronchi which in turn attach to the trachea. Inferiorly the lungs are supported by the diaphragm. In order to facilitate inspiration and avoid high friction between sliding surfaces, the lungs are covered superficially by a thin visceral pleura that forms a continuous sheet with the parietal pleura that lines the deep surface of the rib cage. A net negative pressure is maintained in between the pleura,
allowing the lungs to expand and take in air as the thorax expands. Between the lungs is
the heart, which resides in a pericardial sac anterior to the bronchi. From the base of the
heart the ascending aorta climbs to the level of the second rib. At this level the aorta
arches towards the posterior of the thorax and descends along the spine creating the
descending aorta.

Figure 2.4: Anterior view of thorax with internal organs [7]
Figure 2.5 shows internal organs located in the superior aspect of the abdomen and their relation to the rib cage. Just inferior to the diaphragm, on the subject’s right side mostly anterior of lateral, is the liver. Located on the inferior aspect of the liver is the gallbladder that stores bile and introduces it to the digestive tract as called upon to aid with the digestion of fats. The spleen is located inferior to the diaphragm on the left side of the subject, posterior of lateral.

In the impacts that were done for this study, organs and structures on the left side of the subject were most likely to be injured. Internal organs that were most susceptible
were the left lung, heart, stomach, and spleen. When looking at the response of the thorax to impact it is expected that the boney structures, particularly the rib cage, will have the largest contribution. However, it is also possible that at the speeds that are being investigated in this study the internal organs will also play a large part in thoracic response due to their rate dependent viscoelastic nature.
2.2 Literature Review

Blunt trauma of the thorax has long been a focus of injury biomechanics research. Many studies have been conducted on frontal thoracic loading. Much work has also been done in side impact testing through the use of acceleration sleds. However, very little work has been done using pendulum testing to characterize the response of the thorax to impact. Eppinger [8] conducted four pendulum tests involving impacts to PMHS thoraces as part of a larger series of tests that also involved sled testing. Using a 23.4 kg pendulum, three subjects were impacted laterally at 4.5 m/s and the fourth was impacted laterally at 6.3 m/s. The Thoracic Trauma Index (TTI) was developed from this work, which involved measurements from thorax mounted accelerometers.

One of the other major studies conducted in this area within the last twenty years was by Viano [9]. In this series of tests in 1989, Viano impacted fourteen PMHS in three body regions (thorax, abdomen, and hip) with varying speeds. Many of the subjects were used for multiple tests in order to build a larger collection of data given a limited number of subjects. Figure 2.6 depicts the test setup utilized for these pendulum tests.
Viano indicated that for subjects where multiple tests were completed, initial tests were of a low speed, non-injurious level and successive tests were completed at higher energies on a different body region of the subject. For example, a subject may have been impacted in the thorax at a low speed, 3.6 m/sec, and then it would be impacted in the abdomen at speeds as high as 9.8 m/sec. As testing continued the researchers became more willing to impact subjects repeatedly. The first three subjects were impacted once, but by the end of the test series subjects were impacted as many as six different times. It
was not clear from the paper that any imaging was conducted between test runs to determine if injury had occurred.

The bulk of this data was used to construct force-deflection corridors to define the response of the thorax to blunt trauma. Viano found that $V_{\text{max}}C_{\text{max}}$, the product of the maximum compression value of the thorax and the velocity at which the thorax was impacted, provides the best prediction of severe injury to a human. This measurement was determined to be more reliable than the TTI [9]. Viano argued there was danger in solely using TTI to predict injury because padding, which is very helpful in preventing serious injury, could be ignored because padding would not greatly change the measured accelerations of a subject, thus resulting in a large TTI. Cushioning would change the energy being transferred to a subject in an impact, thus reducing the compression of the subject’s torso.

Pendulum testing using PMHS, is not the only method employed in injury biomechanics research. Sled tests, where subjects are seated on a seat and accelerated down a track, are also often used. Cavanaugh et al. [10] conducted twelve side impact sled tests to study the biomechanical response of the thorax.

In these sled tests, the subject was seated on a Teflon coated seat pan next to either a rigid or padded wall mounted to the sled. Two seated positions were used for the subjects. In one position, the subject was placed flush against the impacting wall and in the other the subject was placed 0.15 meters away from the impacting wall. Instrumentation external to the subject included load cells mounted to the impact wall at
four levels (shoulder, thorax, abdomen, and pelvis) to measure how the subject loads the wall. Internally the subject was instrumented with accelerometers mounted at the level of thoracic vertebrae five on the spine, sternum, and on the ribs at the midaxillary line. Finally high speed cameras were used to track fiducial markers on the subject for displacement data.

In the rigid tests the sled was accelerated to achieve a constant speed between 6.7 m/sec and 9 m/sec. All of the padded tests were conducted at 9 m/sec. Repeatability of the lower speed impact was generally good whereas the high speed tests were conducted from 8.74 m/sec to 10.47 m/sec. While the deviation from 9 m/sec is somewhat large, the achieved speeds would not be expected to cause different levels of injury.

Using high speed cameras and photo markers on the subject, half chest deflections (as is typically measured in anthropomorphic test devices) were calculated. These deflections were compared to the level of injury seen in each subject and used to create injury risk curves. As was seen in the Viano tests, $V_{max}C_{max}$ and compression were found to have the strongest correlation with injury severity. However, it would be difficult to compare the compression values that result in injury from Cavanaugh’s tests with those of Viano’s due to the interaction of the arm with the impacting surface. In the pendulum tests the arms were raised so that the impacting pendulum face would contact only the thorax of the subject. The sled tests left the arms at the side of the subject so that the loading wall would impact the arms as well as the thorax. Transmission of the impact
forces through the shoulder girdle would reduce the energy going through the thorax, potentially resulting in a very different thoracic response and mode of injury.

Pintar et al. [11] worked to develop a more reliable method for measuring the deflection occurring in PMHS during side impact testing. In the previously discussed articles the deflection of a subject’s thorax was determined from video analysis of externally applied markers or markers applied to rods that had been screwed into bony structures in the subject. Based on work by Eppinger [12] in 1989, a chestband was developed that could continuously measure the contour of a subject’s torso during impact.

In a series of side impact pendulum tests Pintar used a side impact dummy (SID) to measure the efficacy of the chestband to measure deflections compared to high speed video analysis. Six pendulum tests were conducted laterally to the SID with square and round face plates. The SIDs were outfitted with a single chestband along with retroreflective markers used for video analysis. Results showed that chestband data, after being processed with RBand-PC software, had an error of 1.3% when compared to the video analysis. To further test the ability of the chestband to be used during side impact tests, five side impact sled tests were conducted using PMHS. Figure 2.7 shows how the impacting wall was setup for Pintar’s tests.
In the PMHS tests, three chestbands were placed on each subject at the levels of rib four, the xyphoid process, and the tenth thoracic vertebra. These PMHS sled tests were conducted in a similar fashion to Cavanaugh et al. [13] tests, except that the impact wall was changed so the shoulder of the subject was not impacted directly. Again using high speed video analysis to validate the chestband data, the band was found to have an error of approximately 2% for all of the PMHS tests conducted. The tests conducted also showed that the chestbands were reliable after multiple uses at speeds of 6.4 m/sec to 9.2 m/sec. When compared to the Cavanaugh data for half chest deflections, under similar testing conditions the chestband provided similar results, lending more credibility to the
chestband’s ability to accurately depict the deflection of the thorax under high speed loading.

Kuppa et al. [14] used data from forty-two PMHS side impact sled tests along with thirty-eight paired dummy tests using the EuroSID-II with rib extension modifications. These sled tests involved a very similar sled setup to the Pintar work described previously. External instrumentation used in these tests included load cells in the impacting wall. Internally the subject was instrumented with tri-axial accelerometer packages at T1, T12, and the sacrum, along with uni-axial accelerometers on the lateral aspect of ribs four and eight on the subject’s impact side and on the sternum body. Chestbands were placed at the levels of the fourth and eighth ribs. The tests were conducted with a mix of rigid and padded walls as well as flush or pelvic-wall-offset positions.

Kuppa modified the measured chestband deflections that were used to analyze the deflection recorded in the thorax. Measurements of the subcutaneous tissue thickness were taken for each subject and an average thickness was subtracted from the chestband deflection measurements, in an attempt to characterize the deflection of only the bony structure of the thorax. This resulted in a slightly different compression value for each test. She found that the modified chest deflections were better predictors of AIS 4+ than they were for AIS 3+ injury. Kuppa’s rationalization for this was that many of the AIS 4+ injuries involved flail chest injuries, where multiple fractures result in an unstable chest wall. Because of this change in the structural integrity of the rib cage, the chestband
readings could indicate a very large deflection. Comparison of thoracic deflections between injured and non-injured subjects using the chestband may not be fair.

Using the paired dummy tests, Kuppa compared all measurements observed in the ES2-res and compared them to the injuries documented in the PMHS. She determined that the dummy data did not provide the same predictive capability as was found with the PMHS data. Instead of the maximum normalized half thorax deflection and TTI being the best predictors of injury, as was the case for the PMHS, the dummies predicted injury best based on maximum rib deflection and average spine accelerations. It was also noted that because many automotive accidents do not result in a strictly lateral loading of the occupant, the ES2-re may not be able to measure true maximum rib deflections from these types of accidents.

In 2005, Yoganandan and Pintar [15] performed lateral sled tests with side impact dummies. These sled tests used a similar setup to the Pintar and Kuppa sled tests shown in Figure 2.7. The dummies that were used in this study were the SID (modified Hybrid II dummy), ES2-re, BioSID, and WorldSID prototype. The biofidelity of these dummies could be determined by comparing these tests to those run in previous sled tests with PMHS.

Chestbands were placed on the SID so that the time history of its thoracic contours could be determined and compared directly to the existing PMHS data. Three chestbands were used, placed at the level of the fourth rib, the xyphoid process, and at the tenth thoracic vertebra. The chestbands from the SID along with the linear (ES-2re),
string (BioSID), and infrared potentiometers (WorldSID) were used to calculate the deflection that occurred during impact.

The location of maximum compression was evaluated for all of the dummies and characterized as occurring at the upper chest level, the mid chest level, and the lower chest level. Each dummy exhibited varying chest levels of maximum compression that also varied by the type of impacting wall that was used. The data presented indicates that the biofidelic response of each of these dummies varies and confirms that side impact ATDs have to be further developed and researched to behave more like humans [15].

The authors concede that these tests were limited by being conducted at one speed (6.7 m/sec) and that more testing needs to be done to understand how these dummies compare to PMHS tests that were conducted at higher speeds. It should be noted that Kuppa et al. also commented on the issue of the ES2-re dummy being hung up during impact due to the interface between its rubber back plate and the seatback used in sled tests. This would result in the dummy under predicting injury because the imparted load would be transferred to the seat back, effectively offloading that force from the dummy. The ES2-re that she tested with had a modified back plate with needle bearings and a Teflon coating. The materials that modern ATDs are constructed with have come very far from the materials that were first used to produce the dummies in the 1970’s, but they are still far from being able to accurately replicate the response of a human in similar crash conditions.
In 2006, Shaw et al. [6] tested PMHS under oblique and lateral loading conditions using a pneumatic ram at 2.5 m/sec. In previous work by Viano and Eppinger pendulum impacts were conducted laterally and obliquely to PMHS. These tests were all conducted between 3.6 m/sec and 10.2 m/sec. Viano concluded that because all of the oblique tests resulted in peak forces that were similar to the lateral forces that Eppinger had observed, all of the tests could be grouped together. In 1999 the International Standards Organization TR9790 [16], used this data to construct a biofidelity corridor that determines how well a dummy mimics a thoracic response to blunt lateral impacts.

Shaw’s tests were conducted with the PMHS in a seated position, suspended from a hook temporarily held in place by an electromagnet. The arms of the subject were positioned anteriorly and raised perpendicular to the floor so as to not interfere with the ram face impacting the thorax. Instrumentation included a chestband located at the level of the fourth rib; tri-axial accelerometers on the lateral aspect, 30° anterior of lateral, and 30° posterior of lateral on the fourth rib on each side of the subject; and tri-axial accelerometers on vertebrae T4 and T8 or T12. To measure the force imparted to each subject a load cell was added behind the face of the impacting ram. Figure 2.8 shows a typical setup for Shaw’s lateral tests.
Figure 2.8: Setup for Shaw (2006) lateral impact tests [6]

From these tests it was concluded that the response of the thorax differed when considering oblique versus lateral impacts. From the force deflection curves that were obtained, the response of the thorax in lateral loading was stiffer than that observed during oblique loading. Peak normalized loads were 1430 N for lateral tests and 1130 N for oblique tests. Normalized deflections were 32.5 mm for the lateral tests and 49.5 mm
for the oblique tests. These normalized deflections translated to average peak compressions of 10% in the lateral tests and 16% in the oblique tests. Based on previous testing, it is possible that this increase in compression could lead to more serious injury.

Pintar et al. [17] in 2007 investigated the severity and frequency of thoracic injury occurring from pole side impacts where vehicles had struck tall, cylindrical objects, such as a light pole or a tree. A search of the Crash Injury Research and Engineering Network (CIREN) database found 49 typical pole crashes. Pintar concluded that 26 of these crashes resulted in severe chest trauma and of these 26 cases, 17 cases had lung contusions and 19 cases had rib fractures, most of which were unilateral. Hypothesizing that the oblique loading from these pole impacts would result in unilateral rib fractures, lung contusions, and possible aortic ruptures, he conducted 20° and 30° oblique PMHS sled tests.

Even though the testing done was only on two PMHS subjects, the injuries that were observed in these 6.7 meters per second tests were consistent with those seen in the CIREN study. It was proposed by Pintar that because of the severity of injury that occurs in these oblique impacts, it is important to further develop data sets that characterize the response of the thorax to oblique loads, especially due to the current generation of ATDs being incapable of accurately measuring the deflection of the thorax in these directions. While the sample size in this particular case is small, it does help support the findings of Shaw et al. that oblique impact scenarios should not be lumped with lateral impact data and that further investigation is warranted.
Chapter 3: Methods

This chapter is comprised of five sections. It begins with a discussion of selection criteria used to determine PMHS suitability for this study. Next the instrumentation process for the subject is described, split into subsections that focus on a particular instrument being used. Third, general setup of each subject and its positioning is explained. Then the impact event is discussed. The final section details the collection of data from the test and how it was analyzed.

3.1 PMHS Selection and Preparation

All subjects used in this study were received from the Ohio State University Willed Body Program, with approval from the Ohio State University Biomedical Sciences Human Subjects Review Committee approved this project.

