Comparison of Q3s Anthropomorphic Test Device Biomechanical Responses to Pediatric Volunteers

Thesis

Presented in Partial Fulfillment of the Requirements for the Degree Master of Science in the Graduate School of The Ohio State University

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Abstract

The biofidelity of pediatric ATDs continues to be evaluated with scaled-down adult data, a methodology that makes inaccurate assumptions about the likeness of biomechanical properties of children and adults. Recently, evaluation of pediatric ATDs by comparison of pediatric volunteer (PV) data has been shown to be a valuable and practical alternative to the use of scaled adult data. This study utilized existing PV data to evaluate a 3 year-old side impact ATD, the Q3s. While ATDs have been compared to volunteer responses in frontal impacts, this study is the first to extend ATD-PV comparison methods to the Q3s ATD, and among the first to extend these methods to side impacts.

Previously conducted experiments were replicated in order to make a direct comparison between the Q3s and PVs. PV data were used from 4-7 year-olds (shoulder tests, n=14) and 6-8 year-olds (sled tests, n=7). Force-deflection data were captured during quasi-static shoulder tests through manual displacement of the shoulder joint. Resulting shoulder stiffness was compared between the Q3s and PVs. Low-speed far-side sled tests were conducted with the Q3s at lateral (90°) and oblique (60°) impacts. Primary outcomes of interest included 1) lateral displacement of the torso, 2) torso rollout angle, and 3) kinematic trajectories of the head and neck.
The Q3s exhibited shoulder stiffness values at least 32 N/mm greater than the PVs for all conditions (PV muscles tensed and relaxed, deflection calculated for full- and half-thoracic). In lateral sled tests, the Q3s demonstrated increased lateral torso displacement (Q3s: 194.6 mm; PVs: 164.3 mm ± 26.6), coronal torso rollout (Q3s: 49.2°; PVs: 35.7° ± 12.4), and maximum excursions in the ΔY and ΔZ directions for trajectories of the top of the head, C4, and T1, compared to PVs. In oblique trials, the Q3s did not exhibit significantly different lateral torso displacement or rollout, but achieved less forward ΔX motion (Q3s: 68 mm; PVs: 113 ± 17 mm) compared to PVs.

Increased Q3s ATD shoulder stiffness suggests an effect on loads seen by the Q3s head, torso, and neck during side impact. Q3s and PV trajectories were of similar shape, although Q3s head and neck kinematics displayed rigid body motion followed by independent lateral bending of the head, suggesting cervical and thoracic spine rigidity compared to PVs.
Dedication

Dedicated to my parents, Tim and Cheryl Ita, for their continuous love and support of anything I do; my sister, Rebecca, for being a constant throughout my time at OSU; and my brother, Daniel, for always sending happy, chill thoughts.
I would like to thank my academic committee: Dr. John Bolte, Dr. Mark Ruegsegger, and Dr. Yun-Seok Kang. Further I would like to acknowledge Dr. Bolte for allowing me the opportunity to take on this study in the IBRC as well as serving as a mentor during my time with the lab as both an undergraduate and graduate student. Thank you to Dr. Kang for serving as the P.I. on my thesis study, and being exceptionally helpful and accessible. I have learned so much through my experience working with Dr. Kang and am honored to have been his first student.

I would like to acknowledge the National Science Foundation (NSF) Center for Child Injury Prevention Studies at the Children’s Hospital of Philadelphia (CHOP) and the Ohio State University (OSU) for sponsoring this study and its Industry Advisory Board (IAB) members for their support, valuable input and advice. The views presented are those of the authors and not necessarily the views of CHOP, OSU, the NSF, or the IAB members. Specifically, thank you to my IAB mentor Eric Dahle and Evenflo, Inc.

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tests, and the students and staff at CHOP for their help in conducting tests and sharing data used for comparison.

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Lastly, knowing I have constant support from all of my family and friends is a rock that gets me through the crazy times. Thank you for putting things into perspective when I get too lost in the numbers.
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1. Introduction

The largest contributing factor to causes of death for children up to 14 years of age, with the exception of those children less than 1 year-old, is unintentional injury. Motor vehicle collisions (MVCs) are the leading cause of unintentional injury for children 5-14 years-old, and the second leading cause of unintentional injury for children less than 4 years-old (CDC NCIPC WISQARS 2010). The side impact collision mode has been identified as a priority in the area of child occupant protection, as side impact has been studied very little relative to frontal impact even though higher fatality rates have been observed in side relative to frontal impacts (Arbogast & Durbin, 2013; Starnes & Eigen, 2002; Viano & Parenteau, 2008).

The advance of child restraint system (CRS) design, and implementation of regulatory laws regarding child protection in motor vehicles, has been instrumental in realizing the decline of child MVC-related fatalities and injuries. Until recently, however, the lack of an adequately biofidelic pediatric side impact anthropomorphic test device (ATD) has been a major limitation in the ability to accurately capture and assess ATD behavior and interaction with restraint systems in side impacts. The 3 year-old targeted Q3s ATD, part of the Q-series ATD family, is one of the first side impact specific pediatric ATDs that is emerging as a tool to fill this need.
Since the initial introduction of pediatric ATDs into the industry, biofidelity and robustness have been evaluated using scaled-down adult post-mortem human subject (PMHS) data. Using scaled adult data for pediatric ATD evaluation requires many assumptions about the biomechanical, anthropomorphic, and injury responses of children relative to adults. Numerous studies have reported the effect of age on material and structural properties of the human anatomy, and the implications that age effects have on injury response and mechanism (Burdi et al., 1969; McCray et al., 2007; Starnes & Eigen 2002). While methodologies originally developed to normalize and scale adult data have been modified to attempt to accommodate pediatric ATDs (Irwin & Mertz 1997; Melvin 1995; van Ratingen et al., 1997; Wolanin et al., 1982), these modifications do not adequately account for material differences between children and adults.

Limited studies have been conducted that compare pediatric ATDs and child-size PMHS in crash-like scenarios which demonstrated non-biofidelic responses of pediatric ATDs originally designed to scaled adult specifications (Ash et al., 2009; Lopez-Valdes et al., 2009; Sherwood et al., 2002). Recently, several studies have taken advantage of pediatric volunteer (PV) data to evaluate the biofidelity of ATDs (Seacrist et al., 2010, 2012, 2013, 2014). Pediatric volunteer data is more readily attainable compared to pediatric PMHS, and is emerging as a valuable tool for evaluation of ATD whole body kinematic responses.

The study presented here utilizes the progressive exploration of biomechanical response in pediatric subjects to evaluate the Q3s ATD. While ATDs have been compared to both adult and pediatric volunteer responses for frontal impacts, this study is
the first to extend ATD-PV comparison methods to the Q3s ATD, and among the first to extend these methods to side impacts. Volunteer experiments were replicated with the Q3s, including quasi-static shoulder experiments and low-speed far-side sled tests, to evaluate the accuracy of Q3s biomechanical responses as compared to human pediatric data.
2. Background

2.1. Pediatric Occupants in Side Impacts

The largest contributing factor to causes of death for children up to 14 years of age, with the exception of those children less than 1 year-old, is unintentional injury fatalities (Table 1). The most recent available data, accessed from the Centers for Disease Control and Prevention, deemed motor vehicle collisions (MVCs) as the leading cause of unintentional injury fatalities for children 5-14 years-old, and the second leading cause of unintentional injury fatalities for children less than 4 years-old (CDC NCIPC WISQARS 2010). Additionally, if we consider non-fatal injuries, for every 1 fatality,

<table>
<thead>
<tr>
<th>Table 1: Top five leading causes of death for children up to 14 years-old in the United States - 2007. (replicated from Arbogast &amp; Durbin, 2013; CDC NCIPC WISQARS 2010)</th>
</tr>
</thead>
<tbody>
<tr>
<td>&lt;1 year</td>
</tr>
<tr>
<td>Congenital anomalies (5,785)</td>
</tr>
<tr>
<td>Short gestation (4,857)</td>
</tr>
<tr>
<td>SIDS (2,453)</td>
</tr>
<tr>
<td>Pregnancy complications (1,769)</td>
</tr>
<tr>
<td>Unintentional injury (1,285)</td>
</tr>
</tbody>
</table>
about 18 children are hospitalized and over 400 receive medical treatment for MVC-related injuries (CDC NCIPC WISQARS 2010). Although the scope of pediatric MVC injury risk is low compared to adults, the fact that over 100,000 children under the age of 14 years-old are treated for MVC-related injuries each year necessitates the need for researchers to address automotive safety for the pediatric population (Ridella, 2013).

As the scope of the study presented here considers those children falling around the 3 year-old age group and in side impacts, we can further narrow our population of interest by age and primary direction of force (PDOF). Children around 3 years of age will most commonly be placed in forward facing child restraint systems (FFCRS). The National Highway Traffic Safety Administration (NHTSA) and the American Academy of Pediatrics (AAP) suggest that children remain rear facing until 2 years of age or until the maximum height or weight requirement of the CRS are reached by the child (O’Neil et al., 2011).

In regards to CRS research, the side impact condition has been tested very little relative to frontal impact (Arbogast & Durbin, 2013). While frontal impacts have been more extensively studied, frontal impacts, side impacts, and rollovers have been shown to account for a similar number of child fatalities. For example, for the 1 to 8 year-old age group between 1991 and 2000, there were 2,914 and 2,384 fatalities for the frontal and side impact directions, respectively (Starnes & Eigen, 2002). A review of FARS data from 1996 to 2005 for rear-seated children 0 to 7 years-old reported similar results regarding fatality and crash mode, with 20.3% of fatalities occurring in rollovers, 20.2% in frontal impacts, and 19.6% in side impacts (Viano & Parenteau, 2008). Additionally,
side impacts and rollovers have a higher fatality rate than frontal impacts. Despite similar fatality numbers, both rollover and side impact crash modes were found to have higher fatality risks than frontal impacts, with fatality risks of 1.37%, 0.47%, and 0.34%, for rollover, right-side, and left-side, respectively.

As the side impact crash mode has gained further attention in the automotive safety research community, far-side and near-side impacts have been separately studied as individual crash conditions. Starnes and Eigen (2002) reported the number of fatalities for near-side impacts as 2.6 times greater than the number of fatalities in far-side impacts. Increased fatality risk in near-side impacts has been attributed to direct loading of the CRS/occupant by intruding vehicular structures (Fildes et al., 2003; Franklyn et al., 2007; Klinich et al., 2005; McCray et al., 2007). Thus, some researchers have focused attention on replicating internal vehicular loading conditions of near-side impacts (Langwieder et al., 1997; Paton et al., 1998). Others have focused on injuries induced by far-side impacts, pointing out that these injuries are highly attributable to interaction between occupant and CRS, versus with an intruding vehicle, and therefore more easily targeted via CRS design modifications (Kamren et al., 1993; Klinich et al., 2005). While higher fatality rates have been reported for near-side impacts, serious non-fatal injuries have been reported for far-side impacts as well, where little or no vehicle intrusion was reported. For example, in a study of 93 children ages 0 to 15 years-old in 55 side impacts, 42% of significant injuries occurred with little or no intrusion (Arbogast et al., 2000).
While these numbers highlight the need to test side impacts as thoroughly as frontal impacts, the progress made to reduce MVC-related child fatalities and injuries in the past few years should not be overlooked. For example, number of fatalities in the 1 to 3 year age group dropped 11% from 1996 to 2000 (Starnes & Eigen, 2002). Similarly, the number of fatalities in the 4 to 8 year-old age group dropped 11% from 1998 to 2000, a trend enhanced as the use of CRSs among 3 to 8 year-olds saw a threefold increase between 1999 to 2007 (CHOP, 2008). Implementation of legislation and regulatory laws regarding child protection in motor vehicles is a major contributor to the progressive steps toward improving child safety in motor vehicles. For example, Federal Motor Vehicle Safety Standard (FMVSS) 213: Child Restraint Systems specifies requirements for CRS use in motor vehicles and aircrafts. FMVSS 213 currently only requires that U.S. sold CRSs meet a dynamic testing requirement in the frontal collision direction (Sullivan & Louden, 2009). In November 2013, however, NHTSA published a notice proposing to test CRSs to new side impact performance requirements as an amendment to FMVSS 213 (NHTSA, 2013). The proposed upgrades would require that all CRSs sold in the U.S. demonstrate that they can safely restrain a child by preventing harmful head contact with an intruding vehicle door, and reduce crash forces transmitted to the pediatric occupant’s head and chest. Continuing proactive legislation regarding child safety and CRS requirements is imperative in realizing the decline of child MVC-related fatalities and injuries.

Anthropomorphic test devices (ATDs) are essential in CRS testing to verify the efficacy of CRSs. In order to predict occupant behavior as accurately as possible,
dimensions and characteristics of ATDs are generally designed based on a certain anthropometry (i.e. 50th percentile male, 95th percentile male) and represent a certain age. Therefore, CRS testing requires ATDs that are designed specific to pediatric age groups (i.e. infant, 10 year-old) and to different crash modes (i.e. frontal, side). Until recently, the lack of an adequately biofidelic side-impact ATD has been a major limitation in the ability to accurately capture CRS and ATD interaction and behavior in side impacts. The following section provides a summary of available pediatric ATDs, with a focus on the Q3s ATD that is evaluated in this study.