Testing for this project required the use of nine unembalmed fresh PMHS. In order to determine a PMHS’s suitability for use in this project, the following criteria were developed:

1) PMHS were received as close to their time of death as possible. Attempts were made to test the subjects within 72 hours of death Subjects of any age or gender
were accepted because age is not always a good predictor of bone integrity and overall health.

2) Subjects were accepted if their Body Mass Index (BMI) was calculated to be in the “Normal” or “Overweight” range because the 50th percentile male has a BMI that falls directly in between these ranges. If the subject was over 210 pounds then the subject was not used because it would have been difficult to move and position during testing.

3) Scarring on the subject was documented and cross referenced with known medical history to determine if the subject received surgery or some other invasive procedure that may have compromised the structural integrity of the thorax.

4) A DEXA scan was performed on each subject to determine their bone mineral density (BMD). If the subject received a Young Adult T-Score of -2.5 or less, then the subject was considered to be osteoporotic and was eliminated from consideration.

5) When available, blood tests were performed to check for the presence of communicable diseases, particularly Hepatitis B Virus and Human Immunodeficiency Virus. Positive results from this blood test immediately resulted in the disqualification of that subject from testing. In cases where blood
tests were unable to be performed, medical histories were queried to ensure the safety of lab personnel.

A quantitative algorithm was not employed for selecting subjects, as violation of any of the above rules resulted in disqualification of the subject. Figure 3.1, shown below, is representative of the selection criteria that were documented for each subject.
Qualifications for Acceptance into Oblique Thorax Study

- Males and females of any age are acceptable, with preference for those under the age of 75
  - 87 years is an acceptable age
- The cadaver should be classified as normal or overweight, per BMI classification
  - The 50th Percentile Male is 170 lb and 5’10”. This is almost exactly in between the normal and overweight BMI range. It is for this reason that we will accept cadavers that fall within either range.
  - Cadaver BMI rating is denoted by red X in Figure 2
- The weight should be below 210 lb
  - As larger cadavers are difficult to move
- Scarring on thorax of cadaver can be acceptable, given the scarring is not due to invasive surgery or other events that may have compromised the structural integrity of the thorax
  - Scarring was noted on the abdomen, inferior of the thorax
- The cadaver should not be osteoporotic, with preference given to subjects with normal BMD levels (Young Adult T-Score of -2.5 or above)
  - Young Adult T-Score of -2.3 is acceptable

Figure 3.1: Representative selection criteria scheme
Once a PMHS was approved for this test series, the subject was washed with a 10% bleach solution externally and in all orifices to remove potential infectious materials. The orifices were then packed with absorbent towels to prevent leakage of fluids from the subject throughout testing. A plastic bag was also placed over the head of the subject to collect any potential emissions from any orifices. The subject was then placed in a supine position on a metal table. Using a fluoroscope, the level of the T4 vertebra was delineated. After marking this location on the skin of the subject (which was later used to identify placement of the spinal 3aω motion block), the subject was ready to be instrumented.
3.2 Instrumentation

3.2.1 Pressurization

Initial instrumentation of the subject involved the placement of pressurization instrumentation. The purpose of this instrumentation was to allow the thoracic vasculature of the subject (including the superior vena cava (SVC), inferior vena cava (IVC), aortic arch, and descending aorta) to be pressurized with saline at a normal physiologic level. Sensors placed inside the thoracic vasculature could also be used to measure the change in pressure that occurred during the impact event.

Figure 3.2 and 3.3 show the pressurization instrumentation that was used in each subject. Binding of the Millar pressure sensor (Millar Instruments, Inc., Texas) and angiographic catheter was done with heat shrink wrap at five locations along the length of the angiographic catheter. The purpose of this binding was to make placement of the Millar sensor in the Foley catheter easier, using the stiffer angiographic catheter as a guide. The sensor and catheter were fed through a compression fitting to prevent leakage, plastic Y-fitting for connecting saline supply lines, and Foley catheter for occluding and supplying saline to the thoracic vasculature. The sensor at the end of the Millar wire is extended beyond the end of the Foley catheter approximately 1-2 cm so that the pressure reading is not affected by flow edge effects at the end of the catheter. Four pressurization instruments were assembled. Two of the assemblies were each fitted with two Millar pressure sensors and two assemblies were each fitted with one.
Figure 3.2: Close-up of pressurization instrumentation
To install the pressurization instrumentation in the subject, incisions were nominally made in three locations. On the right and left sides of the subject’s neck, incisions were made slightly anterior to the sternocleidomastoid muscle. The third incision was made in the right leg, just inferior to the anterior-superior iliac spine, slightly medial to the midline of the thigh. In the right neck incision, taking care not to damage internal structures, the right internal jugular vein was located. A string was tied as superiorly as possible around this vessel to inhibit flow up into the head. Additionally, the right common carotid artery was tied closed to prevent the saline from filling the head.
This process was repeated in the left neck incision for the left common carotid artery and the left internal jugular vein. The right femoral artery and vein were located in the right leg incision and each vessel was tied off as inferiorly as possible to stop flow into the right leg.

Once all of the vessels of the neck and leg were tied and the pressurization instrumentation was ready, incisions were made in the right internal jugular vein, left common carotid artery, right femoral artery, and right femoral vein. The sensor end of the instrumentation bundle was inserted into each of the incisions. The arterial insertions received pressurization instruments with two Millar pressure sensors, while the venous insertions were instrumented with the single Millar pressure sensor assemblies.

The 0905OTH55L01 and 0906OTH45L01 tests included a flushing of the vasculature with saline. These were the only tests in this series to receive this treatment which was a result of continually evolving the pressurization procedure. Flushing of the vasculature was done before the instrumentation was installed, and involved pumping saline into the neck vasculature and allowing it to flow out from the femoral vasculature. Saline was pumped through until the flow leaving the subject changed from red fluid to a colorless fluid. All fluid that came out of the subject through this process was collected in a beaker and compared qualitatively with the volume of fluid that was pumped into the subject to be sure that extra fluid was not left in the subject.

The balloons at the end of the Foley catheters were used to occlude the vasculature so that any saline pumped into the subject for the test would occupy only the
vessels of interest for this study. The superior arterial balloon was placed at the inferior end of the left common carotid artery. This would inhibit saline flow into the head and place the pressure sensor instrumentation in the aortic arch. The superior venous balloon was placed in the right braciocephalic vein. This was also to block saline flow to the head and right arm of the subject, placing the pressure sensor in the SVC superior to the heart. Both the inferior arterial and inferior venous balloons were placed at the level of the diaphragm. The inferior arterial balloon occluded the descending aorta, placing the pressure sensors in the same. The inferior venous balloon occluded the IVC, placing the pressure sensor in the same.

To assist in placing the Foley catheter balloons, real time fluoroscopy was used. Three to five milliliters of radio-opaque contrast solution was injected into the Foley catheter balloons to make them visible in the x-ray. Additionally, contrast fluid was injected into the vasculature through the angiographic catheters, so that the vessel that the instrumentation was in could be identified. Once it was determined that the Foley catheter balloon was in the correct location, air was injected into the balloon until significant resistance was met, indicating the balloon was fully occluding the vessel.

Once the pressurization instrumentation was completed pre-test imaging was completed to detect potential pre-existing fractures in the subject. The imaging was a computed tomography (CT) scan that was used to create a 3D reconstruction of the subject’s bony structures. Figure 3.4 shows a representative reconstructed scan. Note that the Foley catheters are also visible because of the contrast fluid inside the balloons.
Figure 3.4: 3D CT reconstruction, anterior to posterior view
3.2.2  Rib Strain

Uniaxial strain gauges were placed on ribs three through ten on the left and right sides of each subject. These gauges were able to record the strain on the outer surface of each rib throughout the impact event. If a rib fractured during the test, the strain gauge on that rib should indicate a sudden change in its signal. Therefore the strain gauge data could be used to determine a time of fracture, and ultimately a time associated with achieving AIS3+ injury.

Installing the strain gauges involved making 20 cm slits in the sides of the subject along the length of each rib being instrumented. Once the soft tissue had been removed from the rib’s surface it was made dry by wiping it with an ether soaked gauze pad and a strain gauge was glued on the rib. Care was taken to run the strain gauge wires away from the impact site to avoid interactions with the impacting ram. Placement of the strain gauges was also modified based on the type of impact that was to be performed. If the impact was from a lateral direction, then the strain gauges on the left side of the subject were placed on the oblique aspect of the subject’s thorax. Conversely, in an oblique impact the gauges were placed as laterally as possible. On the right side of the subject the gauges were always on the oblique aspect of the thorax. Once the gauges had all been placed the incisions were sutured shut and the strain gauge wires were bundled and sutured to the subject’s skin ensuring strain relief.
3.2.3 Accelerometers and Angular Rate Sensors

To capture the kinematic motion of the thorax during the impact event, 3aω motion blocks were constructed. These blocks, such as the one pictured in Figure 3.5, have three uniaxial accelerometers and three angular rate sensors that allow for linear acceleration and angular velocities to be recorded in the x-, y-, and z-directions. Two 3aω blocks were mounted on each subject.

Figure 3.5: 3aω motion block
One of the motion blocks was mounted on the spine of the subject. Using the site that was identified for T4 using the fluoroscope, a vertical incision was made along the spinous processes of the vertebrae. Taking care to disrupt as little of the musculature attachments as possible, the soft tissue surrounding the spinous process of T4 was pulled laterally to allow for removal of the spinous process of T4. Upon removal, a mount for installing the 3aω block was screwed through the neural arch and into the body of T4. The other motion block was attached to the sternum body. A vertical incision was made at approximately the midline of the sternum body. Subcutaneous tissue and musculature was removed to accommodate a flat mounting plate. This plate was screwed to the sternum body in a vertical manner. Figure 3.6 shows the mounts that were used.

Figure 3.6: 3aω motion block mounts
3.2.4 Chestband

In order to measure the deflection of the thorax an External Peripheral Instrument for Deformation Measurement, or chestband, was used. The chestband was comprised of 40 strain gauges affixed to a metal strip that have been covered with a flexible rubber. Using the strain that was measured by the strain gauges, a radius of curvature was calculated for each of the gauges. This device was wrapped around the subject at the level of the xyphoid process. Because the impact was applied at this level in each test, the chestband measured the deformation of the thorax in the X-Y plane of impact. The chestband was secured tightly around the body of the subject by taping the overlapping ends of the band and then pressurizing the lungs of the subject. Once the chestband was securely in place, measurements were taken of the subject’s chest depth, chest breadth, oblique dimensions, and chest circumference.

Post processing of the strain curvatures was done with a MATLAB (The MathWorks, Massachusetts) script called CrashStar v1.1. This script was developed to take chestband curvature inputs and output the planar coordinates of each gauge site. Using these coordinates and knowing the gauge nearest the impact site and its opposing gauge, the deflection of the thorax was determined. This technique has been used previously by researchers [6, 9, 11, 12, 18].
3.3 Positioning

Once instrumentation phase was complete, the subject was placed on a lift table in front of the impactor ram. All of the tests performed in this series were left side impacts. The subject was fitted with a head harness that hooked onto an electromagnetic release directly above the subject. Both the lift table and the head harness were adjusted along the Z-axis so that the center of the ram face was aligned with the xiphoid process of the subject. The subject was then adjusted in the X-Y plane for the correct loading direction (i.e. lateral or oblique).

In the case of a lateral test, the subject was situated so that the line of action of the ram was perpendicular to the X-Z plane created by the subject’s sternum and spine, with the point of first contact between the ram face and subject at the most lateral aspect of the thorax. For the oblique tests, two beams were bolted together to create a 30° angle. Aligning one of the beams with the line of action of the ram, the other beam was used to line up the back of the subject. The result of this rotation was that the impact face of the ram would make first contact with the thorax 30° anterior of lateral. Figures 3.7 and 3.8 show how the subject was rotated with respect to the impacting ram.
To try to minimize the amount of rotation introduced to the subject during impact the ram was aligned with the center of gravity (CG) of the subject. The CG of the subject was estimated to be 44% of the chest depth of the subject, measured from the spine toward the sternum [18]. Because the ram face plate had a width that was larger than the
average chest depth of the subjects (305 mm and 194 mm, respectively), this estimate of the thorax CG was expected to minimize subject rotation.

To ensure no interaction of the ram face with the subject’s left arm, the arms were folded and lifted out in front of the subject perpendicular to the body. All of the subjects were placed roughly the same distance from the ram face based on first contact occurring just after the ram had passed through its light trap. If necessary, the legs of the subjects were propped up by placing blocks under the feet and tying string around the legs, just proximal to the knee. This was to ensure proper seating of the subject, keeping them from sliding off of the lift table when the electromagnetic hook was released. An example of subject positioning and setup is shown in Figure 3.9.
Figure 3.9: Overview of test setup
Once the subject was in its final position, a FARO Arm point digitizing system [FARO Technologies, Inc., Florida] was used to locate motion blocks, bony landmarks, and the plane of the chestband. The output of this system was global coordinate data in the X-, Y-, and Z-directions for all points taken. This would allow the motion block data to be transformed to the subject’s body coordinate system.

3.4 Test Performance Phase

Once the PMHS was in final position, the subject was ready to be impacted. With two saline reservoirs positioned at heights (venous 47 cm above the subject’s heart and arterial 144 cm above subject’s heart) to pressurize the subject to living physiologic pressures, the superior vascular insertions were connected to the saline tubs. The inferior vascular insertions were connected to vertical tubes to estimate the pressure achieved inside the subject. Kelly Green clothing dye was added to the saline in each reservoir so that during autopsy the perfusion of saline into the subject could be determined. Flow of saline into the subject was turned on and allowed to run for two to three minutes prior to impact to allow the subject to reach a stable internal pressure.

The ram used for the impact event was a 23 kg pneumatic ram. Using an accelerometer mounted to the ram, the velocity of the ram was recorded through impact. Maximum deflection of the subject’s thorax was reached well before the ram was extended. Therefore, maximum extension of the ram is not thought to have affected the
results of the impact. The face plate of the ram was a 15.25 cm x 30.48 cm x 0.635 cm rectangle made of 6061 T6 Aluminum.

The nominal velocity of the impacts was determined using previously obtained data from other researchers. Using a method established by Shaw, an energy balance was used to determine speeds that would result in a 50% or 75% chance of achieving AIS3+ injury. This calculation was based on data provided by Viano and Kuppa. It was determined that the velocity required for a 50% chance of AIS3+ injury was approximately 4.5 m/s and for a 75% chance of AIS3+ injury was approximately 5.5 m/s.

Because the impacts in these tests were meant to cause injury, each subject was impacted only once. Impacts were randomly assigned both a loading direction and impact speed. The resulting test matrix is shown in Table 3.1.

<table>
<thead>
<tr>
<th>Subject</th>
<th>0802</th>
<th>0803</th>
<th>0804</th>
<th>0901</th>
<th>0902</th>
<th>0903</th>
<th>0904</th>
<th>0905</th>
<th>0906</th>
</tr>
</thead>
<tbody>
<tr>
<td>Direction</td>
<td>Lateral</td>
<td>Oblique</td>
<td>Oblique</td>
<td>Oblique</td>
<td>Lateral</td>
<td>Lateral</td>
<td>Lateral</td>
<td>Oblique</td>
<td>Oblique</td>
</tr>
<tr>
<td>Speed</td>
<td>4.5 m/s</td>
<td>4.5 m/s</td>
<td>4.5 m/s</td>
<td>4.5 m/s</td>
<td>4.5 m/s</td>
<td>4.5 m/s</td>
<td>5.5 m/s</td>
<td>5.5 m/s</td>
<td>4.5 m/s</td>
</tr>
</tbody>
</table>

Table 3.1: Testing matrix

When the subject was released from the electromagnetic hook and impacted by the ram, it was free to fall to rest. In order to prevent injury from impacts not related to the test, a Secondary Injury Protection Unit (SIPU) was placed to the right of and behind
the subject. This was a largely foam device that provided support, keeping the PMHS from falling to the floor or otherwise impacting the test apparatus.

Once the test was complete the subject was taken to have post-test CT done, again to indicate injury that may have occurred. The following day a thorax dissection was done to look for injury. After removing the skin and subcutaneous tissue the muscles were inspected for damage. Reflecting back the pectoral muscles revealed the ribs. Each rib was cleaned of all soft tissue and inspected all along its length to determine if fracture had occurred. The chest plate was removed and the pleural surface of the rib cage was also inspected for fracturing. Thoracic viscera were examined for injury, particularly the arch of the aorta at the ligamentum arteriosum to inspect for aortic rupture. The superior aspect of the abdomen was also inspected to determine if damage had occurred to the liver or spleen.

3.5 Data Collection and Reduction

Data was collected using a 96-channel data acquisition system [Yokogawa Electric Corporation, Japan]. Signals from each sensor were sent to a central data collection system where analog to digital conversion was performed at a sampling rate of 20 kHz. Bias was removed and the data was filtered according to SAE J211 standards where appropriate. A configuration file is included in Appendix E that represents a typical sensor setup used in these tests.
The 3aω motion blocks that were mounted to the spine and sternum were able to capture motion data in three orthogonal axes, but the orientation of these sensors was not well aligned with the subject’s body coordinate system (SAE J211). The FARO data that was collected before the impact was used to establish a body coordinate system for the subject and transform the accelerometer and angular rate sensor data so that it better represents the motion of the subject over time. Figure 3.10 depicts the difference between the subject’s body coordinate system (x, y, z) and the 3aω motion block’s coordinate system (x′, y′, z′).
To perform the transformation of the 3aω motion blocks into the subject’s body coordinate system, the coordinate system of the motion blocks must be established. FARO data included points taken at three vertices of the face of the motion block. These point were used to find two axes (typically y’ and z’) and the cross product of these vectors was taken to get the normal vector (x’). Similarly for the subject, anatomical landmark FARO data was used to create a y-z plane through the subject (from the
acromions of the shoulder down to the greater trochanters of the femurs). The normal to this plane was the subject’s x-axis. Unit vectors of the 3aω motion block coordinate system \((i', j', k')\) and the subject’s body coordinate system \((i, j, k)\) were then calculated.

One way to express the transformation matrix is as the cross product of the subject’s body coordinate system with the unit vectors of the 3aω motion block.

\[
A = \begin{bmatrix} x \\ y \\ z \end{bmatrix} \times \begin{bmatrix} i' \\ j' \\ k' \end{bmatrix} = \begin{bmatrix} x \cdot i' \\ x \cdot j' \\ x \cdot k' \end{bmatrix} \times \begin{bmatrix} y \cdot i' \\ y \cdot j' \\ y \cdot k' \end{bmatrix} \times \begin{bmatrix} z \cdot i' \\ z \cdot j' \\ z \cdot k' \end{bmatrix} \tag{3.1}
\]

Additionally, the transformation matrix can be written based on Euler angles based on rotations about the z-axis, \(x'\)-axis, and \(z'\)-axis.

\[
A = \begin{bmatrix} \cos \phi \cos \sigma - \sin \phi \cos \theta \sin \sigma & -\cos \phi \sin \sigma - \sin \phi \cos \theta \cos \sigma & \sin \phi \sin \theta \\
\cos \phi \cos \sigma + \sin \phi \cos \theta \sin \sigma & -\sin \phi \sin \sigma + \cos \phi \cos \theta \cos \sigma & -\cos \phi \sin \theta \\
\sin \theta \sin \sigma & \sin \theta \cos \sigma & \cos \theta \end{bmatrix} \tag{3.2}
\]

Where:
- \(\phi\) is rotation about the z-axis
- \(\theta\) is rotation about the \(x'\)-axis
- \(\sigma\) is rotation about the \(z'\)-axis

Setting the above transformation matrices equal to each other will provide values for the initial Euler angles between the two coordinate systems.
The relation between the angular velocity of the body fixed coordinate system 
\((\omega'_x, \omega'_y, \omega'_z)\) to the angular velocity of the Euler angles is shown below.

\[
\begin{bmatrix}
\dot{\phi} \\
\dot{\theta} \\
\dot{\sigma}
\end{bmatrix} =
\begin{bmatrix}
\sin \theta \sin \sigma & \cos \sigma & 0 \\
\sin \theta \cos \sigma & -\sin \sigma & 0 \\
\cos \theta & 0 & 1
\end{bmatrix}^{-1}
\begin{bmatrix}
\omega'_x \\
\omega'_y \\
\omega'_z
\end{bmatrix}
\] (3.3)

The next set of Euler angles is then simply:

\[
\varphi_2 = \varphi_1 + \dot{\varphi}_1 \times \Delta t
\] (3.4)

\[
\theta_2 = \theta_1 + \dot{\theta}_1 \times \Delta t
\] (3.5)

\[
\sigma_2 = \sigma_1 + \dot{\sigma}_1 \times \Delta t
\] (3.6)

Using these new Euler angles, and the associated time history of the Euler angle rotations, subsequent angles can be found using the following equations:

\[
\varphi_{i+1} = \varphi_i + \left( \frac{\dot{\varphi}_i + \dot{\varphi}_{i+1}}{2} \right) \times \Delta t
\] (3.7)

\[
\theta_{i+1} = \theta_i + \left( \frac{\dot{\theta}_i + \dot{\theta}_{i+1}}{2} \right) \times \Delta t
\] (3.8)

\[
\sigma_{i+1} = \sigma_i + \left( \frac{\dot{\sigma}_i + \dot{\sigma}_{i+1}}{2} \right) \times \Delta t
\] (3.9)

Using these Euler angle time histories, a transformation matrix can be built for each time point sampled. This transformation matrix time history can then be used to transform the accelerometer and angular rate sensor data to get meaningful motion data.
for the thoracic spine and sternum body. An example of the MATLAB scripts used to transform the data is included in Appendix B.

The chestband that was used in this test series provided a time history of radius of curvature changes around the entire thorax in the plane of impact. The MATLAB script CrashStar v1.1, as mentioned previously, was used to turn the curvature data into contours that represent the shape of the thorax at each time step where data was sampled. The script analyzed the curvature data that was provided and created a closed oval-like shape that closely matched the circumference that was measured before the test. Based on these contours, the script was then able to provide planar coordinates that represent the location of each of the gauges of the chestband over time. The gauge that was closest to the impact site and the gauge opposite that site along the line of action of the ram were documented before the test was fired. Using those two gauges the deflection of the thorax was calculated.
It is difficult to compare the responses of two different PMHS directly to each other due to variance between subjects. Because of this it was important to scale the data that were collected from these tests according to a standard. Three methods have been discussed in the literature to achieve this scaling, typically to some parameter of the 50th percentile male. The first method was developed by Eppinger [8] and assumes that density and the elastic modulus across all subjects are equivalent. The second method explored was described by Mertz and Viano [9] and assumes that all subjects exhibit an equivalent stiffness and share geometric similitude in the thorax. The third method proposed by Moorhouse [20] employs similar assumptions from the Mertz and Viano
method, except that stiffness was not considered to be equivalent across subjects. Because of the deviation in impact velocity for the tests in this series, appropriate terms were also scaled by the ratio between the nominal test velocity and the true impact velocity.

In the Eppinger method, simple ratios of the mass of the 50th percentile male (76 kg) to the total body mass of the PMHS are used to scale time, length, force, and acceleration. Following this equation:

\[ \lambda = \frac{M_{50th}}{M_{sub}} \]  

(3.10)

In all of these equations, “50th” refers to the 50th percentile male and “sub” refers to the subject being scaled. Thusly, the PMHS measurements listed above are scaled according to the following equations:

\[ L_{50th} = \left( \frac{V_{50th}}{V_{sub}} \right) \lambda^{1/3} L_{sub} \]  

(3.11)

\[ F_{50th} = \left( \frac{V_{50th}}{V_{sub}} \right) \lambda^{2/3} F_{sub} \]  

(3.12)

\[ A_{50th} = \left( \frac{V_{50th}}{V_{sub}} \right) \lambda^{-1/3} A_{sub} \]  

(3.13)

\[ T_{50th} = \lambda^{1/3} T_{sub} \]  

(3.14)
In the Mertz and Viano method, instead of a total body mass ratio, the effective mass of the thorax is calculated for each subject based on an impulse momentum analysis of the impact event. In all cases, the time over which the loading and associated change in subject thoracic velocity is from time zero out to the time where maximum deflection of the thorax occurs. The change in thorax velocity was calculated from the resultant velocity of T4 in the plane of the impact (namely the x-y plane). The effective mass is calculated from the following equation:

\[
m_{\text{eff}} = \frac{\int_0^T F(t) dt}{v_0}
\]  

(3.15)

Thusly, an effective mass is calculated for each subject independently of the other subjects. Because no 50th percentile male “effective thorax mass” is known, the values calculated for the subjects are averaged together to create a “50th percentile effective thorax mass”. The mass ratio is then calculated similarly to the Eppinger method:

\[
\lambda_m = \frac{M_{\text{eff}} - 50\text{th}}{M_{\text{eff}} - \text{sub}}
\]  

(3.16)

Additionally, a stiffness ratio is calculated. Because this method assumes that there is geometric similitude and an equivalent stiffness across all subjects, the stiffness ratio of some characteristic length of the thorax. In these tests, chest breadth (50th percentile male, 349 mm) was used:
\[ \lambda_k = \frac{\text{Chest Breadth}_{50th}}{\text{Chest Breadth}_{sub}} \]  
(3.17)

The scaling values were then determined by modeling the system as a two mass, single spring system:

\[ T_{50th} = \sqrt{\frac{\lambda_m}{\lambda_k}} \sqrt{\frac{(m_p + m_{sub})}{(m_p + m_{50th})}} T_{sub} \]  
(3.18)

\[ a_{50th} = \left( \frac{V_{50th}}{V_{sub}} \right)^{\frac{1}{2}} \sqrt{\frac{\lambda_k}{\lambda_m}} \sqrt{\frac{(m_p + m_{sub})}{(m_p + m_{50th})}} a_{sub} \]  
(3.19)

\[ L_{50th} = \left( \frac{V_{50th}}{V_{sub}} \right)^{\frac{1}{2}} \sqrt{\frac{\lambda_m}{\lambda_k}} \sqrt{\frac{(m_p + m_{sub})}{(m_p + m_{50th})}} L_{sub} \]  
(3.20)

\[ F_{50th} = \left( \frac{V_{50th}}{V_{sub}} \right)^{\frac{1}{2}} \sqrt{\frac{\lambda_m}{\lambda_k}} \sqrt{\frac{(m_p + m_{sub})}{(m_p + m_{50th})}} F_{sub} \]  
(3.21)

Finally, the Moorhouse method is very similar to the Mertz and Viano “effective mass” method. The scaling factors are calculated in exactly the same way, except instead of using a ratio of characteristic lengths for the stiffness ratio, an effective stiffness of the thorax is calculated based on the energy absorbed by the thorax during impact. To calculate that energy, the loading history is integrated over the deflection history from time zero out to the time of maximum deflection. Using equation 3.22, an effective stiffness for the thorax can then be calculated:
\[ k_{\text{eff}} = \frac{2 \int F dx}{x_{\text{max}}^2} \]  \hspace{1cm} (3.22)

Again, as was the case with the calculated effective stiffness, there is no existing “50th percentile male effective stiffness”. So the stiffnesses of each subject are averaged to create this 50th percentile male value. The stiffness ratio is then calculated as follows:

\[ \lambda_k = \frac{K_{\text{eff-}50\text{th}}}{K_{\text{eff-}sub}} \]  \hspace{1cm} (3.23)

Simply replacing \( \lambda_k \) in the Mertz and Viano scaling equations with the Moorhouse \( \lambda_k \) will provide new scaling factors for all data.

Of course, normalizing the data for all of the tests changes the time over which the data was collected, which in turn modifies the sampling rate. It is then difficult to compare normalized data between subjects that have differing sampling rates. Using built in functions in MATLAB, all of the data was resampled at 20 kHz to restore the sampling rate that was used in the data acquisition process. The routine employed a linear interpolation to recalculate the time histories at the desired sampling rate.

In order to compare the ability of the different normalization methods to group the data, a percent cumulative variation (%CV) was calculated for the load time histories, the deflection time histories, and the force v. deflection curves. This was done for the time histories by determining what data made up the top 80% of the curve. Then, the standard deviation at one time point across grouped tests was divided by the average value of that
group at that time point. This was done for all time points within the top 80% of the curve. The average of these %CV values was reported as the %CV for that grouping of tests. For the force v. deflection data the calculation was slightly different. The error ellipse at each data point was calculated by using the equation for the area of an ellipse:

\[ A = \pi \times a \times b \]  

(3.24)

Where:
- \( a \) is the standard deviation of the force
- \( b \) is the standard deviation of the deflection

These areas were summed over the portion of the force deflection curve that corresponded to the top 80% of the force data and the top 80% of the deflection data. Finally, a percent improvement for the normalization methods was calculated by comparing the %CV and error ellipse sums for each normalization method to the %CV and error ellipse sums for the non-normalized data.
Chapter 4: Results

This chapter details the data that was collected from the PMHS tests that were conducted. Most data involved impact forces, thoracic deflections, and injuries incurred. The normalization process involved only the 4.5 m/s tests because the low number of 5.5 m/s tests only complicated the variance seen in the data due to their higher energies. Appendix C includes a sample processing script that was used to filter and remove bias from the data.