2.2. Pediatric ATDs

Pediatric ATDs are designed specific to certain age ranges and sometimes to collision mode dependent on PDOF. Some of the first child ATDs were those developed by the Society of Automotive Engineers (SAE) Hybrid III Dummy Family Task Group and the Infant Dummy Task Group (Irwin & Mertz, 1997). These pediatric ATDs were designed to represent the 50th percentile child in the following age groups: 6 months, 12 months, 18 months, 3 years, and 6 years-old. The 6, 12, and 18 month ATDs make up the Child Restraint Air Bag Interaction (CRABI) family of ATDs while the 3 and 6 year ATDs make up part of the Hybrid III ATD family. Note that there is also a Hybrid III 10 year-old ATD, which was designed following the 3 and 6 year-old to represent those older children who could be exposed to an inflating airbag (Humanetics, 2014).

Irwin and Mertz (1997) described the biomechanical bases and biofidelic response requirements of the CRABI and Hybrid III 3YO and 6YO ATDs. The dimensions of the
CRABI ATDs and the Hybrid III 3YO are based off of anthropometry studies from the University of Michigan Transportation Research Institute (Weber et al., 1985), while the Hybrid III 6YO is based off of a different anthropometric study (Young et al., 1976). Methods to determine segment masses for the Hybrid III 3YO and 6YO are also described (Wolanin et al., 1982; Young et al., 1976). While the CRABI ATDs were developed to test in all directions of impact, the Hybrid III 3YO and 6YO were developed to be specific to frontal impacts (Humanetics, 2014). During the development of the CRABI and Hybrid III child ATDs, the P-series and eventual Q-series of child ATDs were being developed in Europe. The Q-series is a more biomechanically advanced family of ATDs compared to the P-series; the Q-series was designed with updated biomechanical and anthropomorphic characteristics, to have multi-directionality, and with interchangeable instrumentation (Humanetics, 2014). Aside from the side-specific Q3s discussed in the following section, the other original Q ATDs were developed for frontal- or multi-directionality, although a Q6s is in development (Humanetics, 2014).

The limitation of the lack of an adequately biofidelic side impact ATD, until recently, has resulted in the use of non-side impact designed ATDs in side impact designed tests (Klinich et al., 2005; van Ratingen & Twisk, 1999). It is worth noting that while some CRS manufacturers internally require CRS side impact tests, these tests have been conducted with ATDs that are not designed specifically for the side impact crash mode, as no side impact specific ATD has been commercially available. This means biofidelity for these ATDs in the lateral direction is not optimized and likely very poor, invalidating the advertising claim that CRSs are side impact certified. In an attempt to
modify an existing ATD to be laterally biofidelic, NHTSA designed the Hybrid III 3Cs, a 3 year-old ATD with a modified neck (NHTSA, 2012). The new neck had the ability to twist while maintaining flexion, extension, and lateral bending motions. The 3Cs ATD was originally evaluated in parallel with the Q3s (described later) as a contender for a side impact ATD standard. In biofidelity studies, the 3Cs performed equivalently to the Q3s in the head and neck regions, but was shown to have poor biofidelity relative to the Q3s in other body regions. Concurrent with the proposal for side impact CRS requirements previously mentioned (NHTSA, 2013), NHTSA also proposed to amend current regulations to add specifications and qualification requirements for the Q3s ATD, which would mean integration of a side impact ATD into CRS side impact tests.

2.2.1. The Q3s ATD

The Q-series of ATDs was developed as a successor to the P-series of ATDs, which were originally developed in the 1970’s and 80’s for use in European standards testing (NHTSA, 2013; van Ratingen et al., 1997). When the development of the Q-series began in 1993 under the direction of the European Commission/International Child Dummy Working Group, the design intention of the Q-series was to provide more instrumentation than the P-series offered and to make these ATD’s multi-directional (NHTSA, 2012). The Q-series covers the complete child population up to 12 years-old. The first ATD designed in the Q-series was the Q3 ATD, which was intended to be multi-directional (van Ratingen et al., 1997).
The Q3 ATD was based off of anthropometry data from CANDAT, a pediatric database developed by TNO compiled of several child anthropometric data sets collected in Europe and North America (Twisk, 1994). The anthropometric angles for the neck and pelvis were for that of a reclined-seated child rather than a straight-seated child to capture the reclined position of a child in a CRS (van Ratingen et al., 1997). A correct kinematic response of the head-neck system of the Q3 was a priority during initial design, as the head is important in injury analyses of pediatric occupants (Brown et al., 2006; McCray et al., 2007; Starnes & Eigen, 2002). Some notable design features of the Q3 included separate joints at the occipital condyles and C1-C2, a flexible column for C2 through C7, a floating joint shoulder resulting in side loading to the ribcage, and a flexible lumbar

![Figure 1: Q3s ATD (Carlson et al., 2007; D. Rhule, 2012).](image)
spine. In addition to describing the biomechanical bases for the Q3, van Ratingen (1997) also presented front and side impact performance targets for the 3 year-old ATD. Details on the development of these targets can be found in section 2.4.2. The multi-directional Q3 ATD was shown to perform poorly in biofidelity analyses in both the front and side impact test directions (Berliner et al., 2000). In 2001, First Technology Safety Systems (FTSS, now Humanetics) began development of the side impact only ATD, the Q3s, shown in Figure 1 (NHTSA, 2013).

Since its initial design, the Q3s has seen much evaluation, redesign, and iteration. Initial biofidelity analyses indicated improvements from ATDs such as the Q3 and Hybrid III 3 year-old that were being used as a side impact surrogate at the time (Carlson et al., 2007). Key deficiencies, however, were also discovered, including lack of a biofidelic neck in the lateral bending mode, an unstable hip joint that would frequently dislodge, and poor durability in the thorax due to cracking (Wang et al., 2009). These deficiencies were addressed by integration of a new neck based off of the neck component designed for the Hybrid III 3Cs, stronger aluminum components to increase stability of the hip joint, and use of a polyurethane material in the thorax that improved fatigue life and optimized stress distribution.

The Q3s as it exists today is 14.5 kg (32.0 lb) and has a seating height representing the 50th percentile 3 year-old at 539 mm (NHTSA, 2013). Despite many iterations, the Q3s is still designed to meet the original specifications defined by van Ratingen (Klinich & Reed, 2013; van Ratingen et al., 1997). The Q3s neck (Figure 2) is segmented with three rubber segments containing V-shaped grooves, bonded to 4
aluminum disks (NHTSA, 2013). A symmetrical safety cable of wire rope runs through the center of the neck to provide axial resistance. The one-piece polyurethane ribcage is bolted to an aluminum spine which contains short interface blocks between the lower neck and thoracic spine and between the lumbar column and pelvis. A similar floating shoulder joint to that of the original Q3 is contained within the Q3s, with the ability to buckle when laterally impacted and covered with a urethane flesh. The Q3s is capable of capturing 30 measurement channels, including displacement measurements in both the shoulder and the thorax. Instrumentation capabilities of the Q3s are outlined in Table 2.

Figure 2: The molded Q3s neck optimized for lateral bending and rotation in molded form (top) and in CAD form highlighting ring and cable assembly (bottom).
In the most recent biofidelity evaluation, NHTSA reported good biofidelity for responses in the lateral and frontal head, lateral thorax, and oblique abdomen, and poor biofidelity for responses in the lateral and oblique shoulder and lateral pelvis (NHTSA, 2012). Although not yet commercially available, the Q3s has been used in research focused on targeting the understanding of various aspects of side impacts. Klinich et al. (2005) used the Q3s with a series of sled tests to analyze CRS and pediatric occupant kinematics in side impacts. In this study, Klinich et al. highlighted how ATD arm position affected ATD kinematics and found that placing the arms at the ATD’s side,
versus angled or extended forward, reduced lateral head excursions. The effect of various methods of securing a CRS in the vehicle on CRS and ATD movement were also studied. NHTSA also used the Q3s for evaluation of initial test procedures of side impact CRS testing (Sullivan & Louden, 2009).

2.3. Biofidelic Evaluation of ATDs

Biofidelity has been defined as a measure of how well an ATD replicates the response of a human (NHTSA, 2012). The biofidelity of ATDs can be evaluated qualitatively and quantitatively. Qualitatively, biofidelity can be assessed by comparison of an ATD’s response to corridors representing a possible range of human subject responses. Quantitatively, methodologies have been established to calculate biofidelity rank values which can aid in the comparison of ATD responses to both human subjects and also other ATDs. This section outlines these methods, while first summarizing the scaling and normalization processes involved in ATD evaluation. An understanding of these processes is important in realizing their limitations when applied to the pediatric population.

2.3.1. Scaling and Normalization

The biofidelity of ATDs is evaluated by comparison between an ATD’s response to a stimulus and human response to an equivalent stimulus. Ideally, ATD biofidelity should be evaluated with those human responses closest to the ATD’s anthropometric
representation. The sizes and ages of available human volunteers and PMHS do not always match those of the anthropometric specifications, and volunteer and PMHS responses are gathered for a range of sizes and ages surrounding the desired anthropometry. In other words, many “non-standard” subject responses are used to predict the response of a “standard” subject. In order to accurately predict this response, however, the relationship between a subject’s physical characteristics and associated response to a stimulus must be accounted for (Mertz, 1984). Inherent human variability and the capturing of human response data of various aged and “non-standard” subjects necessitate the need for a methodology to both normalize and scale subject responses.

Exact meanings behind the terms scaling and normalization are inconsistent throughout the literature, and these processes are often referred to interchangeably (Eppinger et al., 1989; Lopez-Valdes et al., 2012; Mertz, 1984; Moorhouse, 2013). For the purposes of this document, scaling will be defined as the process by which the subject responses designated in one desired anthropometry are adjusted to fit into the range of a different desired anthropometry. For example, if responses from a 50th percentile male population are adjusted to represent those of a 10 year-old child population, the 50th percentile responses would be scaled down to those of a 10 year-old. Normalization will be defined as the adjustment of many non-standard subject responses taken for one standard subject, all of the same desired anthropometry, to that standard subject. Normalization should collapse a group of responses down to one mean response to more accurately and concisely represent one standard population. For example, if the responses of many 6 year-old children are mathematically adjusted according to the 50th
percentile 6 year-old, then these data would have been said to be normalized. Additionally, these normalized curves could be further collapsed into one representative curve for a “standard” 6 year-old. Note that the distinction between scaling and normalization defined here is similar to one defined by NHTSA. NHTSA defined scaling as the changing of values from one reference value to another and normalization as the adjusting of a measured value to a reference (as in adjusting measured test values to an ATD reference). Although defined and referenced as distinctly as possible, scaling and normalization methodologies are often combined in the subsequently explained techniques due to the nature of their development.

The two most commonly utilized scaling/normalization methodologies are those developed by Eppinger, et al., 1989 and Mertz, 1984. These methodologies have been referred to as mass-based scaling and impulse momentum-based scaling, respectively, and are outlined below. Additionally, summaries of mass-based scaling (Eppinger et al., 1989) and impulse momentum-based scaling (Mertz, 1984), also referred to as SAE-scaling in later adjustments to the methodology, can be found in Lopez-Valdes, et al., 2012 and Moorhouse, 2013.

2.3.1.1. Mass-based Methodology

Given two subjects introduced to a comparable dynamic stimulus, mass-based scaling identifies ratios between identical geometric measurements, mass, and time. These fundamental ratios are used to define scaling factors for mass, modulus of elasticity, and density. Mass density and modulus of elasticity are assumed to be constant
such that relationships between velocity, acceleration, and force can also be defined between two subjects (Eppinger et al., 1989). Mass-based scaling can be used to normalize responses within a subject set to a “standard” subject (such as a 50th percentile subject) and can also been implemented in scaling from one target subject anthropometry to another.

Mass-based scaling is advantageous because while velocity is not scaled (a 1:1 ratio), other measured responses such as acceleration, force, and time can be scaled, allowing this methodology to be implemented for a variety of test types. Mass-based scaling is simple to implement relative to other methods and is easily adjustable as it is linked directly to the attainable anthropometry measure of total body mass. Even scale models of identical materials, however, have been shown to be approximate given the assumptions of geometrical and material property similitude (Eppinger et al., 1989). One might then predict that translating this assumption to such a variable and inhomogeneous material as the human body might likewise be problematic. Anthropometric measures of mass and dimension are not necessarily good predictors for stimulus response data as these measures do not account for differences in age, gender, pathology, etc. (Moorhouse, 2013). The underlying assumptions of mass-based scaling become further questionable when scaling between subject age groups.