4.1 Subject Data

Anthropometric data taken for each subject is found in Table 4.1. This data was recorded using the Pre-Test Data Sheet that has been included in Appendix D. Of the nine subjects that were used in this study, seven of them were male and two were female. The average age across all subjects was 80. Their stature averaged 171 cm ($\sigma=11.12$ cm) and their masses averaged 75.1 kg ($\sigma=14.3$ kg).
<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age</th>
<th>Height cm</th>
<th>Mass kg</th>
<th>BMD Young Adult T-Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>0802</td>
<td>Male</td>
<td>67</td>
<td>169</td>
<td>81.6</td>
<td>-1.1</td>
</tr>
<tr>
<td>0803</td>
<td>Male</td>
<td>81</td>
<td>167.6</td>
<td>79.4</td>
<td>0.2</td>
</tr>
<tr>
<td>0804</td>
<td>Male</td>
<td>81</td>
<td>175.3</td>
<td>92.5</td>
<td>-1.1</td>
</tr>
<tr>
<td>0901</td>
<td>Male</td>
<td>82</td>
<td>175</td>
<td>72.1</td>
<td>-0.3</td>
</tr>
<tr>
<td>0902</td>
<td>Male</td>
<td>77</td>
<td>178</td>
<td>63.5</td>
<td>-1.2</td>
</tr>
<tr>
<td>0903</td>
<td>Female</td>
<td>81</td>
<td>159</td>
<td>52.3</td>
<td>-0.3</td>
</tr>
<tr>
<td>0904</td>
<td>Female</td>
<td>88</td>
<td>150</td>
<td>65.5</td>
<td>-1.4</td>
</tr>
<tr>
<td>0905</td>
<td>Male</td>
<td>87</td>
<td>186</td>
<td>71.2</td>
<td>-2.4</td>
</tr>
<tr>
<td>0906</td>
<td>Male</td>
<td>76</td>
<td>180</td>
<td>97.5</td>
<td>-1.7</td>
</tr>
<tr>
<td>Average</td>
<td></td>
<td>80</td>
<td>171</td>
<td>75</td>
<td></td>
</tr>
</tbody>
</table>

Table 4.1: Subject data

The goal of this testing was to characterize the response of the PMHS thorax when impacted at speeds that would produce injury. As such, autopsies were completed following each test and the results of those autopsies are listed in Table 4.2. The locations of rib fractures measured during autopsy are included as Figures 4.1 through 4.8. The solid lines in the rib fracture figures are meant to indicate the lateral most aspect of the thorax for each subject, helping to delineate anterior and posterior fractures. The sternal notch served as the origin for all of the measurements that were taken.
<table>
<thead>
<tr>
<th>Subject</th>
<th>Injury Type</th>
<th>Injury Locations (Number of Fractures)</th>
<th>AIS Level</th>
</tr>
</thead>
<tbody>
<tr>
<td>0802</td>
<td>None</td>
<td>None</td>
<td>0</td>
</tr>
<tr>
<td>0803</td>
<td>Rib Fracture</td>
<td>Left Rib 4 (1) Left Rib 5 (1) Left Rib 6 (1)</td>
<td>2</td>
</tr>
<tr>
<td>0804</td>
<td>Rib Fracture</td>
<td>Left Rib 3 (2) Left Rib 4 (2) Left Rib 5 (2) Left Rib 6 (2)</td>
<td>3</td>
</tr>
<tr>
<td>0901</td>
<td>Rib Fracture</td>
<td>Left Rib 3 (1) Left Rib 4 (1) Left Rib 5 (3) Left Rib 6 (2) Left Rib 8 (1)</td>
<td>3</td>
</tr>
<tr>
<td>0902</td>
<td>Rib Fracture</td>
<td>Left Rib 4 (1) Left Rib 5 (1) Left Rib 6 (1) Left Rib 7 (2) Left Rib 8 (2) Left Rib 9 (2) Left Rib 10 (2)</td>
<td>3</td>
</tr>
<tr>
<td>0903</td>
<td>Rib Fracture</td>
<td>Left Rib 4 (3) Left Rib 5 (2) Left Rib 6 (2) Left Rib 7 (2) Left Rib 8 (2) Left Rib 9 (2) Left Rib 10 (2) Left Rib 11 (2) Left Rib 12 (1)</td>
<td>4</td>
</tr>
<tr>
<td>0904</td>
<td>Rib Fracture</td>
<td>Left Rib 3 (3) Left Rib 4 (2) Left Rib 5 (2) Left Rib 6 (2) Left Rib 7 (2) Left Rib 9 (2) Left Rib 10 (2)</td>
<td>4</td>
</tr>
<tr>
<td>0905</td>
<td>Rib Fracture</td>
<td>Left Rib 3 (1) Left Rib 4 (1) Left Rib 5 (2) Left Rib 6 (1) Left Rib 7 (1) Left Rib 8 (1) Left Rib 9 (1) Left Rib 10 (1)</td>
<td>4</td>
</tr>
<tr>
<td>0906</td>
<td>Rib Fracture</td>
<td>Left Rib 4 (1) Left Rib 5 (1)</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td>Spleen Rupture</td>
<td>Major Devascularization, No Hilar Injury</td>
<td>4</td>
</tr>
</tbody>
</table>

Table 4.2: Summary of injury by subject
Figure 4.1: Rib fracture locations for test 0803OTH45L01, only anterior fractures
Figure 4.2: Rib fracture locations for test 0804OTH45L01
Figure 4.3: Rib fracture locations for 0901OTH45L01
Figure 4.4: Rib fracture locations for test 0902LTH45L01
Figure 4.5: Rib fracture locations for test 0903LTH45L01
Figure 4.6: Rib fracture locations for subject 0904LTH55L01
Figure 4.7: Rib fracture locations for test 0905OTH55L01
4.2 Impact Results

Table 4.3 shows a summary of the impact results from each test. Seven of the nine tests had a nominal test velocity of 4.5 m/s and the other impacts had a nominal velocity of 5.5 m/s. Because the impacting ram was driven by a pneumatic system and inherent drag from the linear potentiometer, fine control of the impact velocity was not possible. The resulting average impact velocity for the 4.5 m/s tests was 4.66 m/s ($\sigma=0.14$ m/s) and for the 5.5 m/s tests was 5.57 m/s ($\sigma=0.1$ m/s). In order to calculate the impact force, an accelerometer mounted on the ram was used. The force was determined using the equation below:
\[ \text{Ram Force} = \text{Accelerometer Signal} \times 9.80665 \frac{m}{s^2} \times 22.99 \text{ kg} \quad (4.1) \]

The average of the impact forces for all of the 4.5 m/s tests was 2652 N ($\sigma=363.5$ N) and for the 5.5 m/s tests was 3459 N ($\sigma=8.5$ N). The average maximum compression for the 4.5 m/s tests was 19.8% ($\sigma=3.6\%$) and for the 5.5 m/s tests was 22.9% ($\sigma=2.7\%$).

<table>
<thead>
<tr>
<th>Impact Number</th>
<th>Type of Impact</th>
<th>$V_{\text{impact}}$</th>
<th>$F_{\text{max}}$</th>
<th>$D_{\text{max}}$</th>
<th>$C_{\text{max}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0802LTH45L01</td>
<td>270°</td>
<td>4.77 m/s</td>
<td>3192 N</td>
<td>64 mm</td>
<td>18.5 %</td>
</tr>
<tr>
<td>0803OTH45L01</td>
<td>300°</td>
<td>4.58 m/s</td>
<td>2713 N</td>
<td>56 mm</td>
<td>16.7 %</td>
</tr>
<tr>
<td>0804OTH45L01</td>
<td>300°</td>
<td>4.47 m/s</td>
<td>2557 N</td>
<td>71 mm</td>
<td>19.8 %</td>
</tr>
<tr>
<td>0901OTH45L01</td>
<td>300°</td>
<td>4.88 m/s</td>
<td>2708 N</td>
<td>56.6 mm</td>
<td>18.3 %</td>
</tr>
<tr>
<td>0902LTH45L01</td>
<td>270°</td>
<td>4.58 m/s</td>
<td>2379 N</td>
<td>79.7 mm</td>
<td>24.7 %</td>
</tr>
<tr>
<td>0903LTH45L01</td>
<td>270°</td>
<td>4.68 m/s</td>
<td>2079 N</td>
<td>67 mm</td>
<td>24.7 %</td>
</tr>
<tr>
<td>0904LTH55L01</td>
<td>270°</td>
<td>5.64 m/s</td>
<td>3465 N</td>
<td>66.6 mm</td>
<td>21 %</td>
</tr>
<tr>
<td>0905OTH55L01</td>
<td>300°</td>
<td>5.50 m/s</td>
<td>3453 N</td>
<td>73.5 mm</td>
<td>24.8 %</td>
</tr>
<tr>
<td>0906OTH45L01</td>
<td>300°</td>
<td>4.66 m/s</td>
<td>2939 N</td>
<td>54.7 mm</td>
<td>15.9 %</td>
</tr>
</tbody>
</table>

Table 4.3: Summary of raw impact data peak measurement

4.3 Scaling Factors

Using the normalization method discussed in section 3.5, scaling factors were calculated to adjust the data collected in these tests to represent a 50th percentile male impacted at our nominal test speeds. Table 4.4 contains data that was used to calculate factors for each of the normalization methods. Table 4.5 shows the scaling factors for the Eppinger method, Table 4.6 shows the scaling factors for the Mertz-Viano method, and Table 4.7 shows the scaling factors for the Moorhouse method.
<table>
<thead>
<tr>
<th>Test Number</th>
<th>Chest Breadth</th>
<th>Effective Mass</th>
<th>Characteristic Ratios</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>$\lambda_{m\text{-total body}}$</td>
<td>$\lambda_{m\text{-eff}}$</td>
</tr>
<tr>
<td>0802</td>
<td>314 mm</td>
<td>34.94 kg</td>
<td>0.93</td>
<td>1.03</td>
</tr>
<tr>
<td>0803</td>
<td>315 mm</td>
<td>33.96 kg</td>
<td>0.96</td>
<td>0.96</td>
</tr>
<tr>
<td>0804</td>
<td>342 mm</td>
<td>34.04 kg</td>
<td>0.82</td>
<td>0.95</td>
</tr>
<tr>
<td>0901</td>
<td>315 mm</td>
<td>37.42 kg</td>
<td>1.05</td>
<td>0.87</td>
</tr>
<tr>
<td>0902</td>
<td>314 mm</td>
<td>40.87 kg</td>
<td>1.20</td>
<td>0.88</td>
</tr>
<tr>
<td>0903</td>
<td>268 mm</td>
<td>18.08 kg</td>
<td>1.45</td>
<td>1.99</td>
</tr>
<tr>
<td>0906</td>
<td>322 mm</td>
<td>38.47 kg</td>
<td>0.78</td>
<td>0.84</td>
</tr>
</tbody>
</table>

Table 4.4: Summary of mass and stiffness ratios

<table>
<thead>
<tr>
<th>Test Number</th>
<th>Eppinger Scaling Factors</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Acceleration</td>
<td>Displacement</td>
</tr>
<tr>
<td>0802</td>
<td>0.97</td>
<td>0.92</td>
</tr>
<tr>
<td>0803</td>
<td>1.00</td>
<td>0.97</td>
</tr>
<tr>
<td>0804</td>
<td>1.07</td>
<td>0.94</td>
</tr>
<tr>
<td>0901</td>
<td>0.91</td>
<td>0.94</td>
</tr>
<tr>
<td>0902</td>
<td>0.93</td>
<td>1.04</td>
</tr>
<tr>
<td>0903</td>
<td>0.85</td>
<td>1.09</td>
</tr>
<tr>
<td>0906</td>
<td>1.03</td>
<td>0.89</td>
</tr>
</tbody>
</table>

Table 4.5: Eppinger normalization factors

<table>
<thead>
<tr>
<th>Test Number</th>
<th>Mertz-Viano Scaling Factors</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Acceleration</td>
<td>Displacement</td>
</tr>
<tr>
<td>0802</td>
<td>0.97</td>
<td>0.90</td>
</tr>
<tr>
<td>0803</td>
<td>1.07</td>
<td>0.92</td>
</tr>
<tr>
<td>0804</td>
<td>1.06</td>
<td>0.99</td>
</tr>
<tr>
<td>0901</td>
<td>1.09</td>
<td>0.85</td>
</tr>
<tr>
<td>0902</td>
<td>1.15</td>
<td>0.91</td>
</tr>
<tr>
<td>0903</td>
<td>0.65</td>
<td>0.99</td>
</tr>
<tr>
<td>0906</td>
<td>1.15</td>
<td>0.90</td>
</tr>
</tbody>
</table>

Table 4.6: Mertz-Viano normalization factors
<table>
<thead>
<tr>
<th>Test Number</th>
<th>Moorhouse Scaling Factors</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Acceleration</td>
</tr>
<tr>
<td>0802</td>
<td>0.80</td>
</tr>
<tr>
<td>0803</td>
<td>0.99</td>
</tr>
<tr>
<td>0804</td>
<td>1.28</td>
</tr>
<tr>
<td>0901</td>
<td>0.98</td>
</tr>
<tr>
<td>0902</td>
<td>1.20</td>
</tr>
<tr>
<td>0903</td>
<td>0.62</td>
</tr>
<tr>
<td>0906</td>
<td>1.02</td>
</tr>
</tbody>
</table>

Table 4.7: Moorhouse normalization factors

4.4 Chestband Contours

Deflection data from each impact was gleaned from chestband measurements. Using the output from the MATLAB script CrashStar v1.1, contours were produced for each time point sampled during the impact. Gauges on the chestband were determined prior to impact to be lying on the line of action of the ram, that is, one gauge at the impact site and its opposing gauge on the opposite side of the subject. The distance between these two gauges at the moment the ram face plate begins to engage the subject (the initial length) can be compared to the distance between these gauges at any other point during the impact event. Assuming that the rotation incurred by the thorax during impact is negligible, the change in the distance between these gauges will represent the deflection of the thorax during the impact. Figure 4.9 shows a comparison from a representative impact of the initial chestband orientation to its orientation at the time of maximum deflection. For the purpose of illustration, these contours both have the gauge located at the spine of the subject set as the origin.
4.5 Force vs. Time