2.3.1.2. Impulse Momentum-based Methodology

Impulse momentum-based scaling (Mertz, 1984) was developed based on a simple spring-mass system that simulates sled or drop tests and defines effective mass
and stiffness ratios. An “effective mass” is chosen based on the impacted body region, dependent on testing type and conditions. Since the desired effective mass is typically unknown, an average percentage of the effective mass to total body mass is calculated. For the stiffness ratio, Mertz showed that stiffness is proportional to a characteristic length, assuming geometric and material property similitude between subjects in the desired body region. In the initial introduction of impulse momentum-based scaling, Mertz analyzed thorax tests so chose thoracic depth as the characteristic length. Characteristic length, however, would change dependent on test type and conditions. Note that the impulse momentum-based approach was modified by Viano to model a two-mass spring system to accommodate pendulum impacts tests (Viano, 1989).

Impulse momentum-based scaling is advantageous because its use of effective mass and stiffness ratios based on a desired body region can potentially account for variations other than anthropometry within that body region. However, the effective mass to total body mass ratio can vary considerably from subject to subject, as addressed by Mertz (1984). Additionally, there is subjectivity in choosing the most appropriate characteristic length for the test condition. While the assumption of similitude is not being applied to the whole body as in mass-based scaling, it is still being applied to the body region of interest. Thus, validity of using this method both when scaling between target anthropometries and normalizing to a standard subject is questionable.
2.3.1.3. Methodology Modifications

The limitations and short-comings of these methodologies have presented the opportunity to develop improved and modified mass-based and impulse momentum-based scaling techniques. The impulse momentum-based scaling methodology, as presented by Mertz, has been implemented most commonly in specifying response requirements of specific ATDs. In 1984, 1987, and 1989, impulse momentum-based scaling was utilized to produce response requirements for various side impact ATDs (SID, EuroSID-1, and BioSID), the Hybrid III 50\(^{th}\) percentile adult, and for the 95\(^{th}\) percentile male and 5\(^{th}\) percentile female Hybrid III ATDs, respectively (Irwin et al., 2002; Mertz et al., 1989). In these publications, the modulus of elasticity was consistently assumed to be 1. In 1997, however, a modified methodology was presented that incorporated more age-specific scaling factors for the modulus of hard tissue (Irwin & Mertz, 1997). Modified modulus ratios were based off of experimentally obtained values of parietal bone for newborns, 6 year-olds (McPherson & Kriewall, 1980) and adults (Hubbard, 1971). It should be noted that the modulus of soft tissues was still considered to be the same across all ages and subjects. Modification of the hard tissue modulus scaling factor has been implemented into the impulse-momentum based methodology since its introduction (Irwin et al., 2002; Lopez-Valdes et al., 2012; Mertz et al., 2003). Additionally, Lopez-Valdes, et al. have incorporated an age-specific modulus scaling factor into mass-based scaling (Lopez-Valdes et al., 2012).

Moorhouse suggested further modifications to mass-based and impulse momentum-based methodologies to address prominent limitations (Moorhouse, 2013).
To address unconsidered body mass distribution in mass-based scaling, he proposed replacing the ratio of total body mass with a ratio such as Body Mass Index (BMI), $\propto \frac{\text{mass}}{\text{height}^2}$, or Ponderal Index (PI), $\propto \frac{\text{mass}}{\text{height}^3}$. As an improvement to impulse momentum-based scaling, Moorhouse investigated variations and combinations of the characteristic length measurement as well as the calculation of an “effective stiffness.” Similar to the effective mass calculated in the original Mertz method, the effective stiffness calculation is based on deflection data for a specific body region. The effective stiffness measurement is limited in that it can only be calculated if deflection data is directly or indirectly captured during a test, and can also not be directly related back to anthropometric data. Moorhouse used percent coefficient of variation (%CV) to assess how well original mass-based vs. original impulse momentum-based (effective mass and characteristic length) vs. modified impulse momentum-based (effective mass and effective stiffness) performed. The effective mass and effective stiffness methodology performed the best (highest %CV) in 28 out of 38 signal groups analyzed. Moorhouse concluded that the effective mass and effective stiffness scaling process resulted in the best curve grouping and thus a more effective normalization process. It should be considered that the data sets (Maltese et al., 2002; Shaw et al., 2006) used in this analysis were all from roughly the same target population (while age was targeted, sex was not) and thus can be considered to have been normalized versus scaled. While the effective mass-effective stiffness methodology more closely grouped curves in normalizing to a standard subject, one should be careful when translating these conclusions to scaling between target populations.
2.3.1.4. **Original Methodologies**

In addition to methodologies built upon those developed by Eppinger, et al., 1989 and Mertz, 1984, researchers have also developed and implemented original scaling and normalization methodologies. Lopez-Valdes, et al. developed an analytical model based on the hypothesis that differences in spinal motion between pediatric and adult subjects can be explained by segmenting the spine into portions of varying stiffness values (Lopez-Valdes et al., 2010). This model consists of four rigid links connected by rotational joints with five degrees of freedom representing the biomechanics from the head to the pelvis. Another study by Lopez-Valdes, et al. proposed a scaling methodology based on energy considerations (Lopez-Valdes et al., 2012). This method was specific to the test condition considered in the study (frontal sled test impacts with volunteers and PMHS) and assumed that the energy of the sled equaled the work done by the seat belt which restrained the forward motion of the occupant. This method, along with mass-based scaling and a slightly modified impulse momentum-based scaling (described in the previous section), were applied to adult data to predict a known kinematic response for a child. Although all three methods failed to correctly predict the maximum forward displacements for the child, the proposed energy-based methodology was an improvement to the two other methods. Subject data has also been normalized very simply through consideration of some characteristic length specific to test condition. For example, in frontal impact sled tests conducted with adult and child volunteers, kinematic trajectories were normalized by dividing by environment-specific seated height (Arbogast et al., 2009).
2.3.2. Qualitative and Quantitative Biofidelity Assessment

ATDs can be compared to human subjects for their level of biofidelity through qualitative comparison of like response curves. Unfortunately, biofidelity tests for various ATDs have been conducted by various researchers, and as a whole these experiments are inconsistent. Methodologies for the formation of biomechanical targets, or corridors, also vary. Some define the mean human response of a data set of subjects as the mean human response plus or minus one standard deviation, while others define a biomechanical corridor to encompass an entire target subject data set (Rhule et al., 2002). Methodologies for corridor development can also be separated into time-based and time-independent approaches (Lessley et al., 2004). While time-based approaches are quantifiable and repeatable they may not retain the characteristic response curve. Time-independent approaches, on the other hand, generally retain the characteristic response but are inherently subjective and thus rarely repeatable.

For the Biofidelity Ranking System (BRS) presented by Rhule et al. (2002), biomechanical corridors are defined to be the mean human response plus or minus one standard deviation from the mean. It is further defined in a later publication regarding the BRS (Rhule et al., 2013), that use of a single average standard deviation value, rather than a standard deviation value calculated point by point, is advantageous because this can avoid “necking” at points, often caused by the intersection of response curves. This BRS also focuses on what is described as the most relevant part of the curve for the single standard deviation calculation, which is encompassed by the upper 80% of the mean response. A phase optimization methodology to minimize phase difference
between subjects and the ATD is introduced and is described in more detail later in this section. Lessley, et al. (2004) described an approach to corridor development that was both described to be time-independent but also quantifiable, repeatable, and able to maintain the shape of the characteristic response curve. This method could be useful when considering a structural response (force vs. deflection) rather than a time response, and could prevent skewing of corridors when including data set outliers.

The BRS first presented by Rhule et al. (2002), provides a methodology to quantify those qualitative differences seen between ATD and human response curves. The BRS quantifies 1) the ability of an ATD to load a vehicle as a PMHS does, and 2) the ability of an ATD to replicate those PMHS responses that best predict injury potential. These are deemed “External Biofidelity” and “Internal Biofidelity,” and are captured by those signals measuring the response of the test environment, and by those signals measuring the response of the ATD or human, respectively. Biofidelity scores include both an overall score and individual scores for different body regions, which is advantageous in capturing weak and strong regions of an ATD. For each response requirement, cumulative variance of the ATD response relative to the mean cadaver response (DCV) and the cumulative variance of the mean cadaver response relative to the mean plus one standard deviation (CCV) are calculated. A ratio of these values makes up the Response Comparison Value (DCV/CCV), which quantifies how well an ATD duplicates target human subject responses or, in other words, the biofidelity of the ATD.

Updates to the BRS methodology were published in 2009, and included removal of test condition weights originally incorporated, description of a sensitivity analysis to
help understand reported biofidelity of one ATD relative to another, and explanation of how to identify a significant difference between biofidelity ranks of different ATDs (Rhule et al., 2009). In 2013, the BRS was presented in parallel with a methodology for creating biomechanical targets that has both a statistical basis and takes into account the shape and magnitude (SM) and phase (P) of the response curves (Rhule et al., 2013). The phase optimization methodology presented involves locating the mean curve with respect to time zero so that appropriate timing of the response is not lost, which is shown to preserve the integrity of the shape of the response curves over a non-phase optimized mean curve. The BRS is also updated to account for phase differences of ATD responses through calculation of a Shape and Magnitude Response Comparison Value (SM), as well as a Phase Response Comparison Value (P). These values are combined to provide a sense of the total biofidelity quality of each channel time history.

2.4. Differences between Children and Adults

While many studies have captured the injury responses of adult subjects using crash-imitating test conditions and PMHS, the available pediatric injury response data for crash-like conditions are scarce. The insufficiency of pediatric data can be attributed to lack of pediatric PMHS data in addition to ethical concerns over the use of PMHS pediatric data when it is available. Since the initial introduction of pediatric ATDs into the industry, pediatric ATDs have been evaluated at least in part by scaled-down adult data. The use of scaled adult data requires many assumptions about the biomechanical, anthropomorphic, and injury responses of children relative to adults. This section
addresses both 1) the biomechanical differences between adults and children that bring to question the validity of historical scaling methods and 2) old and new practices of assessing the biofidelity of pediatric ATDs.

2.4.1. Biomechanical Differences

Numerous studies have reported the effect of age on material and structural properties of the human anatomy, a few of which are outlined here. The proportion of different body regions relative to total stature varies with growth, with the most notable body region being the head (Burdi et al., 1969). For infants, the head has been reported to be about one-fourth of total stature while for mature adults the head makes up about one-seventh of total stature. Large relative head size for children is an underlying reason why traumatic brain injuries (TBI) are the most common MVC-related serious injury sustained by children (McCray et al., 2007; Starnes & Eigen, 2002). Sitting height has also been shown to vary with age, as the sitting height relative to total height changes from 70%, to 57%, to 52% for infants, 3 year-olds, and adults, respectively (Burdi et al., 1969). This variation contributes to a changing center of gravity with age, which can certainly factor into injury response in MVC. Much attention has been paid to spinal differences between children and adults as well, as discussed in detail by Arbogast, et al. (2009, 2012). It has been reported that full ossification of the spine is not accomplished until 30 years of age, and that the rate of intervertebral disc growth is additionally influenced by location in the spinal column (Franklyn et al., 2007; Kokoska., 2001; Platzer et al., 2007). Incomplete bone ossification and thus cartilage prevalence has also
explained higher shoulder stiffness reported for the 50th percentile male compared to children (Suntay et al., 2011).

Studies more specifically focused on kinematic behavior in sub-injurious MVC-simulated experiments have also captured response differences with age. In low-speed frontal sled tests, forward head excursion and head rotation were both shown to decrease with age, and the magnitude of spinal flexion was reported highest in youngest subjects (Arbogast et al., 2009). This study targeted head to neck girth ratio, which decreased with increasing age, as the primary factor governing these kinematic differences. Kinematics were also captured for children and adults in the same low-speed sled test apparatus, but with lateral and oblique impacts (Arbogast et al., 2012). In these tests, children demonstrated reduced lateral movement but increased suprasternal notch displacement compared to adults.

Differences in reported MVC-related injuries between children and adults also speak to biomechanical differences between the two groups. For example, in thoracic injuries sustained in side impacts, lung contusions are primarily reported in CRS-aged children while skeletal rib fractures are more common in adults (McCray et al., 2007). Younger children are also more susceptible to upper cervical spine injuries and often exhibit “spinal cord injury without radiographic abnormalities” (SCIWORA), due to the compliance of the pediatric spinal column (Platzer et al., 2007). Alternatively, lower cervical spine injuries are more common in older children and adults. Pelvis fractures are also prevalent in adult occupants, but not in child occupants, which is explained by increased joint elasticity in children (Arbogast et al., 2002).
2.4.2. Assessing Pediatric ATD Biofidelity

Since their development, pediatric ATDs have been assessed for their level of biofidelity and for the robustness of their original design specifications. Methodologies originally developed to normalize and scale adult data to evaluate adult ATDs have been modified to attempt to accommodate pediatric ATDs. The modification to the modulus scaling factor presented by Irwin and Mertz (1997), as discussed previously, was originally utilized in an attempt to capture hard-tissue modulus differences between adults and children. In this publication, established requirements for a 50\textsuperscript{th} percentile male were scaled to form response requirements that served as the bases for development of the CRABI (6, 12, and 18 month) and Hybrid III (3YO and 6YO) pediatric ATDs. In addition to an age specific hard-tissue modulus, the pediatric ATD requirements were also modified based on a relationship for acceleration proposed by Melvin (1995) and a scaling factor for the ATD necks proposed by Wolanin et al. (1982). Melvin’s relationship defined an acceleration factor based on knowledge regarding pediatric skull and tendon properties, which set the acceleration scaling factor equal to the skull bending modulus over the characteristic head length for the CRABI ATDs. Wolanin et al. set the product of the neck circumference ratios in the x, y, and z directions equal to the scaling factor for mass. This mass scaling factor was applied to both the Hybrid III and CRABI ATDs. In defining functional design targets for the Q3 ATD, van Ratingen et al. (1997) also implemented impulse momentum-based scaling with Wolanin’s adjusted mass scaling factor for the neck response. Additional adjustments to injury assessment
reference values (IARVs) to account for the differences between adult and child can be found in Irwin and Mertz, 2003.