Force curves are shown split into two groups, one for lateral impacts and one for oblique impacts. All of the data shown is normalized using one of the three methods discussed above. For comparison, the average of each grouping has been plotted in black and the upper and lower bounds, created by adding and subtracting one standard deviation from the average, respectively, have been plotted in gray. Eppinger normalization of each group is shown in Figures 4.10 and 4.11, Mertz-Viano normalization of each group is shown in Figures 4.12 and 4.13, and Moorhouse normalization of each group is shown in Figures 4.14 and 4.15.
Figure 4.10: Normalized load data from lateral tests, Eppinger method
Figure 4.11: Normalized load data from oblique tests, Eppinger method
Figure 4.12: Normalized load data from lateral tests, Mertz-Viano method
Figure 4.13: Normalized load data from oblique tests, Mertz-Viano method
Figure 4.14: Normalized load data from lateral tests, Moorhouse method
Deflection data vs. time for each test was also normalized. Again, the average of the responses for each group (lateral or oblique) is shown with a bold black line and the upper and lower bounds for each group are shown in gray. Figures 4.16 and 4.17 show Eppinger normalized deflections, Figures 4.18 and 4.19 show Mertz-Viano normalized deflections, and Figures 4.20 and 4.21 show Moorhouse normalized deflections.
Figure 4.16: Normalized deflection data from lateral tests, Eppinger method
Figure 4.17: Normalized deflection data from oblique tests, Eppinger method
Figure 4.18: Normalized deflection data from lateral tests, Mertz-Viano method
Figure 4.19: Normalized deflection data from oblique tests, Mertz-Viano method
Figure 4.20: Normalized deflection data from lateral tests, Moorhouse method
Figure 4.21: Normalized deflection data from oblique tests, Moorhouse method

4.7 Force vs. Deflection

Force vs. deflection curves are shown for all 4.5 m/sec tests. Tests were grouped into lateral and oblique and averaged. Similarly to the above plots, the average for each grouping is plotted in black and the upper and lower bounds are shown in gray. These bounds outline the corridor created from standard deviation, or error, ellipses at each data point. The horizontal axis of the error ellipse has a length of twice the standard deviation.
of the deflections at that data point and the vertical axis of the ellipse has a length twice the standard deviation of the forces at that data point. Figures 4.22 and 4.23 show Eppinger normalized data, Figures 4.24 and 4.25 show Mertz-Viano normalized data, and Figures 4.26 and 4.27 show Moorhouse normalized data.

Figure 4.22: Normalized force vs. deflection data from lateral tests, Eppinger method
Figure 4.23: Normalized force vs. deflection data from oblique tests, Eppinger method
Figure 4.24: Normalized force vs. deflection data from lateral tests, Mertz-Viano method
Figure 4.25: Normalized force vs. deflection data from oblique tests, Mertz-Viano method
Figure 4.26: Normalized force vs. deflection data from lateral tests, Moorhouse method
Figure 4.27: Normalized force vs. deflection data from oblique tests, Moorhouse method

4.8 Pressure

Table 4.8 shows the peak pressures that were measured in the major thoracic vasculature. An entry of N/A in the table means that either the data channel was lost or no instrument was placed in that location.
### Table 4.8: Summary of peak vascular pressures

<table>
<thead>
<tr>
<th>Subject Number</th>
<th>Aortic Arch 1 (kPa)</th>
<th>Aortic Arch 2 (kPa)</th>
<th>Descending Aorta 1 (kPa)</th>
<th>Descending Aorta 2 (kPa)</th>
<th>SVC (kPa)</th>
<th>IVC (kPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0802</td>
<td>10.6</td>
<td>15.1</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>0803</td>
<td>9.7</td>
<td>8.5</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>0804</td>
<td>4.7</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>0901</td>
<td>14.8</td>
<td>16.9</td>
<td>N/A</td>
<td>N/A</td>
<td>21.3</td>
<td>23.8</td>
</tr>
<tr>
<td>0902</td>
<td>15.3</td>
<td>15.9</td>
<td>N/A</td>
<td>18.9</td>
<td>11.6</td>
<td>N/A</td>
</tr>
<tr>
<td>0903</td>
<td>12.9</td>
<td>12.1</td>
<td>N/A</td>
<td>N/A</td>
<td>11.7</td>
<td>13.3</td>
</tr>
<tr>
<td>0906</td>
<td>17.2</td>
<td>16.1</td>
<td>15.9</td>
<td>15.3</td>
<td>5.1</td>
<td>N/A</td>
</tr>
</tbody>
</table>

4.9 Strain Gauge Data

Using the strain gauge data from the ribs, time of fracture could be estimated when the strain recording had a sudden change in value. An example plot of strain gauge readings is shown in Figure 4.28.
After autopsy was completed and all of the observable fractures were identified, timings were associated with the first fracture occurring on each rib. Table 4.9 has shows the estimated times of fracture for each subject, where an N/A indicates fracture was observed at autopsy, but timing estimation was not possible. All of the timings presented in the table are based on the non-normalized time data.
<table>
<thead>
<tr>
<th>Subject Number</th>
<th>Rib Fractured</th>
<th>Time of Fracture (msec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0802</td>
<td>None</td>
<td>N/A</td>
</tr>
<tr>
<td>0803</td>
<td>L4, L5, L6</td>
<td>15, 17</td>
</tr>
<tr>
<td>0804</td>
<td>L3, L4, L5, L6</td>
<td>N/A, N/A, N/A</td>
</tr>
<tr>
<td>0901</td>
<td>L3, L4, L5, L6</td>
<td>N/A, N/A, N/A</td>
</tr>
<tr>
<td>0902</td>
<td>L4, L5, L6, L7, L8, L9, L10</td>
<td>15, 24, 21, 19, 9, 8, 11</td>
</tr>
<tr>
<td>0903</td>
<td>L4, L5, L6, L7, L8, L9, L10</td>
<td>17, 13, 10, 10, 9, 8, N/A</td>
</tr>
<tr>
<td>0906</td>
<td>L4, L5</td>
<td>20, 18</td>
</tr>
</tbody>
</table>

Table 4.9: Summary of rib fracture time estimations
Chapter 5: Discussion

The purpose of this study was to explore the kinematic response of the PMHS thorax to impact at high speeds in both lateral and oblique directions. It has been reported in Shaw et al.’s previous work that the response of the intact thorax to impacts is dependent on direction of loading, where the lateral response of the thorax tends to be generally stiffer than the oblique response. That is to say, when observing Shaw’s force deflection plots, the lateral impacts result in a higher average load with less deflection compared to the oblique impacts, which have both lower average peak loads and larger average peak deflections. The speeds at which the tests in this study were conducted were meant to have a 50% chance of causing serious (AIS3+) injury to the subjects.

This study was largely based on the methods used by Shaw and the conclusions drawn from that data. One major difference between this study and the Shaw study is obviously the role that injury plays in the response of these subjects. With the use of strain gauges to identify times of injury for each subject, one of the hopes was that the response of the thorax could be tracked over the entire impact and compared to the response of the thorax before injury occurred. It will be shown that the timing of injury was very well defined for most cases using the strain gauge data. However, the normalization of the pre-AIS3+ injury data was not useful in observing trends between
lateral and oblique cases. As such, the normalization of the pre-AIS3+ data is not presented in this work.

5.1 Comparison of Normalization Methods

Several methods exist for normalizing data to a 50th percentile male. As was discussed in the methods section, normalization can be based on total mass ratios, effective mass calculations, and effective stiffness calculations. In this study, the time, force, and deflection data was normalized using all three methods in order to determine the method that provided the best grouping of the data by loading direction. In order to compare the grouping ability of all of the methods, percent cumulative variation (%CV) was calculated. By comparing the variation that resulted in each method, a percentage improvement over the non-normalized data groupings could be determined. Table 5.1 shows the percentage improvement for each of the methods.

<table>
<thead>
<tr>
<th></th>
<th>Signal</th>
<th>Eppinger (% improvement)</th>
<th>Mertz-Viano (% improvement)</th>
<th>Moorhouse (% improvement)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lateral</td>
<td>Force</td>
<td>-20.70%</td>
<td>-16.49%</td>
<td>-17.89%</td>
</tr>
<tr>
<td></td>
<td>Deflection</td>
<td>-51.58%</td>
<td>-5.80%</td>
<td>24.65%</td>
</tr>
<tr>
<td></td>
<td>F-D</td>
<td>-141.64%</td>
<td>-11.78%</td>
<td>20.65%</td>
</tr>
<tr>
<td>Oblique</td>
<td>Force</td>
<td>10.73%</td>
<td>-0.09%</td>
<td>-8.24%</td>
</tr>
<tr>
<td></td>
<td>Deflection</td>
<td>-14.69%</td>
<td>-41.42%</td>
<td>47.59%</td>
</tr>
<tr>
<td></td>
<td>F-D</td>
<td>24.90%</td>
<td>-12.88%</td>
<td>49.81%</td>
</tr>
</tbody>
</table>

Table 5.1: Percent improvement for normalization methods
From this analysis, it appears that the Moorhouse method that employs the use of a calculated effective mass and a calculated effective stiffness of the thorax performs the best normalization of the data. There are drawbacks to the method. Namely that a 50th percentile male thorax effective mass does not exist, nor does a 50th percentile male thorax effective stiffness. Because of this an average for each of these values is used, and this average is based on the subjects that were a part of this study. It is also clear that possible outlier subjects can have their peak values changed to a value far from the other subjects in that group. As can be seen the Moorhouse normalized force history from Figure 4.15, subject 0804 had its peak force increased several hundred Newtons above the other lateral subjects. However, the normalization grouped most of 0804’s force curve with the other oblique subjects and the deflection histories are well grouped also. Increasing the sample size across all of the tests would help alleviate this phenomenon. All of the data presented in the rest of this paper uses the Moorhouse normalized data.

5.2 Force Responses

ISO 9790 provides force vs. time corridors for thorax impacts conducted at a nominal speed of 4.3 m/s. A crash test dummy with acceptable biofidelity is meant to exhibit a response that falls within this corridor. Figure 5.1 shows the corridor scaled to the standard’s nominal speed.
As was stated above, the ISO 9790 corridor was created for a nominal test speed of 4.3 m/s. In order to apply the data from the study to the corridor, it is worthwhile to scale the corridor to this study’s nominal speed. In order to do this the equation used to scale force is revisited.

\[ R_F = \frac{V_{sub}}{V_{sub}} \times \sqrt{\lambda_{k-eff} \times \lambda_{m-eff} \times \sqrt{\frac{M_p + M_{sub-eff}}{M_p + M_{50s-eff}}}} \]  \hspace{1cm} (5.1)
The forces that are supplied for the corridor have already been scaled to a standard subject, therefore $\lambda_{k\text{-eff}} = \lambda_{m\text{-eff}} = 1$, and $M_{p} + M_{\text{sub-eff}} = M_{p} + M_{50\text{th-eff}}$, so the resulting scaling factor is simply:

$$R_F = \frac{V_{50\text{th}}}{V_{\text{sub}}}$$

(5.2)

In the equation 5.2, $V_{50\text{th}}$ is the nominal velocity of interest and $V_{\text{sub}}$ is the velocity of the data that needs to be scaled. So, the 4.3 m/s corridor will be scaled up to meet the 4.5 m/s data that was created in this study. $R_F$ is $\frac{4.5 \text{ m/s}}{4.3 \text{ m/s}}$, or 1.05. Figure 5.2 shows the ISO 9790 corridor scaled to this study’s test speed.
Figure 5.2: ISO 9790 impact corridor scaled to 4.5 m/s

Figure 5.3 shows the lateral impact loads overlaying the scaled ISO 9790 corridor and Figure 5.4 shows the oblique impact loads overlaying the scaled corridor.
Figure 5.3: Scaled ISO 9790 corridor with lateral tests overlaid
Most of the data from this study falls well within the corridor as scaled. For much of the time history, the upper bound is well beyond the forces seen in these tests. One important aspect regarding the fit of this data to the corridor is how time zero was established. For all of the data presented in this study, time zero was established based on the last point where the deflection data rose from a value of zero. As can be seen in the deflection data presented in the Results section, all of the tests exhibit a rise in deflection from zero at the beginning of the curve. As such, the data from these tests have had their time shifted, including all of the force data. If the loading that occurs before the chestband registers a deflection were included then all of the force time histories in
Figures 5.3 and 5.4 would be shifted to the right, bringing more of the data within the corridor for each impact type.

5.3 Force vs. Deflection Responses

Figures 5.5 and 5.6 show the force vs. deflection plots that compare the normalized test data to their respective averages and standard deviation corridors. When comparing the average normalized peak forces for each group, there was not much of a difference (lateral average peak was 2557 N and the oblique average peak was 2631 N). There was, however, an observable difference in their average normalized peak deflections (lateral average peak was 70.0 mm and the oblique average peak was 56.1 mm).
Figure 5.5: Normalized force vs. deflection curves for lateral tests
It is possible that the greater deflection observed in the lateral cases could be due to spaces between internal organs in the thorax. In the lateral direction there are more interstitial spaces between organs compared to the oblique direction. When the thorax was impacted laterally, vessels and tissue could be pushed to the side more easily, such as the aorta and IVC being pushed toward the posterior or anterior of the subject. Thus the lungs and heart would be able to move into the space evacuated by the vasculature. In the oblique direction the vasculature would be impinged between the right and left lungs or
the lungs and the posterior wall of the rib cage. Because they could not move out of the line of action of the ram, the oblique impacts would result in a lower deflection. In general, this theory of evacuation or impingement can be seen in the pressure data of Table 4.8 where the oblique tests have higher peak pressure values than the lateral tests.

The corridors defined by the upper and lower bounds of Figures 5.5 and 5.6 are created by taking the standard deviation across both variables (force and deflection) and adding or subtracting that from the average values. Because the standard deviations are plotted in two dimensions there are points where the bounds appear to cross. While there is no error with the data, it is not possible to construct biofidelity corridors for ATDs with these bounds. In order to correct for this one can calculate an ellipse for each data point with the standard deviations for each variable defining the axes of the ellipse. This corridor of error ellipses would then better define how an ATD should respond to impact.
Figure 5.7: Average of lateral and oblique responses, normalized force vs. deflection

It appears that the response of the thorax to impacts at these speeds is not dependent on direction of loading. While this does not follow the findings from the Shaw study, there may be a few explanations for the change. It is possible that the response of the bony thorax, thought to be well represented by the Shaw data observed at lower speeds, is overcome by the viscoelastic response of the internal organs of the thorax. This space is filled with tissue that may be dominating the absorption of energy. It would make sense that when large numbers of rib fractures occur and the bony thorax structure is compromised that the underlying support provided by the internal organs would be
more responsible for the observed thorax response. What might be less intuitive is that even in the tests where serious injury was not caused, the response of the thorax is not significantly different from the seriously injured tests. This supports the theory that impacts at higher speeds are dominated by the viscoelastic response of the subject.

5.4 Injury Criteria

Table 5.2 lists metrics that were used to identify a predictor for AIS3+ level of injury. All of these values are global peak values, without consideration for the time of AIS3+ injury.

<table>
<thead>
<tr>
<th>Test Subject</th>
<th>Age</th>
<th>Normalized Peak Force</th>
<th>Normalized Peak Deflection</th>
<th>Peak Compression</th>
<th>$V_{\text{max}}C_{\text{max}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0802</td>
<td>67</td>
<td>2617 N</td>
<td>70.1 mm</td>
<td>18.5%</td>
<td>0.88</td>
</tr>
<tr>
<td>0803</td>
<td>81</td>
<td>2577 N</td>
<td>56.2 mm</td>
<td>16.7%</td>
<td>0.76</td>
</tr>
<tr>
<td>0804</td>
<td>81</td>
<td>3119 N</td>
<td>59.9 mm</td>
<td>19.8%</td>
<td>0.89</td>
</tr>
<tr>
<td>0901</td>
<td>82</td>
<td>2303 N</td>
<td>53.1 mm</td>
<td>18.3%</td>
<td>0.89</td>
</tr>
<tr>
<td>0902</td>
<td>77</td>
<td>2498 N</td>
<td>69.9 mm</td>
<td>24.7%</td>
<td>1.13</td>
</tr>
<tr>
<td>0903</td>
<td>81</td>
<td>2558 N</td>
<td>70.1 mm</td>
<td>24.7%</td>
<td>1.16</td>
</tr>
<tr>
<td>0906</td>
<td>76</td>
<td>2527 N</td>
<td>55.3 mm</td>
<td>15.9%</td>
<td>0.74</td>
</tr>
</tbody>
</table>

Table 5.2: Metrics used for injury criteria

Risk curves were created using the above tabulated data to determine predictors that may be suitable for predicting serious injury. Binary logistic regressions for each of the predictors were run with the dependent variable being achievement of AIS3+ injury.
Table 5.3 shows the $R^2$ and significance values for each of the predictors, where an $R^2$ value close to one and a significance, or $p$, value less than 0.1 indicate good predictors.

<table>
<thead>
<tr>
<th>Predictor</th>
<th>Nagelkerke $R^2$</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
<td>0.422</td>
<td>0.137</td>
</tr>
<tr>
<td>Normalized Peak Force</td>
<td>0.013</td>
<td>0.796</td>
</tr>
<tr>
<td>Normalized Peak Deflection</td>
<td>0.047</td>
<td>0.618</td>
</tr>
<tr>
<td>Peak Compression</td>
<td>0.787</td>
<td>0.056</td>
</tr>
<tr>
<td>$V_{\text{max}}C_{\text{max}}$</td>
<td>1.000</td>
<td>0.054</td>
</tr>
</tbody>
</table>

**Table 5.3: Summary of goodness of fit for injury risk predictors**

While the $R^2$ value for the $V_{\text{max}}C_{\text{max}}$ predictor appears to be very good, its significance is not commensurate with such a high $R^2$ value. If the sample size were increased then $V_{\text{max}}C_{\text{max}}$ might still be viewed as an acceptable injury predictor with a more realistic goodness of fit. Peak compression appears to also provide a fairly reliable prediction of injury. Again, as sample size is increased a better curve may be created that will more closely match the predictive ability of maximum compression for AIS3+ injury. It should be noted that all of the tests that were a part of this analysis had AIS codes assigned based on rib fractures. If other injuries were found to occur, such as soft tissue injuries, then the ability of maximum compression to predict injury may not be as good as was shown here. Figure 5.8 shows the risk curve based on peak compression and Figure 5.9 shows the risk curve based on the $V_{\text{max}}C_{\text{max}}$ calculation.
Figure 5.8: Risk curve for AIS3+ injury based on peak compression

Figure 5.9: Risk curve for AIS3+ injury based on $V_{\text{max}}C_{\text{max}}$ calculation
5.5 Limitation and Future Work

As is often the case in this field, sample size must be increased in order to have statistically significant data. Basing biofidelity corridors on three or four tests does not provide a sound basis for ATD development. Particularly in this area of pendulum thoracic impacts, more tests should be done at varying speeds to delineate when the directionally dependent response of the thorax disappears. It appears thus far that speed is between 2.5 m/sec and 4.5 m/sec.

Because these tests often caused injury to the subjects, whether serious or not, it is difficult to discern what aspects of the response are affected by the change in structural integrity of the rib cage. While this is certainly a limitation that makes it difficult to compare to non-injured data, such as Shaw et al.’s, this brings up an interesting question: Which data is best to use to represent the response of an ATD to impact at high speeds? While injury does modify the response, an ATD is meant to replicate the response of a human throughout an impact, not just the portion of the impact where injury has not occurred.

Based on the results of this study, the following list is proposed for future work:

1) Test more PMHS using this pendulum setup at varying speeds from 2.5 m/sec to 4.5 m/sec.
2) Analyze the existing pendulum data from this study and previous work collectively using the same zeroing and normalization techniques to identify common responses.

3) Collect more time of injury data and investigate the kinematic thoracic response up to serious injury or first fracture.

4) Continue investigating the relationship between lateral and oblique impacts and the resulting peak thoracic vascular pressures.
Chapter 6: Conclusions

As a result of the testing and analysis of the results presented in this paper, the following conclusions are made:

- Based upon the testing performed there is no dependence on loading direction when the PMHS thorax is loaded at injurious impact speeds

- The average lateral peak impact force was 2557 N and the average oblique peak impact force was 2632 N (normalized responses)

- The average lateral peak deflection was 70.0 mm and the average oblique peak deflection was 56.1 mm (normalized responses)

- The average lateral peak compression value was 22.6% and the average oblique peak deflection value was 17.7%

- Moorhouse’s effective stiffness normalization methodology provided the best grouping of impact response data when compared to Eppinger’s total mass ratio method and Mertz-Viano’s effective mass method
The ISO 9790 response corridor is mostly well formed for high speed impacts, but the upper bound could be refined to better encapsulate the human response to impact.
List of References


Appendix A

Curved Beam Bending Application to Current Study
Deflection of a curved beam can be determined by the means of moment-area theory. Consider the curved beam shown in Figure A.1.

If the beam is loaded by the force at point B, a bending moment will occur. According to moment-area theory the component wise-displacements at a point can be found by

\[
\Delta_x = \int_0^s \frac{M_y}{EI} ds
\]

\[
\Delta_y = \int_0^s \frac{M_x}{EI} ds
\]
and the resultant deflection would be equal to \( \Delta_B = \sqrt{\Delta_x^2 + \Delta_y^2} \) in the direction defined by \( \tan \theta = \frac{\Delta_y}{\Delta_x} \). The location of point B in the x and y directions are

\[
x = R(1 - \cos \theta) \quad \text{and} \quad y = R \sin \theta
\]

The moment and the differential segment are

\[
M = FR \sin \theta \quad \text{and} \quad ds = Rd\theta
\]

Substituting these equalities into the displacement equations above yields

\[
\Delta_{x,B} = \frac{FR^3}{EI} \int_0^{\pi/2} \sin^2 \theta \, d\theta = \frac{\pi FR^3}{4EI}
\]

and

\[
\Delta_{y,B} = \frac{FR^3}{EI} \int_0^{\pi/2} \sin \theta (1 - \cos \theta) \, d\theta = -\frac{FR^3}{2EI}
\]

and the result is

\[
\Delta_B = \sqrt{\left(\frac{\pi FR^3}{4EI}\right)^2 + \left(-\frac{FR^3}{2EI}\right)^2} = \frac{FR^5}{2EI} \sqrt{1 + \frac{\pi^2}{4}}
\]
Appendix B

Sample Transformation Scripts from 3aω Motion Blocks to Subject’s Body Coordinate System
clear all
close all
%-------------------Load Data-------------------
POINTLIST=xlsread ('G:\3aw Transformations\0906\0906_Transformation_Data', 'FARO'); %in units of mm
Accel=xlsread ('G:\3aw Transformations\0906\0906_Transformation_Data', 'Accel'); %in units of G
ARS=xlsread ('G:\3aw Transformations\0906\0906_Transformation_Data', 'ARS'); %in units of deg/s
Time=xlsread ('G:\3aw Transformations\0906\0906_Transformation_Data', 'FARO', 'G1:G60000'); %in units of Sec
ARS= ARS.*pi/180; %convert to units of rad/s

%----------------Determine Global Coordinate System(BCS)-------------
RACR=POINTLIST(1,1:3);
LACR=POINTLIST(2,1:3);
RTROCH=POINTLIST(3,1:3);
LTROCH=POINTLIST(4,1:3);
MIDACR=(RACR+LACR)./2;
MIDTROCH=(RTROCH+LTROCH)./2;
%tip-tail
Ytemp=RACR-LACR;
Ztemp=MIDTROCH-MIDACR;
Xtemp=cross(Ytemp,Ztemp);
Ztemp=cross(Xtemp,Ytemp);
xaxis=Xtemp/norm(Xtemp);
yaxis=Ytemp/norm(Ytemp);
zaxis=Ztemp/norm(Ztemp);

%----------------Correct polarity of transducers-------------------
% Accel(:,1)=-Accel(:,1);
% Accel(:,2)=-Accel(:,2);
% Accel(:,3)=-Accel(:,3);
% Accel(:,4)=-Accel(:,4);
% Accel(:,5)=-Accel(:,5);
% Accel(:,6)=-Accel(:,6);
%
% ARS(:,1)=-ARS(:,1);
% ARS(:,2)=-ARS(:,2);
% ARS(:,3)=-ARS(:,3);
% ARS(:,4)=-ARS(:,4);
% ARS(:,5)=-ARS(:,5);
% ARS(:,6)=-ARS(:,6);

%-----------------Determine initial angles of each block-----------------
[phi_St, theta_St, sigma_St]= Func_Euler_St(POINTLIST(5:7, 1:3), xaxis, yaxis, zaxis);
[phi_T4, theta_T4, sigma_T4]= Func_Euler_T4(POINTLIST(8:10, 1:3), xaxis, yaxis, zaxis);
% Determine the angle Time history for each block and the transformation--
[accel_S1, dts_S1, angle_S1, disp_S1, ang_S1] = Func_trans(phi_S1, theta_S1, sigma_S1, Accel(:,1:3), ARS(:,1:3));

figure (1)
plot(Time, accel_S1(1:60000,1),Time, accel_S1(1:60000,2),Time, accel_S1(1:60000,3),Time,Accel(1:60000,1),Time,Accel(1:60000,2),Time,Accel(1:60000,3));
title('sternum')
axis([-0.01,0.08,-60, 60])
legend('X', 'Y', 'Z', '1', '2', '3')
figure (2)
plot(Time, accel_T4(1:60000,1),Time, accel_T4(1:60000,2),Time, accel_T4(1:60000,3),Time,Accel(1:60000,4),Time,Accel(1:60000,5),Time,Accel(1:60000,6));
title('T4')
axis([-0.01,0.08,-60, 60])
legend('X', 'Y', 'Z', '1', '2', '3')
figure (3)
plot(Time',ang_S1(1:60000,1),Time',ang_S1(1:60000,2),Time',ang_S1(1:60000,3));
title('sternum')
%axis([-0.01,0.08,-60, 60])
legend('X', 'Y', 'Z')
figure (4)
title('T4')
%axis([-0.01,0.08,-60, 60])
legend('X', 'Y', 'Z')
hd1={'Time', 'STRNX_acc', 'STRNY_acc', 'STRNZ_acc', 'T4X_acc', 'T4Y_acc', 'T4Z_acc'; 'Sec' 'G' 'G' 'G' 'G' 'G' 'G'};
hd2={'Time', 'STRNX_ars', 'STRNY_ars', 'STRNZ_ars', 'T4X_ars', 'T4Y_ars', 'T4Z_ars'; 'Sec' 'deg/s' 'deg/s' 'deg/s' 'deg/s' 'deg/s' 'deg/s'};
hd3={'Time', 'STRNX_disp', 'STRNY_disp', 'STRNZ_disp', 'T4X_disp', 'T4Y_disp', 'T4Z_disp'; 'Sec' 'mm' 'mm' 'mm' 'mm' 'mm' 'mm'};
hd4={'Time', 'STRNX_angdisp', 'STRNY_angdisp', 'STRNZ_angdisp', 'T4X_angdisp', 'T4Y_angdisp', 'T4Z_angdisp'; 'Sec' 'deg' 'deg' 'deg' 'deg' 'deg' 'deg'};
xlswrite('0906OTH45L01_Transformed.xls', hd1, 'Acceleration', 'A1');
xlswrite('0906OTH45L01_Transformed.xls', hd2, 'ARS', 'A1');
xlswrite('0906OTH45L01_Transformed.xls', hd3, 'Displacement', 'A1');
xlswrite('0906OTH45L01_Transformed.xls', hd4, 'Angular Displacement', 'A1');
xlswrite('0906OTH45L01_Transformed.xls', [Time, accel_S1, accel_T4], 'Acceleration', 'A3');
xlswrite('0906OTH45L01_Transformed.xls', [Time, dts_S1, dts_T4], 'ARS', 'A3');
% Transformation Matrix