Although attempts have been made to modify scaling factors to accommodate the pediatric population, these scaling factors still do not adequately account for differences between children and adults. There have been limited studies conducted which compare pediatric ATDs and child-size PMHS in crash-like scenarios. Sherwood et al. (2002) compared the response of the Hybrid III 6YO to a 12 year-old PMHS in frontal high-speed impacts and highlighted the poor biofidelity of the ATD’s thoracic spine. When compared to the PMHS, the 6YO ATD exhibited high neck forces and moments that were unrepresentative of true injury potential. Lopez-Valdes et al. (2009) also conducted high-speed frontal tests with the Hybrid III 6YO and an adult PMHS described as “child-size.” In this study, the ATD predicted accurate peak values but exhibited differences from the PMHS in torso angle and head CG and mid-spine resultant accelerations. The biofidelity of the Hybrid III 10 year-old and Hybrid III 5th percentile female were evaluated with a 13 year-old pediatric PMHS by Ash, et al. (2009). Observed differences between the ATDs and the cadaver were also attributed in part to the non-biofidelic motion of the rigid thoracic spine of the ATDs.

Recently, many studies have taken advantage of pediatric volunteer data to evaluate the biofidelity of ATDs. Pediatric volunteer data is experimentally attainable compared to pediatric PMHS, and is emerging as a valuable tool for evaluation of ATD whole body kinematic responses. Arbogast et al. (2009) captured the kinematic responses of 20 child volunteers in sub-injurious frontal impacts, which has served as a
robust dataset for ATD comparison. When compared to pediatric volunteers, the Hybrid III 6YO ATD exhibited decreased spinal excursions and increased reaction loads (Seacrist et al., 2010). This study was extended to the Hybrid III 10YO and the Q-series 6YO and 10YO ATDs in 2012, as the first study to compare whole body kinematics of the Hybrid III and Q-series pediatric ATDs to living children in a simulated frontal impact (Seacrist et al., 2012). Results indicated that both 6YO ATDs exhibited significantly greater belt reaction loads compared to volunteers, and 10YO ATDs exhibited similar reaction loads but had significantly reduced head rotation compared to volunteers. All ATDs showed significant reductions in neck excursion, a response difference which is hypothesized to again be due to the overly rigid ATD thoracic spine. Arbogast et al. (2012) also captured kinematic responses of 20 pediatric subjects in sub-injurious lateral and oblique impacts, which, until now, have not yet been used for ATD comparison.

2.5. Summary

The development of protection principles for child occupants in side impacts has been defined as a future priority in the area of child occupant protection (Arbogast & Durbin, 2013). This is especially significant given the recent implementation of side impact performance requirements into CRS regulations (NHTSA, 2013). The proposal of the Q3s ATD as the side impact ATD necessary to conduct side impact regulation tests indicates NHTSA’s overall satisfaction with the biofidelity of the Q3s ATD. To evaluate Q3s ATD biofidelity, NHTSA utilized values published in different data sets as ATD
biomechanical targets (Bolte et al., 2003; Elhagediab & Moss, 2001; Irwin et al., 2002; Irwin & Mertz, 1997). While these sources are commonly implemented in ATD assessment, all of these targets are based off of scaled adult data, primarily the 50th percentile adult male. NHTSA addressed this concern in the 2013 notice of proposed rulemaking, stating, “Given the lack of pediatric biomechanical data and the many assumptions made in the scaling process, there is greater uncertainty associated with child biofidelity targets compared to the adult targets from which they are derived. Therefore, NHTSA does not consider the biofidelity targets applied herein to be strict prerequisites to accept the dummy” (NHTSA, 2013). While utilization of scaled adult criteria may indeed be the most efficient ATD evaluation tool to move forward and avoid delays of advancements in pediatric occupant safety regulations, a myriad of studies have demonstrated underlying scaling assumptions to be invalid, and the inaccuracy of responses of pediatric ATDs designed based off of scaled adult targets.

The study presented here utilizes the progressive exploration of the understanding of injury mechanism and causation in pediatric subjects to evaluate the Q3s ATD. The insufficiency of pediatric PMHS data has been and will continue to be an on going limitation of pediatric ATD design and evaluation. Comparison of ATD response to pediatric volunteer response is emerging as an accessible and valuable alternative to the use of scaled adult criteria. While ATDs have been compared to volunteer responses in frontal impacts, this study is the first to extend ATD - pediatric volunteer comparison methods to the Q3s ATD and one of the first to extend these methods to side impacts. The following methods sections describe the volunteer experiments replicated with the
Q3s, including quasi-static shoulder experiments and low-speed far-side sled tests. Through volunteer comparison, we can evaluate the accuracy of Q3s biomechanical responses with human pediatric data and achieve insight into the nature of a child’s interaction with the vehicle and restraint components during a typical side impact.
3. Methods

Previously conducted experiments were replicated in order to make a direct comparison between the Q3s and pediatric volunteers (PVs). The two experiments, quasi-static shoulder experiments and low-speed far-side sled tests, are explained in the following two sections. Detailed methods as described in the original publications can be found in Suntay et al. (2011) and Arbogast et al. (2012).

3.1. Quasi-static Shoulder Experiments

3.1.1. Experimental Setup and Instrumentation

Force and displacement measurements were captured using a custom force applicator with an attached load cell, and a motion analysis system. The custom force applicator contained a load cell on its end (Honeywell Model 31 Mid-Range Precision Miniature). The frame on which the force applicator was mounted was designed to allow for movement up and down so as to align the force applicator with the subject’s shoulder joint (Figure 3). The base of the force applicator additionally allowed for rotational movement such that the applicator could be positioned for a medial loading direction ($0^\circ$) or a posteromedial loading direction ($30^\circ$). Three additional load cells (Honeywell
Model 31 Mid-Range Precision Miniature) were mounted to a wall that was positioned on the non-loading side of the Q3s. This wall was positioned flush with the hip joint up to the shoulder joint of the subject to prevent lateral excursion and “tilt.” It should be noted that the far-side wall was not incorporated into the experimental setup until after the study published by Suntay et al. (2011).

![Figure 3](image)

**Figure 3:** Custom force applicator used to manually displace the shoulder joint. Top view shows positioning for medial and posteromedial loading directions (replicated from Suntay et al., 2011).

Displacement measurements were captured using an 8-camera, 100 Hz motion analysis system (Vicon Motion Systems, Oxford, UK). The global coordinate system (with respect to the motion analysis system) was defined as positive x in front of the occupant, positive z above the occupant, and positive y to the left of the occupant. Reflective markers were placed on the Q3s right and left acromion, the sternum, on either side of the structure directly superior to the sternum, the force applicator load cell, and on
the far-side load cell wall, as seen in Figure 4. Deflection measurements were also acquired from the linear potentiometer internal to the Q3s. The linear potentiometer was configured to capture deflection measurements between the impacted shoulder joint and the spine, deep to the sternum marker. While linear potentiometer data were not used in

![Figure 4: Reflective markers for displacement capture on the Q3s and test environment.](image)

the Q3s to PV comparison, measurements from the linear potentiometer served as a validation of the Vicon motion capture displacement measurements in the medial loading direction.
The Q3s was positioned on the testing bench such that the ATD spine was aligned with the edge of the bench. The force applicator frame was positioned so that the load cell was in line with the ATD shoulder joint (Figure 5). Each trial consisted of 10 seconds of shoulder displacement through manual force application to the shoulder joint. Three tests were performed on the Q3s right shoulder for both the medial and posteromedial (30° from medial) loading directions.

3.1.2. Data Reduction and Analysis

Motion capture data were acquired at 100 Hz and processed using Vicon Nexus software. Force and linear potentiometer data were captured at 1,000 Hz and filtered with a low-pass butterworth filter at 100 Hz. Data were processed using a custom MATLAB script. This script, and pseudocode for this script (Figure 22) can be found in
Appendix A. Shoulder deflection was calculated from both the acromion to sternum (half-thoracic) and from the right acromion to the left acromion (full-thoracic). Data were truncated at the point where subject tilt, defined as the angle formed between the line made by the markers located on the sternum and the left acromion, and the horizontal, exceeded 4° from initial position (Bolte et al., 2003; Suntay et al., 2011). Data truncation due to subject tilt was rarely observed, as the far-side wall was purposed to prevent this.

Since shoulder stiffness values were calculated as the slope of the resulting force-deflection (F-D) curves, a viably linear portion of the force-deflection curve had to be defined. To accomplish this, the middle 20-80% of the curves was defined based off of the peak force measurement. An exemplar curve from a posteromedial trial, with the middle 20-80% bolded, is shown in Figure 6. In some cases, the middle 20-80% of the

![Figure 6: Example F-D curves post-processing. 20-80% of peak force (bolded portion) captured a relatively linear area in a.), but the boundaries were modified in b.) to 10-70% to avoid the horizontal portion of the curve.](image-url)
curve did not eliminate negative slopes or otherwise very nonlinear portions of the curve. In these cases, a linear portion of the curve was manually defined by adjusting the 20% and 80% boundaries to eliminate extreme nonlinearities. An exemplar curve that was manually adjusted to define the linear portion as 10% to 70% of peak force is also shown in Figure 6. The force-deflection curves for all trials displaying 0-100% of peak force and the corresponding linear portion can be found in Figures 23-37 in Appendix B. Note that during this processing stage, PV force-deflection data were omitted from the remaining analyses if the output force-deflection curves either 1) displayed exclusively negative deflection or 2) were deemed to contain no viably linear portion.

In order to plot Q3s shoulder response against pediatric volunteer shoulder response, individual F-D curves were created for each subject for each test condition. To accomplish this, repeated subject trials within each test condition group were interpolated onto common values of deflection (Suntay et al., 2011). Interpolation was limited by the response of shortest deflection within a trial group, so trials less than 50% of the maximum deflection within the trial group were omitted from the interpolation. Further, if subject mean curves exhibited a deflection less than 2 mm, they were also omitted from the interpolation calculation for the PV group mean curve, for reasons explained below.

During initial experiments conducted by Suntay, et al. (2011), less than 2 mm deflection was consistently observed in the volunteers in the medial loading direction. This was described as the amount of deflection needed to deform only the skin layer (0.5 mm to 4 mm for eyelids and hand/feet soles, respectively). Additionally, the resolution of the Vicon motion capture system was on the order of magnitude of 0.1 mm, which
Suntay explained introduced a 5% error into the deflection measurements. Therefore, only the posteromedial data were presented previously for the volunteers, and are the only data compared here between the pediatric volunteers and Q3s.

3.2. Low-Speed Far-Side Sled Tests

3.2.1. Experimental Setup and Instrumentation

Low-speed far-side sled tests were conducted with a pneumatically actuated - hydraulically controlled low-speed crash sled (Figure 7) located at the Applied Biomechanics Lab at Rowan University. The sled consisted of three sub-assemblies: the frame, actuator, and seating buck. Namely, the sled frame was primarily constructed of extruded aluminum tubing (MiniTec Framing Systems LLC, Victor, NY) and a 2-way high dynamics proportional throttle cartridge valve (Model LIQZO-LE, Atos, Italy) was

![Figure 7: Low-speed far-side sled test used for lateral and oblique impacts.](image-url)
used in the hydraulic control circuit. More details of the sled construction can be found in Arbogast, et al. (2009).

The sled was designed such that it could be rotated to allow for sled pulses applied to various directions relative to the sled occupant. In this study, sled pulses were applied to mimic a purely lateral (90°) and an oblique impact (60°). The design for the sled acceleration pulse was based off of an amusement park bumper car ride (Six Flags Great Adventure, Jackson, NJ) which provided a basis for safety considerations (Arbogast et al., 2009). For volunteers included in this study, the average sled pulse acceleration and rise time were 1.8 g, 59.9 ms and 1.9 g, 53.9 ms for lateral and oblique trials, respectively. For trials conducted with the Q3s, the average sled pulse acceleration and rise time were 1.9 g, 59.1 ms and 2.0 g, 55.7 ms for lateral and oblique trials, respectively.