% Transformation Matrix

function A=Func_A(Q)
A = zeros(3,3);
A(1,1)=cos(Q(4))*cos(Q(6))-sin(Q(4))*cos(Q(5))*sin(Q(6));
A(1,2)=-cos(Q(4))*sin(Q(6))-sin(Q(4))*cos(Q(5))*cos(Q(6));
A(1,3)= sin(Q(4))*sin(Q(5));
A(2,1)=sin(Q(4))*cos(Q(6))+cos(Q(4))*cos(Q(5))*sin(Q(6));
A(2,2)=-sin(Q(4))*sin(Q(6))+cos(Q(4))*cos(Q(5))*cos(Q(6));
A(2,3)=-cos(Q(4))*sin(Q(5));
A(3,1)=sin(Q(5))*sin(Q(6));
A(3,2)=sin(Q(5))*cos(Q(6));
A(3,3)=cos(Q(5));

function [Phi_0, Theta_0, Sigma_0] = Func_Euler_St(block_pts, xaxis, yaxis, zaxis)
a=block_pts(2,:)-block_pts(1,:);
j=a/norm(a);
b=block_pts(3,:)-block_pts(2,:);
k=b/norm(b);
i=cross(j,k);
i=i/norm(i);
j=cross(k,i);
j=j/norm(j);

matrix=zeros(3,3);
matrix(1,1)=dot(i,xaxis);
matrix(1,2)=dot(j,xaxis);
matrix(1,3)=dot(k,xaxis);
matrix(2,1)=dot(i,yaxis);
matrix(2,2)=dot(j,yaxis);
matrix(2,3)=dot(k,yaxis);
matrix(3,1)=dot(i,zaxis);
matrix(3,2)=dot(j,zaxis);
matrix(3,3)=dot(k,zaxis);
Ini_A=matrix
Check= matrix*matrix'

% Determine initial Euler angle (3-1-3)
theta_1 = acos(Ini_A(3,3))
Phi_1_1_1 = atan(-Ini_A(1,3)/Ini_A(2,3))
Sigma_3_1_1 = atan(Ini_A(3,1)/Ini_A(3,2))

Phi_1_1_2 = pi+atan(-Ini_A(1,3)/Ini_A(2,3))
Sigma_3_1_2 = pi+atan(Ini_A(3,1)/Ini_A(3,2))

q1 = [0;0;0;Phi_1_1_1;theta_1; Sigma_3_1_1];
q2 = [0;0;0;Phi_1_1_2;theta_1; Sigma_3_1_1];
q3 = [0;0;0;Phi_1_1_1;theta_1; Sigma_3_1_2];
q4 = [0;0;0;Phi_1_1_2;theta_1; Sigma_3_1_2];

% Check the initial Euler angle
K1=Func_A(q1)
K2=Func_A(q2)
K3=Func_A(q3)
K4=Func_A(q4)

er=0.0001;
if (abs(Ini_A(2,1) - K1(2,1)+Ini_A(2,2)-K1(2,2)+Ini_A(2,3)-K1(2,3))<er)
    ini_euler = [Phi_1_1_1, theta_1, Sigma_3_1_1];
else
    if (abs(Ini_A(2,1) - K2(2,1)+Ini_A(2,2)-K2(2,2)+Ini_A(2,3)-K2(2,3))<er)
        ini_euler = [Phi_1_1_2, theta_1, Sigma_3_1_1];
    else
        if (abs(Ini_A(2,1) - K3(2,1)+Ini_A(2,2)-K3(2,2)+Ini_A(2,3)-K3(2,3))<er)
            ini_euler = [Phi_1_1_1, theta_1, Sigma_3_1_2];
        else
            if (abs(Ini_A(2,1) - K4(2,1)+Ini_A(2,2)-K4(2,2)+Ini_A(2,3)-K4(2,3))<er)
                ini_euler = [Phi_1_1_2, theta_1, Sigma_3_1_2];
            else
                ini_euler = [Phi_1_1_2, theta_1, Sigma_3_1_2];
            end
        end
    end
end
end

Ini_Euler = ini_euler;
Phi_0=Ini_Euler(1);
Theta_0=Ini_Euler(2);
Sigma_0=Ini_Euler(3);

function [Phi_0, Theta_0, Sigma_0] = Func_Euler_T4(block_pts, xaxis, yaxis, zaxis)
a=block_pts(3,:)-block_pts(2,:);
j=a/norm(a);
b = block_pts(2,:) - block_pts(1,:);
k = b / norm(b);
i = cross(j, k);
i = i / norm(i);
j = cross(k, i);
j = j / norm(j);

matrix = zeros(3, 3);
matrix(1, 1) = dot(i, xaxis);
matrix(1, 2) = dot(j, xaxis);
matrix(1, 3) = dot(k, xaxis);
matrix(2, 1) = dot(i, yaxis);
matrix(2, 2) = dot(j, yaxis);
matrix(2, 3) = dot(k, yaxis);
matrix(3, 1) = dot(i, zaxis);
matrix(3, 2) = dot(j, zaxis);
matrix(3, 3) = dot(k, zaxis);

Ini_A = matrix
Check = matrix * matrix'

% Determine initial Euler angle (3-1-3)
theta_1 = acos(Ini_A(3,3))
Phi_1_1_1 = atan(-Ini_A(1,3)/Ini_A(2,3))
Sigma_3_1_1 = atan(Ini_A(3,1)/Ini_A(3,2))
Phi_1_1_2 = pi + atan(-Ini_A(1,3)/Ini_A(2,3))
Sigma_3_1_2 = pi + atan(Ini_A(3,1)/Ini_A(3,2))

q1 = [0; 0; 0; Phi_1_1_1; theta_1; Sigma_3_1_1];
q2 = [0; 0; 0; Phi_1_1_2; theta_1; Sigma_3_1_1];
q3 = [0; 0; 0; Phi_1_1_1; theta_1; Sigma_3_1_2];
q4 = [0; 0; 0; Phi_1_1_2; theta_1; Sigma_3_1_2];

% Check the initial Euler angle
K1 = Func_A(q1)
K2 = Func_A(q2)
K3 = Func_A(q3)
K4 = Func_A(q4)
er = 0.0001;
if (abs(Ini_A(2,1) - K1(2,1) + Ini_A(2,2) - K1(2,2) + Ini_A(2,3) - K1(2,3)) < er)
    ini_euler = [Phi_1_1_1, theta_1, Sigma_3_1_1];
else if (abs(Ini_A(2,1) - K2(2,1) + Ini_A(2,2) - K2(2,2) + Ini_A(2,3) - K2(2,3)) < er)
    ini_euler = [Phi_1_1_2, theta_1, Sigma_3_1_1];
else if (abs(Ini_A(2,1) - K3(2,1) + Ini_A(2,2) - K3(2,2) + Ini_A(2,3) - K3(2,3)) < er)
    ini_euler = [Phi_1_1_1, theta_1, Sigma_3_1_2];
else if (abs(Ini_A(2,1) - K4(2,1) + Ini_A(2,2) - K4(2,2) + Ini_A(2,3) - K4(2,3)) < er)

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ini_euler = [Phi_1_1_2, theta_1, Sigma_3_1_2];
end
end
end
Ini_Euler = ini_euler;
Phi_0=Ini_Euler(1);
Theta_0=Ini_Euler(2);
Sigma_0=Ini_Euler(3);

%Given the input of the initial angles as well as the accel and DTS data,
%it will return the transformed accel, DTS data, and displacement
function [accel_xyz, dts_xyz, angle_xyz, disp_xyz, ang_xyz] = Func_trans(phi_0, theta_0, sigma_0, accel, omega)
n=length(accel);
delt=1/20000;
accel=[accel(:,1), accel(:,2), accel(:,3)];

%---------------- Initialize variables ----------------
phi(1)=phi_0;
theta(1)=theta_0;
sigma(1)=sigma_0;
i=1
B=zeros(3,3);
B(1,1)=sin(theta(i))*sin(sigma(i));
B(1,2)=cos(sigma(i));
B(1,3)=0;
B(2,1)=sin(theta(i))*cos(sigma(i));
B(2,2)=-sin(sigma(i));
B(2,3)=0;
B(3,1)=cos(theta(i));
B(3,2)=0;
B(3,3)=1;
omega_x=omega(i, 1);
omega_y=-omega(i, 2);
omega_z=omega(i, 3);
ang(i,1:3)=inv(B)*[omega_x; omega_y; omega_z];
phi_dot(i)=ang(i,1);
theta_dot(i)=ang(i,2);
sigma_dot(i)=ang(i,3);
phi(2)=phi(i)+phi_dot(i)*delt;
theta(2)=theta(i)+theta_dot(i)*delt;
sigma(2)=sigma(i)+sigma_dot(i)*delt;

%---------------- Define time history of phi, theta, and sigma in global
coordinates---------
for i=2:n
B=zeros(3,3);
B(1,1)=sin(theta(i))*sin(sigma(i));
B(1,2)=cos(sigma(i));
\[
\begin{align*}
B(1,3) &= 0; \\
B(2,1) &= \sin(\theta(i)) \cos(\sigma(i)); \\
B(2,2) &= -\sin(\sigma(i)); \\
B(2,3) &= 0; \\
B(3,1) &= \cos(\theta(i)); \\
B(3,2) &= 0; \\
B(3,3) &= 1; \\
\omega_x &= \omega(i, 1); \\
\omega_y &= -\omega(i, 2); \\
\omega_z &= \omega(i, 3); \\
\text{ang}(i, 1:3) &= \text{inv}(B) \cdot [\omega_x; \omega_y; \omega_z]; \\
\phi_{dot}(i) &= \text{ang}(i, 1); \\
\theta_{dot}(i) &= \text{ang}(i, 2); \\
\sigma_{dot}(i) &= \text{ang}(i, 3); \\
\phi(i+1) &= \phi(i) + 0.5 \left(\phi_{dot}(i-1) + \phi_{dot}(i)\right) \cdot \text{delt}; \\
\theta(i+1) &= \theta(i) + 0.5 \left(\theta_{dot}(i-1) + \theta_{dot}(i)\right) \cdot \text{delt}; \\
\sigma(i+1) &= \sigma(i) + 0.5 \left(\sigma_{dot}(i-1) + \sigma_{dot}(i)\right) \cdot \text{delt}; \\
\end{align*}
\]

\[
\text{angle}_xyz = [\phi', \theta', \sigma']; \\
%--------------------------
\text{Transform the data}--------------------------
\text{for}\ i=1:n
\begin{align*}
A &= \text{zeros}(3,3); \\
A(1,1) &= \cos(\phi(i)) \cos(\sigma(i)) - \\
&\quad \sin(\phi(i)) \cos(\theta(i)) \sin(\sigma(i)); \\
A(1,2) &= -\cos(\phi(i)) \sin(\sigma(i)) - \\
&\quad \sin(\phi(i)) \cos(\theta(i)) \cos(\sigma(i)); \\
A(1,3) &= \sin(\phi(i)) \sin(\theta(i)); \\
A(2,1) &= \sin(\phi(i)) \cos(\sigma(i)) + \cos(\phi(i)) \cos(\theta(i)) \sin(\sigma(i)) + \\
&\quad \cos(\phi(i)) \cos(\theta(i)) \sin(\sigma(i)); \\
A(2,2) &= \sin(\phi(i)) \sin(\sigma(i)) + \cos(\phi(i)) \cos(\theta(i)) \cos(\sigma(i)) + \\
&\quad \cos(\phi(i)) \cos(\theta(i)) \cos(\sigma(i)); \\
A(2,3) &= -\cos(\phi(i)) \sin(\theta(i)); \\
A(3,1) &= \sin(\theta(i)) \sin(\sigma(i)); \\
A(3,2) &= \sin(\theta(i)) \cos(\sigma(i)); \\
A(3,3) &= \cos(\theta(i)); \\
\text{accel}_xyz(i, :) &= A \cdot \text{accel}(i, 1:3)'; \\
\end{align*}
\text{end}
\text{for}\ i=1:n
\begin{align*}
A &= \text{zeros}(3,3); \\
A(1,1) &= \cos(\phi(i)) \cos(\sigma(i)) - \\
&\quad \sin(\phi(i)) \cos(\theta(i)) \sin(\sigma(i)); \\
A(1,2) &= -\cos(\phi(i)) \sin(\sigma(i)) - \\
&\quad \sin(\phi(i)) \cos(\theta(i)) \cos(\sigma(i)); \\
A(1,3) &= \sin(\phi(i)) \sin(\theta(i)); \\
A(2,1) &= \sin(\phi(i)) \cos(\sigma(i)) + \cos(\phi(i)) \cos(\theta(i)) \sin(\sigma(i)) + \\
&\quad \cos(\phi(i)) \cos(\theta(i)) \sin(\sigma(i)); \\
A(2,2) &= \sin(\phi(i)) \sin(\sigma(i)) + \cos(\phi(i)) \cos(\theta(i)) \cos(\sigma(i)) + \\
&\quad \cos(\phi(i)) \cos(\theta(i)) \cos(\sigma(i)); \\
A(2,3) &= -\cos(\phi(i)) \sin(\theta(i)); \\
A(3,1) &= \sin(\theta(i)) \sin(\sigma(i)); \\
A(3,2) &= \sin(\theta(i)) \cos(\sigma(i)); \\
A(3,3) &= \cos(\theta(i)); \\
\end{align*}
\text{end}
A(3,3)=cos(theta(i));
dts_xyz(i, :)=A*omega(i, 1:3)'; %units of deg/s
end

disp_xyz=dintegrate(accel_xyz); %units of mm
ang_xyz=180/pi*aintegrate(dts_xyz); %units of deg

function disp= dintegrate(acc)
    ONE=acc(:, 1)';
    TWO=acc(:, 2)';
    THREE=acc(:, 3)';
    delt=1/20000;
    DISPONE=(delt*cumtrapz(delt*cumtrapz(ONE*9.8066)))*1000;
    DISPTWO=(delt*cumtrapz(delt*cumtrapz(TWO*9.8066)))*1000;
    DISPTHREE=(delt*cumtrapz(delt*cumtrapz(THREE*9.8066)))*1000;
    disp=[DISPONE', DISPTWO', DISPTHREE'];
end

function angl= aintegrate(ARS)
    ONE=ARS(:, 1)';
    TWO=ARS(:, 2)';
    THREE=ARS(:, 3)';
    delt=1/20000;
    DISPONE=(delt*cumtrapz(ONE));
    DISPTWO=(delt*cumtrapz(TWO));
    DISPTHREE=(delt*cumtrapz(THREE));
    angl=[DISPONE', DISPTWO', DISPTHREE'];
Appendix C

Sample Processing Scripts for Filtering and Removing Bias from Raw Data
Call FormulaCalc("ch('[1]/Time (Sec)') := ch('[1]/Time')")
ChnDim("[1]/Time (Sec)") = "Sec"
Call CHNDELETE("[1]/Time")
Call CHNMOVE("[1]/Time (Sec)",1,1)