Reaction forces were acquired from various sled and seat instrumentation channels, including a sled accelerometer (Model 7264-200, Endevco, San Juan, CA), and lightweight belt webbing load cells on the shoulder belt (located approximately five inches away from the D-ring between the surrogate and the D-ring) and left and right lap belts (Model 6200FL-41-30, Denton ATD Inc, Rochester Hills, MI). Single six-axis load cells were also placed under the seat pan (Model IF-217, FTSS, Plymouth, MI) and under the foot rest (Model IF-234, FTSS, Plymouth, MI) (Arbogast et al., 2009). Q3s instrumentation data were acquired for the channels listed in Table 3. For tests conducted with volunteers, subjects wore an instrumented headband equipped with three accelerometers (Model 7264B-500, Endevco, San Juan, CA) and three angular rate
sensors (ARS) (ARS-1500, DTS Inc, Seal Beach, CA) to capture signals in the x, y, and z-directions, which can be seen on the image of a volunteer in Figure 8. Reflective markers were placed on the Q3s in the orientation previously used for pediatric subjects and tracked using a 3D motion analysis system (Model Eagle 4, Motion Analysis Corporation, Santa Rosa, CA). This included markers on the head, torso, spine, and extremities. Reflective markers were also placed on the seat pan and sled. A comprehensive list of reflective marker location can be found in Arbogast et al. (2009), Table A1.

The Q3s was positioned as described in detail by Arbogast et al. (2012). An image of a 6 year-old volunteer and the Q3s seated in the sled at initial position is shown in Figure 8. Q3s positioning included setting the initial torso and knee flexion angles to 110° through adjustments of the footrest position. The shoulder and lap belt angles, defined as the angle made with the horizontal, were set to 70° and adjusted to

<table>
<thead>
<tr>
<th>Sensor Type</th>
<th>Location in ATD</th>
<th>Measurements</th>
<th>Total # Channels</th>
</tr>
</thead>
<tbody>
<tr>
<td>Accelerometers</td>
<td>Head C.G.</td>
<td>Ax, Ay, Az</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>Thorax (T1, top and side of the spine box near neck base)</td>
<td>Ay</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Thorax (upper spine box)</td>
<td>Ax, Ay, Az</td>
<td>3</td>
</tr>
<tr>
<td>Load Cells</td>
<td>Upper Neck</td>
<td>Fx, Fy, Fz, Mx, My, Mz</td>
<td>6</td>
</tr>
<tr>
<td>Angular Rate Sensors (ARS)</td>
<td>Head</td>
<td>ωx, ωy, ωz</td>
<td>3</td>
</tr>
</tbody>
</table>
accommodate the decreased seated height of the Q3s by adding a lower backrest bar. Surrogates were restrained with an automotive three-point seatbelt (Takata Corp., Tokyo, Japan) with an emergency locking retractor (locking threshold of 2 g, pretensioning load of approximately 300 N). The test matrix found in Table 4 was used for both the PVs analyzed in this study and for the Q3s. For this study, only EMSR OFF (no retractor) and D-Ring Forward conditions were analyzed.

### 3.2.2. Data Reduction and Analysis

Motion capture data were acquired at 100 Hz and analyzed using Cortex 2.6 software (Motion Analysis, Inc.). On-board accelerometer and load cell data, as well as Q3s instrumentation channels, were captured at 10,000 Hz with a built-in anti-aliasing
filter at 4,300 Hz. Sled and Q3s instrumentation signals were filtered according to SAE J211 standards at SAE CFC60. ARS signals were filtered at CFC1000.

Motion capture data were processed using a custom MATLAB script. For motion capture data, a reflective marker at the right rear of the seat pan was designated as the origin for the local sled coordinate system. The coordinate system was defined as positive x in front of the occupant, positive z above the occupant, and positive y to the left of the occupant. Primary outcomes of interest for this study included: 1) lateral displacement of the torso, 2) torso rollout angle projected onto the coronal and transverse planes, and 3) trajectories over time for markers on the top of the head (HT), C4, and T1. Lateral displacement of the torso was calculated using movement of the suprasternal notch (SSN) marker (Arbogast et al., 2012). Torso rollout angle, also defined by Arbogast et al., was the projected angle made between the line connecting the SSN and xiphoid process and the line connecting the first and third markers on the shoulder belt (Figure 9). Maximum excursions were calculated for marker trajectories of interest as the change from initial position to maximum excursion. Time at maximum was also recorded for all outcomes.

**Table 4:** Sled test matrix used with PVs and the Q3s. Shaded cells indicate test data analyzed for this study.

<table>
<thead>
<tr>
<th>Loading Direction</th>
<th>D-Ring Forward</th>
<th>D-Ring Rearward</th>
</tr>
</thead>
<tbody>
<tr>
<td>90°</td>
<td>EMSR ON</td>
<td>EMSR ON</td>
</tr>
<tr>
<td></td>
<td><strong>EMSR OFF</strong></td>
<td>EMSR OFF</td>
</tr>
<tr>
<td>60°</td>
<td>EMSR ON</td>
<td>EMSR ON</td>
</tr>
<tr>
<td></td>
<td><strong>EMSR OFF</strong></td>
<td>EMSR OFF</td>
</tr>
</tbody>
</table>
Average trajectories, excursions, SSN lateral displacement, and torso rollout angles were calculated for lateral and oblique trials for the PV group and compared to the Q3s. Peak and time at peak were calculated and averaged for the PV group and a 95% confidence interval (CI) for the PVs was calculated for each outcome using JMP software (JMP 10 SAS, Cary, NC). Differences between the Q3s and PVs were assessed by comparing the Q3s mean value to the corresponding PV 95% CI.

Figure 9: Representation of torso rollout angle (replicated from Arbogast, et al., 2012).

3.2.3. Length Scaling

In order to compare the kinematics of the PV group to the Q3s, PV kinematic data were length scaled based on dimensional analysis (Ash et al., 2009; Irwin et al., 2002). Kinematic trajectories and torso displacement were scaled using a scaling factor ($\lambda_L$)
defined as the ratio between the seated height of the Q3s \(L_{Q3s}\) and the seated height of the PV \(L_{PV}\), as shown in Eq. (1).

\[
\lambda_L = \frac{L_{Q3s}}{L_{PV}}
\]  

(1)

Scaling was deemed necessary because although the smallest age-group of available PV data were used (6-8 years-old), the PVs were not size-matched to the Q3s. As only kinematic data were compared in the current study, no other scaling factors were required.

### 3.3. Pediatric Volunteers

Pediatric volunteer data for shoulder stiffness experiments were used from 14

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Sex</th>
<th>Age (years)</th>
<th>Mass (kg)</th>
<th>Seated Height (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>15</td>
<td>F</td>
<td>6</td>
<td>18.6</td>
<td>63.9</td>
</tr>
<tr>
<td>20</td>
<td>F</td>
<td>5</td>
<td>19.5</td>
<td>61.0</td>
</tr>
<tr>
<td>24</td>
<td>F</td>
<td>4</td>
<td>21.3</td>
<td>60.5</td>
</tr>
<tr>
<td>25</td>
<td>M</td>
<td>6</td>
<td>20.9</td>
<td>63.5</td>
</tr>
<tr>
<td>26</td>
<td>F</td>
<td>6</td>
<td>24.9</td>
<td>62.0</td>
</tr>
<tr>
<td>27</td>
<td>M</td>
<td>4</td>
<td>20.9</td>
<td>57.0</td>
</tr>
<tr>
<td>29</td>
<td>M</td>
<td>6</td>
<td>23.6</td>
<td>60.5</td>
</tr>
<tr>
<td>30</td>
<td>F</td>
<td>4</td>
<td>18.8</td>
<td>54.5</td>
</tr>
<tr>
<td>33</td>
<td>M</td>
<td>7</td>
<td>24.0</td>
<td>68.0</td>
</tr>
<tr>
<td>34</td>
<td>M</td>
<td>6</td>
<td>19.5</td>
<td>64.0</td>
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<tr>
<td>40</td>
<td>M</td>
<td>5</td>
<td>20.4</td>
<td>66.0</td>
</tr>
<tr>
<td>43</td>
<td>F</td>
<td>6</td>
<td>25.4</td>
<td>68.5</td>
</tr>
<tr>
<td>44</td>
<td>F</td>
<td>7</td>
<td>23.1</td>
<td>67.0</td>
</tr>
<tr>
<td>45</td>
<td>M</td>
<td>5</td>
<td>20.0</td>
<td>65.0</td>
</tr>
<tr>
<td>Q3s</td>
<td>-</td>
<td>3</td>
<td>14.5</td>
<td>56.9</td>
</tr>
</tbody>
</table>
subjects ages 4 to 7 years-old (4YO: n=3, 5YO: n=3; 6YO: n=6; 7YO: n=2) (Suntay et al., 2011). PV subject information can be found in Table 5. Force-displacement data were collected in both the medial and posteromedial (30° from medial) loading directions and in both the muscle tensed and muscle relaxed condition for every volunteer.

Pediatric volunteer data for the low-speed far-side sled tests were used from 7 subjects ages 6 to 8 years-old (6YO: n=2, 7YO: n=3; 8YO: n=2) (Seacrist et al., 2014). Sled tests were conducted for each subject in either the lateral or oblique condition, never both. PV subject information can be found in Table 6. Q3s information is also included. It should be noted that the listed Q3s seated height was measured manually by the same method as was used for the PVs, and deviates from the nominal seated height specification by 3 cm.

### Table 6: PV and Q3s subject and test information for sled tests.

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Test Condition deg</th>
<th>Sex</th>
<th>Age years</th>
<th>Mass kg</th>
<th>Seated Height cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>PV Lat 1</td>
<td>90</td>
<td>M</td>
<td>8</td>
<td>36.7</td>
<td>73.5</td>
</tr>
<tr>
<td>PV Lat 2</td>
<td>90</td>
<td>M</td>
<td>6</td>
<td>28.8</td>
<td>68.5</td>
</tr>
<tr>
<td>PV Lat 3</td>
<td>90</td>
<td>M</td>
<td>7</td>
<td>32.7</td>
<td>67.0</td>
</tr>
<tr>
<td>PV Lat 4</td>
<td>90</td>
<td>M</td>
<td>7</td>
<td>29.7</td>
<td>69.8</td>
</tr>
<tr>
<td>PV Obl 1</td>
<td>60</td>
<td>M</td>
<td>8</td>
<td>28.1</td>
<td>68.5</td>
</tr>
<tr>
<td>PV Obl 2</td>
<td>60</td>
<td>M</td>
<td>7</td>
<td>27.4</td>
<td>64.8</td>
</tr>
<tr>
<td>PV Obl 3</td>
<td>60</td>
<td>M</td>
<td>6</td>
<td>20.2</td>
<td>66.0</td>
</tr>
<tr>
<td>Q3s</td>
<td>90, 60</td>
<td>M</td>
<td>3</td>
<td>14.5</td>
<td>56.9</td>
</tr>
</tbody>
</table>
4. Results

Results for both the quasi-static shoulder experiments (section 4.1) and the sled test experiments (section 4.2) are presented in the subsequent sections.

4.1. Quasi-static Shoulder Experiments

Data acquired from the Q3s linear potentiometer were plotted against the half-thoracic deflection calculated from motion capture data for three medial loading trials conducted with the Q3s (Figure 10). The motion capture peak deflection percent difference from the potentiometer peak deflection was calculated for each trial to quantify the difference in deflection measurements between the two methodologies. Percent differences were found to be -7.14%, +8.78%, and +16.32% for the three trials, respectively.

Stiffness was calculated for each individual trial per subject by applying a linear fit to the linear portion of the F-D curve (Figures 23-37 in Appendix B). Individual stiffness values can be found in Table 10 in Appendix A. Interpolated subject mean curves can be found in Appendix C, Figures 38-51. An exemplar interpolated subject mean curve is shown for the Q3s in Figure 11 for both full and half-thoracic deflections.
F-D curves for both the Q3s and the PV group, shown in Figure 12, were plotted
according to both 1) deflection calculation (full-thoracic or half-thoracic) and 2) PV test condition (muscles relaxed or tensed). Stiffness values with mean and standard deviation, as listed in Figure 12, were calculated using the average and standard deviation from individual stiffness values found in Table 10 (Appendix B). The calculated Q3s shoulder stiffness values were much higher than any of the stiffness values (relaxed and tensed) exhibited by the PVs.

**Figure 12**: F-D curves plotted for the Q3s and PV group for the posteromedial loading direction. Each PV curve represents the mean response from one pediatric subject. Stiffness values (K) are listed as mean ± one standard deviation in N/mm.
During initial data analysis, subject mean curves were calculated in two different ways: 1) through point-by-point averaging (time history method) and 2) interpolation onto common points of deflection as previously explained in 3.1.2 (interpolation method). The interpolation method was used for the current study, as the manually applied loading rate for each quasi-static load was not controlled, and therefore a mean could not be calculated by a method that assumes consistent time histories relative to a common event, as is the case in the time history method (Lessley et al., 2004; Rhule et al., 2002). However, subject mean curves were calculated using both methods and are compared in Table 7.

**Table 7:** Shoulder stiffness calculated for the PV group using both 1) time history and interpolation methods and 2) slope of mean curve and avg. subject curves methods.

<table>
<thead>
<tr>
<th></th>
<th>Relaxed Half</th>
<th>Relaxed Full</th>
<th>Tensed Half</th>
<th>Tensed Full</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Time History</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stiffness (N/mm) (slope of mean curve)</td>
<td>2.69</td>
<td>1.97</td>
<td>2.82</td>
<td>2.42</td>
</tr>
<tr>
<td>Stiffness (N/mm) (avg. subject curves)</td>
<td>2.64</td>
<td>1.97</td>
<td>3.01</td>
<td>2.51</td>
</tr>
<tr>
<td><strong>Interpolation</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stiffness (N/mm) (slope of mean curve)</td>
<td>2.52</td>
<td>1.85</td>
<td>2.78</td>
<td>2.41</td>
</tr>
<tr>
<td>Stiffness (N/mm) (avg. subject curves)</td>
<td>2.52</td>
<td>1.85</td>
<td>2.78</td>
<td>2.41</td>
</tr>
<tr>
<td><strong>Percent Difference (Time History as base)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>% difference (slope of mean curve)</td>
<td>6.83</td>
<td>6.24</td>
<td>1.52</td>
<td>0.56</td>
</tr>
<tr>
<td>% difference (avg. subject curves)</td>
<td>4.79</td>
<td>6.38</td>
<td>8.22</td>
<td>4.16</td>
</tr>
</tbody>
</table>
Further, a mean stiffness value for a group of subjects can be calculated by either 1) applying a linear fit to the overall mean curve (slope of mean curve method) or 2) averaging the stiffness values found for individual subject mean curves (avg. subject curves method). In the current study, individual trial stiffness values (Table 10) were averaged to capture the full variability (standard deviation) in the whole PV data set. To investigate the effects of both different subject mean curve methods and stiffness calculation methods, shoulder stiffness values were calculated for the PV group and are summarized in Table 7. The interpolation stiffness percent difference from the time history stiffness was calculated for each test condition to quantify difference between the two subject mean curve creation methods.

4.2. Low-Speed Far-Side Sled Tests

Plots presented show one mean curve per subject. If data from one trial were lost (n=2 for each condition), the remaining trial was plotted individually. Note that the legend for Figure 13 and following sled test plots indicates age of PV per curve, but like lines across lateral and oblique trials do not indicate the same subject.

The sled pulse for low-speed far-side sled tests from event onset can be seen in Figure 13. Individual trial responses can be found in Appendix D, Figure 52. Event onset was defined as the time at which the sled acceleration reached 5% of its peak value (Arbogast et al., 2012). Pulses were very similar between the PVs and Q3s, with acceleration and rise time differing by less than .1 g and 2 ms, respectively.
Comparison of torso displacement, as represented by lateral displacement of the
torso. Q3s plotted against PV mean curves for lateral and oblique
trials.

Figure 13: Sled pulse from event onset. Q3s plotted against PV mean curves for lateral and oblique
trials.

Figure 14: Lateral displacement (y-direction) of the SSN marker. Q3s plotted against PV mean curves for lateral and oblique trials.

Comparison of torso displacement, as represented by lateral displacement of the
suprasternal notch (SSN) marker, can be seen between the Q3s and PVs in Figure 14. Individual trial responses can be found in Appendix D, Figure 53. Maximum SSN displacement values and time at maximum for the Q3s and PVs, as well as a 95% CI for the PVs, are located in Table 8. The Q3s exhibited significantly delayed (Q3s peaks not within 95% CI) times to maximum torso displacement for both lateral and oblique trials. While torso displacement was larger for the Q3s than the PVs in lateral trials, torso displacement was not significantly different (Q3s peaks did lie within 95% CI) for oblique trials.

![Figure 15: Torso rollout angle projected onto the transverse (x-y) plane. Q3s plotted against PV mean curves for lateral and oblique trials.](image)

Torso rollout angle was projected onto both the transverse (x-y) (Figure 15) and coronal (y-z) (Figure 16) planes. Individual trial responses for transverse and coronal
rollout angles can be found in Appendix D, Figures 54 and 55, respectively. Maximum rollout angles and times at maximum for the Q3s and PVs, as well as a 95% CI for the PVs, are located in Table 8. In the lateral trials, the Q3s exhibited a similar maximum but delayed rollout as compared to the PVs in the transverse plane. In the coronal plane, the Q3s exhibited a similar time to maximum but greater rollout angle. In the oblique trials, the Q3s did not exhibit differences from the PVs in terms of maximum or time at maximum in either plane projection. It should be noted that for the oblique trials, marker acquisition was lost in one subject such that rollout angles could only be calculated for two of the three PVs tested in the oblique condition. Thus, 95% CI were calculated with a small n, resulting in rather large intervals. Sample size should be considered along with statistical significance.

**Figure 16:** Torso rollout angle projected onto the coronal (y-z) plane. Q3s plotted against PV mean curves for lateral and oblique trials.
Trajectories of the head top (HT), C4, and T1 markers were plotted in the coronal (y-z) plane for lateral and oblique trials in Figure 17 and 18, respectively. Trajectories are shown 1) relative to the local sled origin and 2) relative to the T1 trajectory. Maximum excursion from initial position and time at maximum for the Q3s and PVs, as well as a 95% CI for the PVs, are located in Table 9. In some cases, maximum excursion of a marker (largest deviation from initial position at event onset) was not consistently in the negative or position direction across PVs. This sometimes resulted in a PV mean value around 0 with a large standard deviation. Specifically, inconsistency of excursion polarity was observed in the C4 and T1 markers in the z-direction for both lateral and oblique trials. These markers are still included in Table 9, however, as the maximum values of the Q3s in comparison to the PVs demonstrated kinematic differences of the head and neck.

**Table 8:** Maximum and time at maximum for SSN Y-displacement and torso rollout angles. Shaded cells indicate a significant difference between the Q3s and PVs at significance level of .05.

<table>
<thead>
<tr>
<th></th>
<th>Q3s (Mean)</th>
<th>PV (Mean±SD)</th>
<th>PV 95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Lateral</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SSN Y-displacement</td>
<td>Max (mm)</td>
<td>194.6</td>
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<td>Time (msec)</td>
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<tr>
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<td>166.2 ± 24.5</td>
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<td>31.8 ± 1.7</td>
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<td>37.9 ± 10.8</td>
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<td>39.5 ± 2.3</td>
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<td>Time (msec)</td>
<td>38.5</td>
<td>34.3 ± 4.9</td>
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To further investigate the relative movement between the HT, C4, and T1
landmarks, relative displacement was calculated in the y-z plane between HT and C4, and between HT and T1. Relative displacement was zeroed from initial distance between markers in order to isolate the relative displacement (movement) between landmarks.

Relative displacements were calculated for lateral (Figure 19) and oblique (Figure 20).

Table 9: Maximum excursion and time at maximum for $\Delta Y$ and $\Delta Z$ trajectories. Shaded cells indicate a significant difference between the Q3s and PVs at significance level of .05.

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<th></th>
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<th>HT</th>
<th>Max (mm)</th>
<th>$\Delta Z$</th>
<th>Time (msec)</th>
<th>C4</th>
<th>Max (mm)</th>
<th>$\Delta Y$</th>
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<th>Max (mm)</th>
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<td>PV (Mean±SD)</td>
<td>PV 95% CI</td>
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<td>25±2</td>
<td>23 - 28</td>
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</tr>
</tbody>
</table>

57
trials. Trajectories of HT, C4, and T1 markers were also plotted onto the sagittal (x-z) and transverse (x-y) planes in Figure 56 and Figure 57, respectively, in Appendix D. Maximum excursion from initial position and time at maximum for the Q3s and PVs, as well as a 95% CI for the PVs, are located in Table 11 for ∆X.

Subject plots for non-normalized responses are located in Appendix E. This includes curves for lateral torso displacement (Figures 58 and 59), as well as for all head and neck trajectories (Figures 60-63).

Figure 19: Displacement over time between HT-T1 and C4-T1 in the y-z plane. Initial displacement between markers has been set to zero. Q3s plotted against PV mean curves for lateral trials.
Figure 20: Displacement over time between HT-T1 and C4-T1 in the y-z plane. Initial displacement between markers has been set to zero. Q3s plotted against PV mean curves for oblique trials.
5. Discussion

This study aimed to evaluate biomechanical responses of the Q3s ATD against the youngest PV data available in side impact loading conditions. Stiffness was calculated for the Q3s and PVs in order to evaluate structural properties of the Q3s shoulder. Low-speed far-side sled test results highlighted differences in kinematic movement during lateral and oblique loadings.

5.1. Pediatric Comparison Group

In choosing a PV group for Q3s comparison, the 4-8 year-old group (4-7 year-old shoulder tests, 6-8 year-old sled tests) was chosen because it was the youngest data set available and therefore closest to the target age of the Q3s. While maturation changes existing between 3 year-old and 4-8 year-old children should not be ignored in Q3s-PV comparisons, it should be noted that maturation changes between these age groups will be less drastic than those observed between a 3 year-old and 18 year-old, or between a 3 year-old and adult PMHS, and so on. It is believed that a comparison to younger PVs will help mitigate implications of the assumptions between ATD target age and comparison group often problematic in scaling methodologies. Despite an age gap, 4-8 year-old PVs provide a robust data set for Q3s evaluation. For example, cervical spine
range of motion (ROM) measurements were compared between children 3-5 years-old and 6-8 years-old (Arbogast et al., 2007). ROM measures that did not show a difference with age included right and left lateral bending. Studies have also identified a transition in cervical spine injury mechanism in children around the age of 8 or 9 years-old (Platzer et al., 2007). Comparing the Q3s to PVs below this range becomes important when linking pediatric injury risk and Q3s metrics found during testing. These findings partially help to validate conclusions drawn from side impact kinematics between the target-aged 3 year-old Q3s and the 4-8 year-old PVs.

5.2. Quasi-static Shoulder Experiments

5.2.1. Motion Capture Validation

The comparison between the acquired data for the linear potentiometer and the motion capture, shown in Figure 11, indicates similar deflection response between the two methodologies. The overall shape of the curves is qualitatively similar across all three trials. While phase differences between the potentiometer and motion capture curves are not qualitatively distinguishable, the percent differences of peak deflection present concern. Interestingly, motion capture did not consistently over or under estimate deflection relative to potentiometer deflection, making it difficult to relate differences to the experimental setup. Reflective markers were not removed in between trials, although it is possible that markers experienced some shifting between trials.
While these differences do not affect results of the current study (Q3s-PV comparison), differences could have implications on calculated shoulder stiffness for the volunteers as used in other applications. These findings should be taken into consideration with the understanding of the small sample size (n=3). Additionally, potentiometer deflections are considered here as the “true” deflections, as the potentiometer detects deflection between the rigid Q3s spine and acromion. This is appropriate if stiffness of the rigid shoulder complex (acromion to thorax) is desired, but a potentiometer measurement would not be an appropriate comparison metric for a stiffness calculation incorporating surrounding soft tissue in a human volunteer. Comparison of potentiometer data to motion capture provides for a means to validate the quasi-static shoulder testing that has not previously been available. The methodology developed to measure shoulder stiffness of volunteers has previously been concluded as repeatable (Suntay et al., 2011), but comparison to potentiometer data in ATD replicated experiments could provide validation for the magnitude of calculated volunteer shoulder stiffness.

5.2.2. Shoulder Stiffness

The increased shoulder stiffness of the Q3s relative to the PVs is consistent with results of dynamic testing of the Q3s shoulder conducted by NHTSA, where the Q3s shoulder exhibited high shoulder stiffness against lateral and oblique corridors (Bolte et al., 2003; Irwin et al., 2002; NHTSA 2013). While the argument exists that a more compliant shoulder joint would compromise durability of the Q3s as a testing tool, a
A high-stiffness shoulder could have implications on the response of the thorax and head. In side impacts, the shoulder of a pediatric occupant often interacts first with an intruding vehicular structure or restraint system (Suntay et al., 2011). Shoulder deflection due to medial loading results in initial impact load distribution to the spinal column and head, so a high-stiffness shoulder complex may affect loads seen by the head and thorax (Thollon et al., 2001). This is extremely relevant to pediatric occupants, as children in side impacts compared to frontal are significantly more likely to suffer severe injuries to the head and thorax (Brown et al., 2006). The effect of high shoulder stiffness on ATD torso, neck, and head response should be further investigated.

5.2.3. Calculation of the Subject Mean Curve

Discussion regarding different methods to calculate the subject mean curve for quasi-static data is included as an aside to analyzing these data in future quasi-static experiments conducted with human volunteers, as, again, outcomes of the Q3s-PVs comparison of shoulder stiffness will be the same regardless of method. These variations on methods could provide for an interesting discussion on how best to analyze quasi-static F-D data for stiffness. While the principle of using the interpolation method over the time history method is certainly valid, the percent difference between the two methods in less than 10% in all cases analyzed (Table 7). Interestingly, the time history stiffness calculation exceeded the interpolation calculation in all instances. This information provides some insight into whether or not interpolation is necessary, which is limited in these experiments by the trial with the smallest measured deflection and is a
complicated calculation compared to the time history method. Additionally, there were relative differences within the time history calculated stiffness between the two methods for finding one overall mean shoulder stiffness (slope of mean curve v. avg. subject curves). It should be noted that if experimental design is not balanced (same amount of trials per subject being analyzed), then simply averaging stiffness values for all trials (as was done in the current study) is flawed in that subjects’ contribution to the overall mean are unevenly weighted dependent on how many trials with viable data were captured. Even if the stiffness values presented here indicate that method does not make a difference to calculated stiffness for these experiments, the theoretical mathematical principles behind these two methods should be reviewed, as manipulation of the data should ultimately depend on if a calculation is mathematically appropriate.