Call FormulaCalc("ch('[1]/Time (mSec)') := ch('[1]/Time (Sec)')*1000")
ChnDim("[1]/Time (mSec)") = "mSec"
Call CHNMOVE("[1]/Time (mSec)",1,2)

'Call FormulaCalc("ch('[1]/RAMXD') := ch('[1]/RAMXD')*-1")
Call CHNOFFSET("[1]/RAMXD","[1]/RAMXD12001",12001,"mean value offset")
Call FormulaCalc("ch('[1]/RAMXD (CM)') := ch('[1]/RAMXD12001')")
Call FormulaCalc("ch('[1]/RAMXD (MM)') := ch('[1]/RAMXD (CM)')*10")
ChnDim("[1]/RAMXD (MM)") = "MM"
Call CHNDELETE("[1]/RAMXD")
Call CHNDELETE("[1]/RAMXD12001")

'Call FormulaCalc("ch('[1]/RAMXG') := ch('[1]/RAMXG')*-1")
Call CHNOFFSET("[1]/RAMXG","[1]/RAMXG12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMXG12001","[1]/RAMXG Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMXG Class 600") = "G"
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMXG12001","[1]/RAMXG Class 180","CFC_180",0,"EndPoints",10)
ChnDim("[1]/RAMXG Class 180") = "G"
Call CHNDELETE("[1]/RAMXG12001")

'Call FormulaCalc("ch('[1]/RDRAMXG') := ch('[1]/RDRAMXG')*-1")
Call CHNOFFSET("[1]/RDRAMXG","[1]/RDRAMXG12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RDRAMXG12001","[1]/RDRAMXG Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RDRAMXG Class 600") = "G"
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RDRAMXG12001","[1]/RDRAMXG Class 180","CFC_180",0,"EndPoints",10)
ChnDim("[1]/RDRAMXG Class 180") = "G"
Call CHNDELETE("[1]/RDRAMXG12001")

Call FormulaCalc("ch('[1]/A') := ch('[1]/RAMXG Class 180')*9.80665")
Call ChnIntegrate("[1]/Time (Sec)","[1]/A","[1]/RAMXG Class 180 Velocity")
ChnDim("[1]/RAMXG Class 180 Velocity") = "M/Sec"
Call ChnIntegrate("[1]/Time (Sec)","[1]/RAMXG Class 180 Velocity","[1]/RAMXG Class 180 Distance (M)")
ChnDim("[1]/RAMXG Class 180 Distance (M)") = "M"
Call FormulaCalc("ch('[1]/RAMXG Class 180 Distance (MM)') := ch('[1]/RAMXG Class 180 Distance (M)')*1000")
ChnDim("[1]/RAMXG Class 180 Distance (MM)") = "MM"
Call ChnDelete("[1]/A")

Call FormulaCalc("ch('[1]/A') := ch('[1]/RDRAMXG Class 180')*9.80665")
Call ChnIntegrate("[1]/Time (Sec)","[1]/A","[1]/RDRAMXG Class 180 Velocity")
ChnDim("[1]/RDRAMXG Class 180 Velocity") = "M/Sec"
Call ChnIntegrate("[1]/Time (Sec)","[1]/RDRAMXG Class 180 Velocity","[1]/RDRAMXG Class 180 Distance (M)")
ChnDim("[1]/RDRAMXG Class 180 Distance (M)") = "M"
Call FormulaCalc("ch('[1]/RDRAMXG Class 180 Distance (MM)'):=ch('[1]/RDRAMXG Class 180 Distance (M)')*1000")
ChnDim("[1]/RDRAMXG Class 180 Distance (MM)") = "MM"
Call ChnDelete("[1]/A")

Call FormulaCalc("ch('[1]/RAMXG Class 600 Force'):= ch('[1]/RAMXG Class 600')*9.80665*22.99")
ChnDim("[1]/RAMXG Class 600 Force") = "N"

Call FormulaCalc("ch('[1]/RAMXG Class 600 Force Compensated'):= ch('[1]/RAMXG Class 600')*9.80665*1.70979")
ChnDim("[1]/RAMXG Class 600 Force Compensated") = "N"

Call FormulaCalc("ch('[1]/RAMXG Class 180 Force'):= ch('[1]/RAMXG Class 180')*9.80665*22.99")
ChnDim("[1]/RAMXG Class 180 Force") = "N"

Call FormulaCalc("ch('[1]/RAMXG Class 180 Force Compensated'):= ch('[1]/RAMXG Class 180')*9.80665*1.70979")
ChnDim("[1]/RAMXG Class 180 Force Compensated") = "N"

Call FormulaCalc("ch('[1]/RDRAMXG Class 600 Force'):= ch('[1]/RDRAMXG Class 600')*9.80665*22.99")
ChnDim("[1]/RDRAMXG Class 600 Force") = "N"

Call FormulaCalc("ch('[1]/RDRAMXG Class 600 Force Compensated'):= ch('[1]/RDRAMXG Class 600')*9.80665*1.70979")
ChnDim("[1]/RDRAMXG Class 600 Force Compensated") = "N"

Call FormulaCalc("ch('[1]/RDRAMXG Class 180 Force'):= ch('[1]/RDRAMXG Class 180')*9.80665*22.99")
ChnDim("[1]/RDRAMXG Class 180 Force") = "N"

Call FormulaCalc("ch('[1]/RDRAMXG Class 180 Force Compensated'):= ch('[1]/RDRAMXG Class 180')*9.80665*1.70979")
ChnDim("[1]/RDRAMXG Class 180 Force Compensated") = "N"

'Call FormulaCalc("ch('[1]/RAMFX'):= ch('[1]/RAMFX')*-1")
Call FormulaCalc("ch('[1]/RAMFX '):= ch('[1]/RAMFX')")
Call CHNOFFSET("[1]/RAMFX ","[1]/RAMFX 12001",12001,"mean value offset")
Call CnCFCFiltCalc("[1]/Time (Sec)"","[1]/RAMFX 12001","[1]/RAMFX Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMFX Class 600") = "N"
Call CHNDELETE("[1]/RAMFX")
Call CHNDELETE("[1]/RAMFX 12001")

'Call FormulaCalc("ch('[1]/RAMFY'):= ch('[1]/RAMFY')*-1")
Call FormulaCalc("ch('[1]/RAMFY '):= ch('[1]/RAMFY')")
Call CHNOFFSET("[1]/RAMFY ","[1]/RAMFY 12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMFY 12001","[1]/RAMFY Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMFY Class 600") = "N"
Call CHNDELETE("[1]/RAMFY")
Call CHNDELETE("[1]/RAMFY 12001")

Call FormulaCalc("ch('[1]/RAMFZ'):= ch('[1]/RAMFZ')*-1")
Call ChnDim("[1]/RAMFZ") = "N"
Call CHNOFFSET("[1]/RAMFZ ","[1]/RAMFZ 12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMFZ 12001","[1]/RAMFZ Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMFZ Class 600") = "N"
Call CHNDELETE("[1]/RAMFZ")
Call CHNDELETE("[1]/RAMFZ 12001")

Call ChnSub("[1]/RAMXG Class 600 Force","[1]/RAMFZ Class 600","[1]/AccelForce-LoadCellForce Class 600")
ChnDim("[1]/AccelForce-LoadCellForce Class 600") = "N"

Call ChnAdd("[1]/RAMXG Class 600 Force","[1]/RAMFZ Class 600","[1]/AccelForce+LoadCellForce Class 600")
ChnDim("[1]/AccelForce+LoadCellForce Class 600") = "N"

Call ChnSub("[1]/RAMFZ Class 600","[1]/RAMXG Class 600 Force Compensated","[1]/LoadCellForce-AccelForceCompensated Class 600")
ChnDim("[1]/LoadCellForce-AccelForceCompensated Class 600") = "N"

Call ChnAdd("[1]/RAMFZ Class 600","[1]/RAMXG Class 600 Force Compensated","[1]/LoadCellForce+AccelForceCompensated Class 600")
ChnDim("[1]/LoadCellForce+AccelForceCompensated Class 600") = "N"

Call ChnSub("[1]/RAMXG Class 180 Force","[1]/RAMFZ Class 180","[1]/AccelForce-LoadCellForce Class 180")
ChnDim("[1]/AccelForce-LoadCellForce Class 180") = "N"

Call ChnAdd("[1]/RAMXG Class 180 Force","[1]/RAMFZ Class 180","[1]/AccelForce+LoadCellForce Class 180")
ChnDim("[1]/AccelForce+LoadCellForce Class 180") = "N"

Call ChnSub("[1]/RAMFZ Class 180","[1]/RAMXG Class 180 Force Compensated","[1]/LoadCellForce-AccelForceCompensated Class 180")
ChnDim("[1]/LoadCellForce-AccelForceCompensated Class 180") = "N"

Call ChnAdd("[1]/RAMFZ Class 180","[1]/RAMXG Class 180 Force Compensated","[1]/LoadCellForce+AccelForceCompensated Class 180")
ChnDim("[1]/LoadCellForce+AccelForceCompensated Class 180") = "N"
Call ChnDifferentiate("[1]/Time (Sec)","[1]/RAMXG Class 180 Force","[1]/Averages RAMXG Class 180 Force","[1]/Rate of change RAMXG Class 180 Force")
ChnDim("[1]/Rate of change RAMXG Class 180 Force")="N/Sec"

Call ChnDifferentiate("[1]/Time (Sec)","[1]/RDRAMXG Class 180 Force","[1]/Averages RDRAMXG Class 180 Force","[1]/Rate of change RDRAMXG Class 180 Force")
ChnDim("[1]/Rate of change RDRAMXG Class 180 Force")="N/Sec"

'Call FormulaCalc("ch('[1]/RAMMX'):= ch('[1]/RAMMX')*-1")
Call FormulaCalc("ch('[1]/RAMMY'):= ch('[1]/RAMMY')")
Call CHNOFFSET("[1]/RAMMX ","[1]/RAMMX 12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMMX 12001","[1]/RAMMX Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMMX Class 600") = "Nm"
Call CHNDELETE("[1]/RAMMX")
Call CHNDELETE("[1]/RAMMX 12001")

'Call FormulaCalc("ch('[1]/RAMMY'):= ch('[1]/RAMMY')*-1")
Call FormulaCalc("ch('[1]/RAMMY'):= ch('[1]/RAMMY')")
Call CHNOFFSET("[1]/RAMMY ","[1]/RAMMY 12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMMY 12001","[1]/RAMMY Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMMY Class 600") = "Nm"
Call CHNDELETE("[1]/RAMMY")
Call CHNDELETE("[1]/RAMMY 12001")

'Call FormulaCalc("ch('[1]/RAMMZ'):= ch('[1]/RAMMZ')*-1")
Call FormulaCalc("ch('[1]/RAMMZ'):= ch('[1]/RAMMZ')")
Call CHNOFFSET("[1]/RAMMZ ","[1]/RAMMZ 12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/RAMMZ 12001","[1]/RAMMZ Class 600","CFC_600",0,"EndPoints",10)
ChnDim("[1]/RAMMZ Class 600") = "Nm"
Call CHNDELETE("[1]/RAMMZ")
Call CHNDELETE("[1]/RAMMZ 12001")

'Call FormulaCalc("ch('[1]/STRNXG'):= ch('[1]/STRNXG')*-1")
Call CHNOFFSET("[1]/STRNXG","[1]/STRNXG12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/STRNXG12001","[1]/STRNXG Class 1000","CFC_1000",0,"EndPoints",10)
ChnDim("[1]/STRNXG Class 1000") = "G"
Call CHNDELETE("[1]/STRNXG12001")

'Call FormulaCalc("ch('[1]/STRNYG'):= ch('[1]/STRNYG')*-1")
Call CHNOFFSET("[1]/STRNYG","[1]/STRNYG12001",12001,"mean value offset")
Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/STRNYG12001","[1]/STRNYG Class 1000","CFC_1000",0,"EndPoints",10)
ChnDim("[1]/STRNYG Class 1000") = "G"
Call CHNDELETE("[1]/STRNYG12001")

'Call FormulaCalc("ch('[1]/STRNZG'):= ch('[1]/STRNZG')*-1")
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Call ChnCFCFiltCalc("[1]/Time (Sec)","[1]/STRNZG12001","[1]/STRNZG Class 1000","CFC_1000",0,"EndPoints",10)
ChnDim("[1]/STRNZG Class 1000") = "G"
Call CHNDELETE("[1]/STRNZG12001")

'Call FormulaCalc("ch('[1]/STRNXdeg'):= ch('[1]/STRNXdeg')*-1")
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ChnDim("[1]/STRNXdeg Class 600 Integrated") = "Deg"
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ChnDim([1]/SPNEZdeg Class 600 Integrated") = "Deg"
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Appendix D

Pre-Test Data Sheet
**Subject Pre Test Data Sheet**

(Justification for Test Subject Study Qualification)

Cadaver Reference Number: __________________________________________________________

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<th>Weight: ______</th>
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Cause of Death: __________________________________________________________

Cadaver Appearance/Anomalies: _______________________________________

Subcutaneous Tissue Depth at each rib (L2-L10 and R2-R10): _______________

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Anthropometry Measurements (mm):

1. **STATUR** (Stature): __________________________
2. **SHLDHT** (Shoulder Height): __________________________
3. **VRTSYM** (Vertex to Symphysion Length): ______________
4. **SHLDBD** (Shoulder Breadth): __________________________
5. **CHSTBD** (Chest Breadth): Axillary:______ Xyphoid:______ Average:______
6. **WASTBD** (Waist Breadth): __________________________
7. **HIPBD** (Hip Breadth): __________________________
8. **HDTROC** (Head to Trochanterion Distance): ______________
9. **SEATHT** (Seated Height): __________________________
10. **INSCYE** (Interscy Distance): __________________________
11. **NECKCR** (Neck Circumference): __________________________
12. **CHSTCR** (Chest Circumference): Axillary:______ Xyphoid:______ Average:______
13. **WASTCR** (Waist Circumference): __________________________
14. **CHSTDPS** (Chest Depth): Axillary:______ Xyphoid:______ Average:______
15. **WASTDP** (Waist Depth): __________________________
Appendix E

Sample Data Acquisition Configuration File
**Config File:**
D:\Thorax\0906OHT45L01\0906OTH45L01.cfg

**Date/Time:**

**Output File:**
D:\Thorax\0906OHT45L01\0906OTH45L01.dx3

**DataBase Path:**
D:\Yokogawa\DataBase\_

**Pre Event:**
1 Sec

**Post Event:**
2 Sec

**Sample Rate:**
20000 Hz

**Description:**
Left Oblique Impact, 4.5 m/s, Speed

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