5.3. **Low-Speed Far-Side Sled Tests**

5.3.1. **Kinematic Differences**

In the lateral trials, the combined increased torso displacement and coronal plane trajectories demonstrated that the Q3s experienced an overall total body lateral excursion greater than the PVs. Increased torso rollout in the coronal plane, along with increased excursions of the HT-C4-T1 complex, are additionally consistent with increased lateral bending observed in the Q3s. Especially large ΔY and ΔZ were observed for the Q3s HT marker. A qualitative analysis of the lateral trajectories in Figure 17 indicates rigid body motion of the Q3s between the HT, C4, and T1. Toward the end of the trajectory,
however, the Q3s HT marker appears to have achieved greater lateral bending than the C4 and T1 markers. Rigid body motion followed by increased lateral bending of the head (resembling a “head snap” motion) was consistent with high speed videos of the Q3s. In contrast, the PVs did not exhibit this rigid body motion to the same extent as the Q3s, as near the end of the trajectory the C4 and T1 markers deviated upwards from initial position, toward HT.

It is hypothesized that lack of relative movement between HT, C4, and T1 (prior to head snap), could be due to increased rigidity of the Q3s cervical spine relative to PVs. Analysis of the T1 marker also provides some insight into thoracic spine rigidity differences between the Q3s and PVs. The Q3s T1 followed the general shape of the Q3s HT throughout the trajectory and maintained downward motion, while the PVs T1 stayed in line with, or moved slightly above, initial position. To further investigate this, HT and C4 trajectories were plotted relative to T1 (Figure 17). The majority of the HT trajectory relative to T1 can likely be attributed to the Q3s head snap. Trajectories relative to T1 highlight similar motion between T1 and C4 for Q3s and PVs. A look at C4 and HT trajectories also confirms negative relative displacement (compression between the two markers) between C4 and HT for PVs that is not exhibited by the Q3s. Similar findings were observed in the oblique trials with trajectories plotted relative to T1 (Figure 18).

To further investigate relative movement between the head and neck landmarks, relative displacement in the y-z plane was plotted over time for the head and neck markers (Figure 19 and Figure 20). In lateral trials, Q3s HT-T1 and C4-T1 relative displacement did not exceed 5 mm in either tension or compression. One volunteer,
However, experienced over 10 mm of relative movement in compression. In oblique trials, Q3s HT-T1 relative displacement did not exceed 10 mm in compression, while PV HT-T1 relative displacement reached up to 30 mm in compression. These data suggest that while there may be non-biofidelic components of the Q3s thoracic spine, non-biofidelic movement is also occurring superior to T1.

Spine rigidity of both the cervical and thoracic spines likely contributed to observed kinematic differences. Kinematic differences between pediatric ATDs and PVs (Seacrist et al., 2010, 2012) or PMHS (Ash et al., 2009; Sherwood et al., 2002) have previously been attributed to non-biofidelic motion of the ATD thoracic spine. Little data, however, is available in terms of ATD-PV comparison in side impact or with side impact ATDs. Conclusions drawn regarding an overly rigid cervical and/or thoracic spine in the Q3s should be considered with the understanding that kinematic differences may not hold true for higher severity impacts.

While the most relevant plane for side impact is the y-z plane, x-y plane trajectories also demonstrated differences, especially in the oblique trials. For the oblique trials, both PVs and Q3s achieved HT ΔX in flexion (Q3s: 68 mm; PVs: 113 ± 17 mm). The Q3s HT ΔX, however, also exhibited a comparable excursion in extension (-60 mm), whereas the PVs did not achieve any motion rearward of initial position. It should be noted that ΔX excursions (Table 11, Appendix D) only list maximum excursions in either the positive or negative direction (whichever had a greater magnitude). Because the Q3s had a maximum excursion in flexion in one trial and in extension in the second, these values average close to 0. Interestingly, significantly decreased HT ΔX excursions have
been demonstrated in other pediatric ATDs in frontal impacts, specifically the Hybrid III 6 year-old and 10 year-old, Q6, and Q10 ATDs (Seacrist et al., 2012). While side impact ATDs in oblique impacts, and frontal impact ATDs in frontal impacts cannot be directly compared, underestimation of head $\Delta X$ excursion could be a design flaw consistent across pediatric ATDs.

It is possible that lack of muscle response in the ATD contributed to observed kinematic differences. Differences associated with lack of muscle responses in pediatric ATDs relative to PVs have been previously discussed as a factor contributing to lack of rebound in frontal low-speed sled tests (Seacrist et al., 2010, 2012). Similar to findings with other pediatric ATDs, the Q3s did not exhibit the same mechanistic rebound as seen with PVs in either lateral or oblique impacts.

5.4. Limitations

Several limitations of this study need to be addressed. First, there are sources of error associated with the motion capture systems used in both quasi-static shoulder and sled tests. The error as measured for the shoulder and sled test motion capture systems has been estimated as (up to) 5% (Suntay et al., 2011) and 0.3% (Arbogast et al., 2009), respectively, in the different setups. Additionally, displacements and deflections as measured with the motion capture markers assume that the markers exactly match up with the skeletal structures that they represent. At low speeds, Arbogast estimated this error to be less than 2% by assessing two markers on the same skeletal body (2012).
Exact replication of the low-speed far-side sled tests was necessary in order to make a direct comparison between the Q3s and PVs. A limitation exists here, however, in that the setup utilized a three-point seat belt, while a 3 year-old occupant would (ideally) be constrained to a CRS. Additionally, the Q-series ATDs have more cylindrical ribcages and rounded abdomens compared to Hybrid III pediatric ATD counterparts that cause suboptimal fit of the seat belt across the thorax (Seacrist et al., 2012). This leads to seat belt slip up to the neck (also seen in Figure 8) which was also observed in Q3s testing. It is possible that the effect of poor seat belt fit on kinematics could be exacerbated in the Q3s due to decreased seated height. These limitations should be considered along with differences observed between the Q3s and PVs. While the sled test environment was not age specific to the Q3s, it is valuable in that it provided a dataset of unconstrained movement ideal for ATD-PV comparison. Sled modification for a CRS should be considered for future work, although such a modification would also depend on the practicality of recruitment of younger CRS-aged subjects.

While differences in shoulder stiffness between the Q3s and PVs were distinct, and thus a larger sample size would likely not identify further insights into Q3s-PV differences, kinematic differences between Q3s and PVs in sled tests were not as clear. A larger sample size for the 6-8 year-old age group, especially younger subjects, would strengthen conclusions drawn from kinematic comparison to the Q3s. The sample size presented here (n=7), however, is comparable to other PV-ATD studies in the literature.
5.5. Implications and Future Work

In the design and construction of ATDs, there must be a compromise between durability and compliance to optimize ATD biofidelity and efficacy of the ATD as a testing tool. This is certainly true with the shoulder joint of the Q3s. While it may be unrealistic to ever design a Q3s shoulder joint to have a stiffness in the range of the volunteers, the effect of increased shoulder stiffness on the kinematics of the head and the neck should be further investigated.

While only kinematic responses from the low-speed far-side sled tests are presented here, environment reaction forces and Q3s instrumentation data were acquired as a part of these tests. These data should be further analyzed to identify similarities and/or differences between the Q3s and PVs. Head acceleration and upper neck forces and moments are an area of future work and will be especially useful in identifying if differences observed in Q3s neck and head kinematics have a significant effect on forces experienced by the Q3s head-neck junction.

Evaluating the biofidelity of pediatric ATDs with pediatric volunteer data is valuable because 1) volunteer data is attainable relative to pediatric PMHS and 2) assumptions required when scaling adult data are mitigated. The increasing importance for biofidelic ATDs at low speeds has been highlighted as an area of concern as evolving restraint systems limit the crash loads and accelerations transmitted to the occupant (Seacrist et al., 2010, 2012). There are obvious limitations, however, in that volunteer data is obtained at sub-injurious levels not indicative of crash scenarios, and the relationship between behavior at low impacts and behavior at high impacts is largely
unknown. However, the collection of pediatric data in different conditions over the past years has provided robust data sets for sub-injurious comparisons. While ATD-PV comparison at matched conditions is certainly valid and expected, there is opportunity to translate these data to use at higher severities. As indicated schematically in Figure 21, from these data sets biofidelity at low severities can be quantified for a pediatric ATD. If some level of low severity biofidelity can be compared with ATD response at higher severities, perhaps these data could be used to create a transfer function to predict either 1) biofidelity rating for the ATD in crash-like conditions or 2) pediatric response in crash-like conditions. By combining available data (green cells in Figure 21), there is the opportunity for insight into pediatric subject interaction with vehicle and restraint systems in various crash modes. However, care would need to be taken that any transfer

**Figure 21:** Schematic of future work. Green cells indicate those areas where data are available; yellow cells indicate unknown areas.
function creation did not make cyclical assumptions about ATD biofidelity (i.e. ATD response should not be used as a predictor of pediatric response if biofidelity of the ATD in some fashion cannot be affirmed).

Higher severity sled tests have been conducted which could aid in investigating biofidelity of the Q3s. These tests were conducted at low, medium, and high severity levels, with a low severity condition that additionally matched those low-speed far-side sled tests presented in this study. These data should be investigated along with the findings of this study to provide insight into methodologies of pediatric ATD biofidelity evaluation, as well as the behavior of the Q3s at higher severity impacts.
6. Conclusion

Based on the research presented in this document, the following observations were compiled:

- The motion capture methodology for capturing deflection was comparable to linear potentiometer measurements.
- The Q3s exhibited shoulder stiffness values at least 32 N/mm greater than pediatric volunteers for all conditions analyzed.
- The effect of increased shoulder stiffness on ATD head and neck kinematics should be further investigated.
- In lateral sled tests, the Q3s exhibited increased lateral torso displacement and head bending compared to pediatric volunteers.
- In lateral and oblique sled tests, the Q3s exhibited significantly delayed times to peak torso displacement compared to volunteers.
- Overall, Q3s and volunteer trajectories of the head and neck were similar, although the Q3s trajectory mechanism displayed rigid body motion followed by an independent head snap not observed in volunteers.
• Differences in kinematics between the Q3s and volunteers could be attributed to increased rigidity in the cervical and thoracic spine; this should be further investigated.
7. References


Mertz HJ, Irwin AL, Melvin JW, Stalnaker RL, Beebe MS. Size, weight and biomechanical


Ridella SA. Pediatric Injury Biomechanics - Foreward. In: Crandall JR, Myers BS, Meaney DF,


Twisk D. Anthropometric Data Of Children For the Development of Child Dummies. TNO; 1994. TNO 75061275-B.


Appendix A: MATLAB Script
Figure 22: Pseudocode for raw data processing of shoulder F-D data.
Shoulder Raw Data Processing - Posteromedial Loading
Written by Meagan Ita 8.27.13, Updated 3.25.14

close all
clear all
clc
date

% Inputs required here
FileNameR = 'P24pm11_R.txt';
FileName2080 = 'P24pm11_2080.txt';
FileNameFull = 'P24pm11_F.txt';

ans =

25-Mar-2014

Load Q3s data
For each time you run this script for the 12 Vicon tests you need to update the following fields: 1. Vicon input file 2. G5 input file 3. G5 input A column cell (takes care of time phase difference)

% Vicon input trial #, G5 input file, 'A' column cell
% 1, Q3ob1N001, 1675
% 2, Q3ob1N002, 1622
% 3, Q3ob1N003, 1579
% 4, Q3latN001, 1637
% 5, Q3latN002, 1577
% 6, Q3latN003, 1501

% Inputs required here
% ViconFile = 'Trial03.xls';
% G5File = 'Q3ob1N003_DIADEMHeader_BINARY.xls';
% Range = 'A1579:111003';

Load volunteer data
Loads data from .mat file, load cell and EMG data (analog).

% Inputs required here
load 'P24_P24pm11_.mat';
analog = dlmread('P24pm11.txt',',',2,0);
% Manually input time 'start' at beginning of the response curve
start = 70;

**Input Q3s Vicon and G5 data**

```matlab
% Vicon = xlsread(ViconFile,'A5:V974');
% % All position data inputted in millimeters
% % Inputs all data and then zeroed
% % Time = Vicon(:,1)./100; % Sampling rate = 100 Hz, vector is in sec
% RightX = Vicon(:,2) - Vicon(1,2);
% RightY = Vicon(:,3) - Vicon(1,3);
% RightZ = Vicon(:,4) - Vicon(1,4);
% LeftX = Vicon(:,5) - Vicon(1,5);
% LeftY = Vicon(:,6) - Vicon(1,6);
% LeftZ = Vicon(:,7) - Vicon(1,7);
% SternumX = Vicon(:,11) - Vicon(1,11);
% SternumY = Vicon(:,12) - Vicon(1,12);
% SternumZ = Vicon(:,13) - Vicon(1,13);
% % % Input Q3s G5 data
% G5 = xlsread(G5File,Range);
% Ram = (G5(:,3)-G5(1,3)).*4.448.*1; % N, converted from lbs
% Linear_Pot = (G5(:,7)-G5(1,7)).*1; % mm
% G5_Time = G5(:,2) - G5(1,2);
```

**Input volunteer data**

```matlab
% The variable data is a multidimensional array. Impacted acromion is the
% first "page", sternum/manubrium is the third, opposite acromion is the
% fourth. The following three lines pulls the 2D matrix elements out. (For
% some of the volunteer trials, the Iacr, Manu, and Oacr are 9,11,12. To
% figure this out, have to check original script files from 2011).
Iacr = data(:,:,1);
Manu = data(:,:,3);
Oacr = data(:,:,4);

% when marker variables are assigned they are rows, whereas Q3s marker
% variables are columns. After row variables are assigned and zeroed they
% are transposed into column vectors.
RightX = Iacr(1,:) - Iacr(1,1); RightX = RightX';
RightY = Iacr(2,:) - Iacr(2,1); RightY = RightY';
RightZ = Iacr(3,:) - Iacr(3,1); RightZ = RightZ';
LeftX = Oacr(1,:) - Oacr(1,1); LeftX = LeftX';
LeftY = Oacr(2,:) - Oacr(2,1); LeftY = LeftY';
LeftZ = Oacr(3,:) - Oacr(3,1); LeftZ = LeftZ';
```
SternumX = Manu(1,:) - Manu(1,1); SternumX = SternumX';
SternumY = Manu(2,:) - Manu(2,1); SternumY = SternumY';
SternumZ = Manu(3,:) - Manu(3,1); SternumZ = SternumZ';
Ram = (analog(:,8) - analog(1,8)).*10/1.*4.448; % Ram is already a column so doesn't need transposed.

Filter data

% Bandpass Butterworth filter 10-350Hz
[b,a] = butter(6, 100/500, 'low');

RightX = filtfilt(b,a,RightX); RightY = filtfilt(b,a,RightY); RightZ = filtfilt(b,a,RightZ);
SternumX = filtfilt(b,a,SternumX); SternumY = filtfilt(b,a,SternumY); SternumZ = filtfilt(b,a,SternumZ);
LeftX = filtfilt(b,a,LeftX); LeftY = filtfilt(b,a,LeftY); LeftZ = filtfilt(b,a,LeftZ);
Ram = filtfilt(b,a,Ram);

Calculate tilt

Truncate data whenever 4 degrees of subject tilt is observed Add values to the tiltangle vector as long as the angle is less than 4 degrees AND we haven't exceeded the length of the Vicon data vectors Start this calculation from the point after the loading response has been designated to start.

% with volunteer variables
initialangle = atand((Oacr(3,start)-Manu(3,start))/(Oacr(2,start)-Manu(2,start)));
tiltangle = 0;
s = 1;
z = start + s;
while tiltangle < 4 & z < length(Manu)
    tiltangle(s) = initialangle - atand((Oacr(3,s+start)-Manu(3,s+start))/(Oacr(2,s+start)-Manu(2,s+start)));
s = s+1;
z = start + s;
end

% % with Q3s variables
% initialangle = atand(Vicon(start,7)-Vicon(start,13))/(Vicon(start,6)-Vicon(start,12));
% tiltangle = 0;
% s = 1;
% z = start + s;
% while tiltangle < 4 & z < length(Vicon(:,11))
%     tiltangle(s) = initialangle - atand((Vicon(s+start,7)-Vicon(s+start,13))/(Vicon(s+start,6)-Vicon(s+start,12)));
%     s = s+1;
% end
% z = start + s;
% end

**Downsample force data**
Downsample the ram force to match the sampling rate of the Vicon data.

```matlab
Ram = downsample(Ram,10);
Ram_Full = Ram; % Use this to plot full data set if desired.
```

**Calculate deflections**

% Deflections used for Lateral Trials
Half_Girdle_Y = (RightY - SternumY);
Full_Girdle_Y = (RightY - LeftY);

% Additional Deflections used for Oblique Trials
Half_Girdle_X = abs(RightX - SternumX);
Half_Girdle_XY = sqrt((RightX-SternumX).^2 + (RightY-SternumY).^2);

Full_Girdle_X = abs(RightX - LeftX);
Full_Girdle_XY = sqrt((RightX-LeftX).^2 + (RightY-LeftY).^2);

% z exceeded the length(Ram) in a few instances because sometimes the
% downsmapled Ram vector dropped the last value. This if loop takes care of
% this.
if z>length(Ram)
    z = length(Ram);
end

% If all data is "viable" (no tilt exhibited) then these commands won't
% change any of the vectors.
Half_Y = Half_Girdle_Y(start:z)-Half_Girdle_Y(start);
Full_Y = Full_Girdle_Y(start:z)-Full_Girdle_Y(start);
Half_X = Half_Girdle_X(start:z)-Half_Girdle_X(start);
Full_X = Full_Girdle_X(start:z)-Full_Girdle_X(start);
Half_XY = Half_Girdle_XY(start:z)-Half_Girdle_XY(start);
Full_XY = Full_Girdle_XY(start:z)-Full_Girdle_XY(start);
Ram = Ram(start:z)-Ram(start);
```

**Calculate 20-80% of maximum displacement**
Find peak force after the start of the response. Cut the data at 20% to 80% of peak force. The while loops find the points in time where 20% and 80% maximum displacement occurred. These values may need to be manually adjusted if large nonlinearities are encompassed in the 20-80% range.
[maxforce,indx_maxforce] = max(Ram);
force80 = .80*maxforce;
force20 = .20*maxforce;

indx80=0;
i=1;
while indx80 == 0
    x80 = Ram(i);
    if x80>(force80)
        indx80 = i;
    else
        i=i+1;
    end
end
clear i
indx20=0;
i=1;
while indx20 == 0
    x20 = Ram(i);
    if x20>(force20)
        indx20 = i;
    else
        i=i+1;
    end
end

% From these loops, our counters of interest are
indx80
indx20

indx80 =
  184

indx20 =
  38

Create plot files

% For oblique trials
FileR = [Half_XY(1:indx_maxforce),Full_XY(1:indx_maxforce),Ram(1:indx_maxforce)];
File2080 = [Half_XY(indx20:indx80),Full_XY(indx20:indx80),Ram(indx20:indx80)];
FileF = [Half_Girdle_XY, Full_Girdle_XY, Ram_Full];
Create plots for viewing data

```matlab
figure
plot(Half_Girdle_XY,'b')
hold on
plot(Full_Girdle_XY,'r')
plot(Ram_Full,'g')
title('Time Histories')
legend('Half-XY','Full-XY','Ram')
xlabel('Time(sec)')
ylabel('Displacement(mm),Force(N)')
hold off

figure
plot(Half_Girdle_XY,Ram_Full,'b',Full_Girdle_XY,Ram_Full,'r')
legend('HalfXY','FullXY')
xlabel('Displacement(mm)')
ylabel('Force(N)')

figure
plot(Half_XY(1:indx_maxforce),Ram(1:indx_maxforce),'r',Full_XY(1:indx_maxforce),Ram(1:indx_maxforce),'b');
hold on
plot(Half_XY(indx20:indx80),Ram(indx20:indx80),'r',Full_XY(indx20:indx80),Ram(indx20:indx80),'b','LineWidth',2.5);
legend('HalfXY2080','FullXY2080')
xlabel('Displacement(mm)')
ylabel('Force(N)')
```

*Published with MATLAB® R2013b*
Appendix B: Processed Force-Deflection Curves
Figure 23: Subject 15 processed F-D curves for relaxed and tensed test conditions.
Figure 24: Subject 20 processed F-D curves for relaxed and tensed test conditions.
Figure 25: Subject 24 processed F-D curves for relaxed and tensed test conditions.
Figure 26: Subject 25 processed F-D curves for relaxed and tensed test conditions.
Figure 27: Subject 26 processed F-D curves for relaxed and tensed test conditions.
Figure 28: Subject 27 processed F-D curves for relaxed and tensed test conditions.
Figure 29: Subject 29 processed F-D curves for relaxed and tensed test conditions.


**Figure 30:** Subject 30 processed F-D curves for relaxed and tensed test conditions.
Figure 31: Subject 33 processed F-D curves for relaxed and tensed test conditions.
Figure 32: Subject 34 processed F-D curves for relaxed and tensed test conditions.
Figure 33: Subject 40 processed F-D curves for relaxed and tensed test conditions.
**Figure 34**: Subject 43 processed F-D curves for relaxed and tensed test conditions.
Figure 35: Subject 44 processed F-D curves for relaxed and tensed test conditions.
Figure 36: Subject 45 processed F-D curves for relaxed and tensed test conditions.
Figure 37: Q3s processed F-D curves.
Table 10: Stiffness values calculated from the linear portions of individual trial F-D curves. If subject interpolated mean curves exhibited total deflection less than 2 mm, they were omitted from the analyses, as shown by the shaded cells.

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Table 10: Continued
Appendix C: Interpolated Subject Mean Curves
Figure 38: Subject 15 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 39: Subject 20 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 40: Subject 24 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 41: Subject 25 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 42: Subject 26 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 43: Subject 27 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 44: Subject 29 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 45: Subject 30 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 46: Subject 33 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 47: Subject 34 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 48: Subject 40 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Subject 43 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 50: Subject 44 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Figure 51: Subject 45 interpolated mean curves for relaxed and tensed muscle conditions for full- and half-thoracic deflections. Dotted lines represent individual trials; solid lines represent the interpolated mean.
Appendix D: Subject Sled Test Responses
Figure 52: Sled pulse from event onset for all trials. Q3s plotted against individual PV curves for lateral and oblique trials.

Figure 53: Lateral displacement (y-direction) of the SSN marker for all trials. Q3s plotted against individual PV curves for lateral and oblique trials.
**Figure 54:** Torso rollout angle projected onto the transverse (x-y) plane for all trials. Q3s plotted against individual PV curves for lateral and oblique trials.

**Figure 55:** Torso rollout angle projected onto the coronal (y-z) plane. Q3s plotted against individual PV curves for lateral and oblique trials.
**Figure 56:** Kinematic trajectories of markers on the top of the head (HT), C4, and T1 in the sagittal (x-z) plane for lateral and oblique trials. Q3s plotted against PV mean curves for lateral and oblique trials. Trajectories were adjusted to begin at the average initial (x,z) point. Rectangles indicate one standard deviation of initial (x,z) starting point.

**Table 11:** Maximum excursion and time at maximum for ΔX trajectories. Shaded cells indicate a significant difference between the Q3s and PVs at significance level of .05.

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<th>Q3s</th>
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<td>34</td>
<td>43 ± 5</td>
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<td>37 ± 10</td>
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<tr>
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<tr>
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<td>17</td>
<td>29 ± 8</td>
<td>20 - 37</td>
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Figure 57: Kinematic trajectories of markers on the top of the head (HT), C4, and T1 in the transverse (x-y) plane for lateral and oblique trials. Q3s plotted against PV mean curves for lateral and oblique trials. Trajectories were adjusted to begin at the average initial (x,y) point. Rectangles indicate one standard deviation of initial (x,y) starting point.
Appendix E: Non-Normalized Sled Test Responses
**Figure 58:** Non-normalized lateral displacement (y-direction) of the SSN marker for all trials. Q3s plotted against individual PV curves for lateral and oblique trials.

**Figure 59:** Non-normalized lateral displacement (y-direction) of the SSN marker. Q3s plotted against PV mean curves for lateral and oblique trials.
Figure 60: Non-normalized kinematic trajectories of markers on the top of the head (HT), C4, and T1 in the coronal (y-z) plane (left). HT and C4 trajectories relative to T1 (right). Q3s plotted against PV mean curves for lateral trials. Trajectories were adjusted to begin at the average initial (y,z) point. Rectangles indicate one standard deviation of initial (y,z) starting point.

Figure 61: Non-normalized kinematic trajectories of markers on the top of the head (HT), C4, and T1 in the coronal (y-z) plane (left). HT and C4 trajectories relative to T1 (right). Q3s plotted against PV mean curves for oblique trials. Trajectories were adjusted to begin at the average initial (y,z) point. Rectangles indicate one standard deviation of initial (y,z) starting point.
Figure 62: Non-normalized kinematic trajectories of markers on the top of the head (HT), C4, and T1 in the sagittal (x-z) plane for lateral and oblique trials. Q3s plotted against PV mean curves for lateral and oblique trials. Trajectories were adjusted to begin at the average initial (x,z) point. Rectangles indicate one standard deviation of initial (x,z) starting point.

Figure 63: Kinematic trajectories of markers on the top of the head (HT), C4, and T1 in the transverse (x-y) plane for lateral and oblique trials. Q3s plotted against PV mean curves for lateral and oblique trials. Trajectories were adjusted to begin at the average initial (x,y) point. Rectangles indicate one standard deviation of initial (x,y) starting point.